Intelligent In-House Monitoring for the Elderly

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Abstract- **Heat emergencies such as heatstroke, high fever and occupational heat exposure are all serious medical conditions that require emergency response. These are of particular concern for the elderly or infirmed, who may be home alone and unable to summon help. We present a paper on an Intelligent In-House Monitoring system for the elderly, designed to collect both physiological and environmental data in order to determine if an individual is experiencing a heat emergency. This system was designed to non-intrusively and non-invasively measure pulse rate, perspiration and body temperature to determine whether or not a person is experiencing some sort of heat emergency. The monitoring system is comprised of a wearable monitor that transmits the information wirelessly to a base station that records both ambient air temperature and humidity. The base station combines the data from the monitor with the environmental data to determine if the wearer is experiencing a heat emergency. If it determines that a heat emergency is occurring then it will attempt to reach an emergency contact, as well as warn the wearer of the need to cool down.**

I. Introduction

Heat emergencies, such as heat stroke, are serious medical conditions that require immediate medical attention. The elderly and the infirmed are at particular risk since they may live alone and be unable to summon help. Because of this, it is desirable to devise some sort of system that will be able to determine if a person is experiencing a heat emergency and then summon help, as well as warn the person to try to cool down. A heat emergency occurs when the body is unable to regulate its temperature due to excessive heat exposure. Under normal circumstances, the average body temperature tends to be approximately 37°C. When the ambient air temperature increases, the body attempts to cool itself by perspiration and dilating the blood vessels under the skin so that the heat can escape via convection and radiation. After a period of time, dehydration will cause perspiration to cease. At this point, the core body temperature will begin to rise very quickly and the person can quickly enter a state of heat hyperpyrexia. The body temperature at this point is typically 41°C. The pulse rate will also begin to rise, reaching in excess of 130 beats per minute. If the person does not seek immediate medical attention, they will die once their body temperature exceeds 42°C [5].

In the case of the elderly and the infirmed there are several options to preventing a heat emergency, such as hiring a helper, moving into a nursing home or purchasing a central air-conditioning system. Unfortunately, these options may not always be desirable since there is either some loss of independence or it may be financially prohibitive. For these reasons, a simple, easy to use, unobtrusive and inexpensive monitoring system may be the best option.

II. System Description

The In-Home monitoring system consists of two major parts: the monitoring unit and base-station unit.

1) *The Monitoring Unit*: This unit can be further broken down into three different monitoring sub-systems: A photoplethysmograph (PPG), skin conductance and temperature.

   a. Photoplethysmograph (PPG): The principle behind a PPG is when light intensity I , is traveling through a uniform medium containing an absorbent substance its intensity will decrease exponentially. This is known as Beer's Law seen in Equation 1:

$$I = I_o e^{-\varepsilon\lambda cd} \qquad (1)$$

where c is the concentration of the absorbing substance, d is optical path, $\varepsilon\lambda$ is the extinction coefficient and I_o is the intensity of the incident light. If there is a larger amount of the absorbent substance, the intensity is decreased. In the case of the human blood stream, there are several species of hemoglobin which act as the absorbent substance.

In the design of this sub-system, an Infrared Light Emitting Diode (IR LED) with a wavelength of 940 nm is used to target the oxy-hemoglobin in the blood stream. During the systolic phase (contraction of the heart), the concentration of hemoglobin is increased significantly, and as a result, the intensity of the transmitted light is lower relative to the diastolic phase (relaxation of the heart). The circuitry for the IR LED takes account two issues. The first is the intensity of the light, which is adjusted by the Cypress CY8C29466 Programmable System on a Chip (PSoC) to provide optimal measurements. This is necessary because the skin's characteristics, such as pigmentation, thickness and optical length vary from person to person. Thus, affecting the absorption of light in the tissue. The second is to improve signal noise ratio. This is done by pulse-width modulating (PWM) the IR LED at a frequency of 488 Hz with 25% a duty cycle [8].

The reflected light is then captured by the photodiode as seen in Figure 1. According to Mendelson [10], this signal consists mainly of the opacity of the skin, reflection of the bone, tissue backscatter, and amount of blood present. The

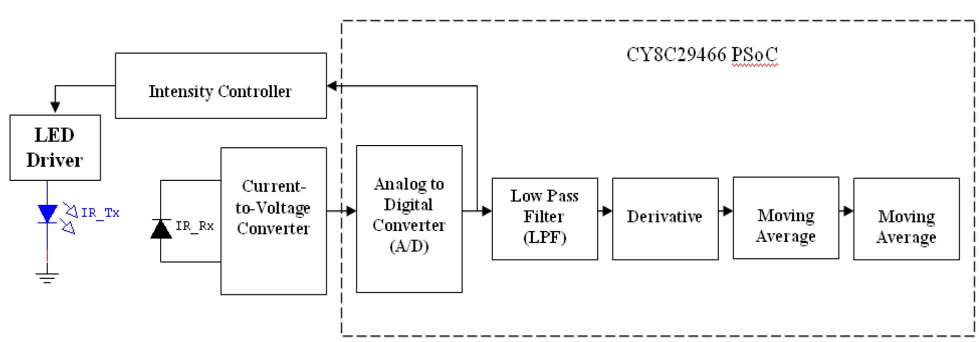

Fig. 1. System Block Diagram of the Photoplethysmograph (PPG)

intensity of this light is approximately 0.05-1 percent of the transmitted or backscattered from the tissue. The photodiode represents this signal as a current, and therefore must be converted into voltage where the A/D of the PSoC acquires the data.

b. Skin Conductance: The objective of finding the conductance of the skin is to be able to determine the amount of perspiration a person exhibits. This is done by iontophoresis. Iontophoresis refers to passing a low level DC current through the membrane of the skin. The typical resistance of the skin may range from 15 kΩ to 2 MΩ, however when moisture or sweat are present, the skin's resistance can drop to 100Ω to 200Ω.

   In the design of the skin conductance system a constant current LED driver produces low DC current for Iontophoresis. The voltage potential drop between the two Ag/AgCl electrodes is measured by using an instrumentation amplifier (AD620, Analog Devices Inc, USA). This value corresponds to the impedance of the skin.

c. Temperature Measurement: The temperature sensor is an integrated circuit (IC) device that operates by exploiting the temperature dependent characteristic of a Zener diode. The IC measures the increase in current as the temperature increases (or decrease in current as the temperature decreases). The increase in current is then translated by the IC into a voltage increase (or decrease) equal to 10mV/°K. At room temperature (approximately 25°C) the IC outputs a voltage of 2.98V which corresponds to 298°K. The voltage is then measured by a microcontroller via a 10-bit analogue to digital (A/D) converter.

2) *The Base-Station Unit:* The base station is comprised of three elements: a microcontroller, liquid crystal display (LCD) and a communication module.

a. Microcontroller: The microcontroller is responsible for managing the flow of information to the LCD display on the base station. The microcontroller is also responsible for determining whether the person has reached a state of heat emergency.

b. Liquid Crystal Display (LCD): The LCD will be located on the base station for viewing from anyone who chooses to look at it. The LCD will display the heart rate last transmitted by the wireless unit along with information on the temperature and perspiration of the body.

c. Communication Module: Its main purpose is to transmit an emergency alarm to a medical professional or guardian, in the case of a heat emergency. This element is not defined as being any specific form of communication. It is meant to have many interchangeable components such that the user may choose his or hers method of communication in the case of an emergency. The preliminary units will attempt to incorporate a Global System for Mobile Communication (GSM) modem that will send a text message in the case of a heat emergency.

III. In Vivo Experiments

Preliminary data from the monitoring system have been acquired using the system described in Figure 1. In the case of the PPG, MATLAB was first used to estimate the type of filters needed and to test heart rate algorithm. The MATLAB simulations demonstrated that the algorithm was able clean up the signal and calculate the average heart rate. The algorithm was then implemented in the PSoC. As it can be seen in Figure 2 the first stage of the signal processing is a derivative filter with the following equation:

$$y(n-1) = 0.5y(n-1) - 4[x(n) - x(n-1)] \quad (2)$$

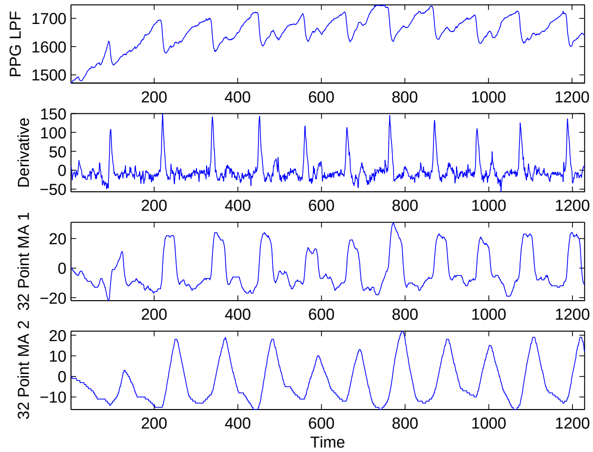

Fig. 2 Results of the signal processing of the algorithm.

By taking the derivative of the signal, the drifting DC off-set is suppressed. These offsets are often caused by motion. The next stage of signal processing is a 32 point Moving Average Finite Impulse Response (FIR) filter. The moving average filter helps smooth out the derivative output signal. However, it can be seen in Figure 2 that the pulse still exhibits some noise during its peaks. Therefore, another iteration of the moving average is applied. Once the data is filtered, an adaptive thresholding algorithm is used to estimate the signal and noise peaks in order to determine the average instantaneous heart rate. The algorithm which is implemented in this design takes account of the amplitude of a peak. This value then starts to decay at a constant rate until the next peak occurs, as seen in green in Figure 3. The hardware implementation of the filters (in blue) is very similar to the MATLAB simulation. The main difference is the first 32 points of the hardware implementation. This initial signal is generally very erratic due to the initial values in memory on the PSoC not being able to initialize properly.

IV. Conclusion

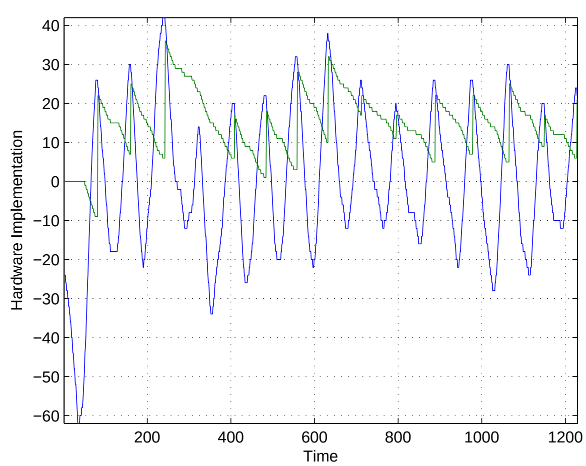

Fig. 3 Output of hardware implementation of the digital filters (Blue) and adaptive thresholding (Green)

The heart rate monitoring subsystem was implemented by designing a PPG using infrared light. A derivative and two 32-point moving averages filter were used to highlight the peaks in order for detection using an adaptive thresholding algorithm. It was shown that both the MATLAB simulation and hardware implementation illustrates that the proposed algorithm can be used to determine heart rate.

Unfortunately, at this point in time, both the temperature measurement and skin conductance subsystems have not been implemented yet. It should be noted that the design of this device is still in its preliminary stage. The end product will feature data logging of the monitored temperature, perspiration, and heart rate in intervals or in real time. We hope that this concept will be implemented in a piece of clothing. If any person who wears this device shows signs that he or she is not operating under normal conditions, then the device will contact the appropriate authorities.

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