Wireless Physiological Sensor System for Ambulatory Use

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Abstract

A wearable Physiological Sensor System (PSS) has been built that can be used with a wireless control unit to monitor the physiological status of a mobile user. Two physiological variables, heart rate and respiratory rate, were selected for initial study. The PSS was required to be non-invasive, lightweight, and low-power. Initial studies indicated that heart rate could be effectively measured using infrared light technology and so this was implemented in a wrist strap. Respiratory rate was recorded using a novel implementation of conducting polymer strain gages. For this purpose, polypyrrole-coated Lycra was fabricated and a chest strap was made that incorporated this sensor. Both sensors were connected to the PSS unit that was designed to provide the necessary signal processing; the output signals were then analyzed by the microcontroller in the control unit that is used to activate a wearable tactile display.

1. Introduction

A range of technologies have been developed for monitoring physiological variables that have been integrated with wireless communication methods. These wireless sensor systems are used in a variety of medical, military and sports settings where they provide valuable information about potential emergencies, intervention requirements and injury status [1]. In general these devices have been designed to be noninvasive, lightweight and easily worn so that the wearer's normal activity is not impeded. The range of variables that has been measured includes heart rate, pulse waveform, respiratory rate, physical activity, body core temperature, and blood oxygen saturation.

Heart rate can be monitored using a variety of methods including wireless chest straps that house electrodes to measure the electrocardiogram (ECG), ultrasound Doppler, and pulse oximeters which use the differential absorbency of light through the capillary bed to measure the changes in arterial blood volume associated with each heart beat [2]. The amplitude of the pulse oximetry signal depends on the amount of blood pulsing through the vascular bed, the optical absorption of the blood, skin and tissue and the particular wavelength used to illuminate the blood.

Respiratory rate monitors include those that use direct measurements of air flow and lung volume such as spirometers and nasal thermocouples, and those that use indirect methods such as pneumatic respiration transducers, gas pressure belt sensors, transthoracic inductance and impedance plethysmographs, and strain gage measurements of thoracic circumference [3]. Direct measurements are typically the most accurate, but are more intrusive and interfere with normal respiration. The main method suitable for respiratory rate monitoring that does not impede the normal activities of the individual is a transducer belt based on strain gages. These sensors respond to changes in length, due to an applied force, with changes in resistance. The bridge circuit and applied voltage used in conjunction with strain gages enables small changes in resistance to be recorded by measuring the voltage changes. As the resistance value pulses, the voltage will also pulse, thereby allowing the respiratory rate to be measured.

1.1. Motivation

The goal of this project was to design and build a low-power, non-invasive, light-weight, wireless Physiological Sensor System (PSS) to measure heart rate and respiratory rate. The main factors considered in the design of this system were power and energy efficiency, non-invasive sensing methods, and wearable sensor harnesses that were comfortable,

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unobtrusive, and positioned so as to not hinder limb movements nor impede normal hand use. The PSS is to be used in conjunction with a Wireless Tactile Control Unit (WTCU) developed in the laboratory [4], which receives navigation commands from a host computer that are then transmitted to an array of vibration motors (tactile display) worn around the torso [5].

To meet the requirements of monitoring physiological signals in healthy individuals and interfacing this system with the WTCU, the wearable sensors and signal conditioning units were designed and built in-house. The technology selected for recording heart rate was IR Light Monitoring, and respiratory rate was measured using a novel application of conducting polymers which were fabricated to function as strain gages. The heart rate sensor was implemented in a wrist strap so as to not interfere with the functionality of the hands and the respiratory rate sensor was implemented in a chest strap.

2. General methods

2.1. Sensors

Sensors based on IR light monitoring measure the changes in arterial blood volume that are associated with each heart beat. The sensor system is noninvasive as it detects the levels of IR light reflected by the skin and only requires that an IR emitter and detector (a phototransistor) be placed near the skin. With this method, infrared light is directed to a localized area of skin, usually on the finger, and an IR light detector is placed on either the other side of the finger or a small distance along the finger in the direction of blood flow. The detector measures the amount of IR light that is absorbed which varies with blood volume. Current commercial systems using IR monitoring technology apply the sensors directly on the fingertip or earlobe for measuring heart rate. Due to the environments it which it was envisaged that the PSS would be used (outdoors, rugged terrain), neither of these sites were appropriate for locating the sensor.

A prototype signal processing circuit was constructed and a number of different IR emitters and detectors were evaluated by mounting them directly on the skin at various sites on the body. Based on the amplitude of the signal recorded and the consistency in measuring a signal, the wrist was selected as the optimal location for this application. The IR emitter and detector (Digikey 365-1043 and 365-1070 respectively) were positioned 20 mm apart and glued between two layers of neoprene in a wrist strap with Velcro enclosures. The wrist pulse signal was amplified and bandpass filtered to attenuate baseline drift and movement artifacts. A block diagram of the heart rate signal processing is shown in Figure 1. The conditioned heart pulse signal from which a digital signal was created and used to measure heart rate is shown in Figure 2.

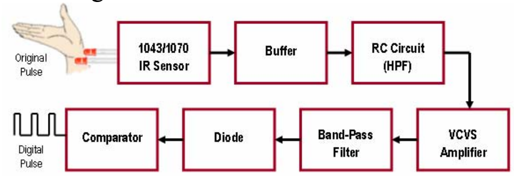

Figure 1. Signal conditioning of the heart rate sensor output to produce a digital pulse signal

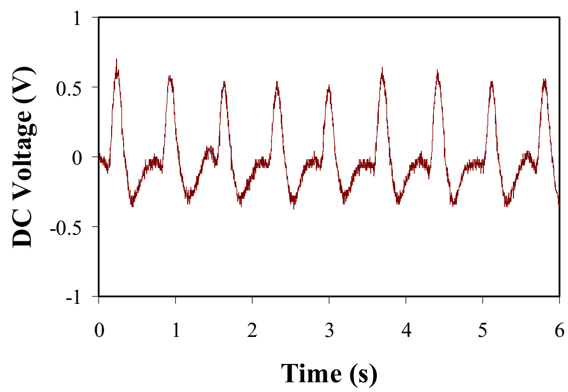

Figure 2. Band-pass filtered output of the IR sensor applied to the wrist.

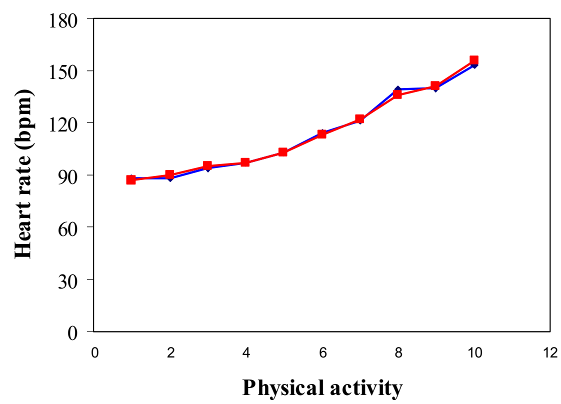

Figure 3. Heart rate measured with the Polar M32 (blue) and the PSS sensor (red) during a range of physical activities from walking to running up stairs.

The performance of the PSS heart rate sensor was compared to that of the Polar M32 heart rate monitor to evaluate its validity during a range of physical activities (e.g., walking, climbing stairs). To calculate the PSS heart rate (in bpm) the times between the rising edges of the five most recent digital pulses were averaged and the inverse of the average time per pulse was calculated and then multiplied by 60. The PSS heart rate was rounded to the nearest integer to be comparable to the Polar M32 readings. The results are shown in Figure 3 where it can be seen that the average difference in heart rate (bpm) recorded from the two devices was extremely small (1%), indicating that the PSS heart rate sensor is as accurate as this commercial system.

Strain gage technology was the main method pursued for measuring respiratory rate. Rigid strain gages were ineffective for measuring the displacement of the thoracic cavity as it expands and contracts during respiration as they had to be mounted in a relatively inflexible strap which did not permit normal expansion of the chest. Elastic fabrics, such as Lycra, coated with a layer of conducting polymer have been used as flexible strain gages [6]. By chemically and then electrochemically depositing pyrrole on a strip of stretchable Lycra, a conducting PPy film can be created that allows the Lycra to act as a strain gage [6-8]. Although conducting polymer strain gages are still a relatively immature technology, a novel application of PPy Lycra strain gages explored in this project was their use for measuring respiratory rate.

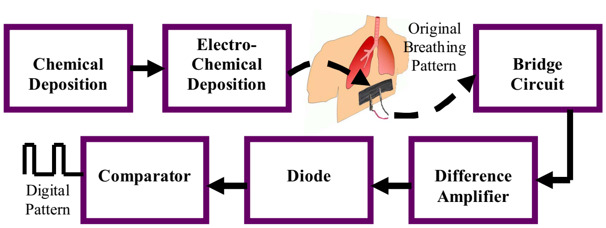

Figure 4. Fabrication and application of PPy Lycra strain gages as respiratory rate sensors

Polypyrrole Lycra strain gages were created in a two-stage process in which the conducting polymer was first chemically deposited onto the Lycra fabric and then an electrochemical deposition was performed to improve the conductivity of the polymer coating. The fabricated strain gages were then stretched and measurements made of the changes in resistance. Although there was considerable variability in the measured resistances between different PPy Lycra strips (180-3000 ohms), the initial resistance for a single strip (i.e .strain gage) remained within a relatively narrow range of values (e.g. 180-200 ohms) with repeated testing. The gage factor, a value that specifies the relation between a strain gage's resistance and its strain, for the fabricated PPy Lycra strain gages was generally between 1 and 2.

It is not necessary to know precisely how much the strain gage is being stretched in order to measure respiratory rate. All that is required is that the changes in resistance are sufficient to trigger the system to record the rate with which the thoracic cavity is expanding and contracting during breathing. The circuitry that was designed for the PSS respiratory rate sensor took into consideration the variability between different PPy Lycra strain gages and focused on creating a pulsating signal that corresponded to the breathing pattern. A Wheatstone bridge circuit was implemented to capture the changes in resistance and the output of the difference amplifier then went into comparator circuitry similar to that used for the heart rate sensor. The resulting output digital signal was used to calculate the respiratory rate. A block diagram of the respiratory rate sensing process is illustrated in Figure 4 and the signals recorded from the two sensors concurrently are shown in Figure 5.

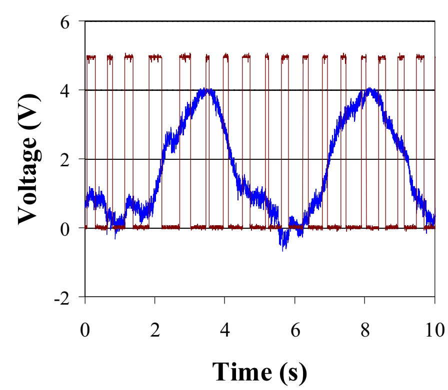

Figure 5. Amplified respiration signal (blue) and concurrent digital heart pulse (red) signal

2.2 Wireless Tactile Control Unit and PSS Unit

The WTCU was designed to control a tactile display comprising a 4-by-4 array of vibrating motors positioned on the lower back. The motors are activated in different spatial patterns, with each pattern associated with a unique navigational or informational instruction. A notebook computer is used to transmit these commands using Bluetooth technology [4]. The WTCU is worn by the user, and the embedded processor in the WTCU receives commands from the host and translates these into patterns of motor actuation. The microcontroller (ATMEGA8535L) has a USART interface, a reasonably fast clock speed, low-power sleep modes, a watchdog timer, large Flash memory space, 16-bit timers for long delays and timing events, and a number of I/O pins for motor control and sensor interfacing. The PSS makes use of the open pins on the microcontroller for analyzing the physiological signals processed by the PSS. Only two interrupt pins on the microcontroller are used, one in conjunction with the heart rate sensor and the other for the respiratory rate sensor. Most of the signal processing is done in analog on the PSS board which converts the AC components of the heart rate and respiratory rate signals into digital signals that are sent to the microcontroller on the WTCU. An algorithm determines the heart rate or respiratory rate from the received digital signal, and the result is transmitted wirelessly to a supervisory agent, who monitors the status of the individual. The wireless component of the PSS is connected to the WTCU using the same 2.4 GHz Bluetooth Class 1 device which communicates wirelessly with the host computer.

The software required for the PSS is written in assembly language for the microcontroller and in Microsoft Visual Basic .NET for the GUI run on the notebook computer. The assembly program for the PSS sensors finds and returns the average time between the heart or respiratory pulses.

3. Discussion

Two sensors have been fabricated together with the signal processing unit to measure heart rate and respiratory rate in ambulatory individuals. The respiratory rate sensor is based on a novel application of conducting polymer strain gages. The PSS is to be used in conjunction with a torso-based tactile display (WTCU). With the wireless communication technology implemented at present on the WTCU (2.4 GHz Bluetooth), the range of this system outdoors is 100 m, and indoors around 30 m.

One challenge in designing and fabricating wearable sensors for ambulatory use in rugged environments is to ensure that the sensor remains in contact with the body as the individual moves around, but does not impede the wearer's movements. Movement of the IR heart rate sensor on the skin does create high frequency peaks in the heart pulse signal and although band-pass filtering has been implemented to eliminate high frequency noise it does not fully compensate for rapid body movements. Further digital signal processing of the heart rate signal will be implemented that should improve the sensor's performance during movement.

The PPy Lycra strain gage respiratory sensor was sewn into a soft but non-flexible chest strap that was worn circumferentially around the chest to measure breathing rate. The properties of the PPy Lycra strain gage changed over time probably due to solvent loss and oxidation by the environment. Further isolation of the strain gage from the environment and development of a better coating may extend its life.

The PSS and WTCU together provide a wireless system for monitoring and communicating with a person navigating through an environment. The system is non invasive, low power and light weight and has the potential to be expanded to incorporate a broader range of body sensors.

Acknowledgements

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