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Portable Health Monitoring Device for Electrocardiogram and Impedance Cardiography Based on Bluetooth Low Energy

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Abstract

Portable health monitoring device could benefit the life quality of patients as well as decreasing health care costs. Monitoring Electrocardiogram (ECG) and Impedance Cardiography (ICG) simultaneously can help evaluate cardiovascular disease comprehensively. However most of the ICG monitoring devices are only used in clinical measurement instead of family monitoring. Based on the above concern, this paper designed a portable health monitoring device based on Bluetooth Low Energy (BLE) which is valued for its ultra-low power consumption for long-term continuous non-invasive cardiac activity monitoring by measuring ECG, ICG, acceleration and angular acceleration of body. There are mainly four advantages of this device: 1) High integration: ECG, ICG, acceleration and angular acceleration can be monitored in a single circuit; 2) Low power consumption: The device can monitor continuously for over 83 hours when powered by a 3.7 V lithium polymer battery with 1000 mAh; 3) Small size: The area of the PCB is only 30 mm×40 mm, so it is very easy to be attached to clothes; 4) Wireless communication: The acquired data can be transmitted to a personal computer or a smart phone wirelessly via Bluetooth. Different body postures (upright or supine) can result in change of ICG signal, which can be used for adjusting patients' hemodynamic parameters.

Keywords: Portable Health Monitoring Device; Impedance Cardiography; Electrocardiogram; Bluetooth Low Energy; Hemodynamic Parameters

1 Introduction

Cardiovascular disease is one of the leading causes of mortality in most countries. Although

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invasive techniques, such as thermodilution, give satisfactory results on the estimation of Stroke Volume (SV), they still are not quite suitable for repeated measurements or long-term monitoring due to the pain caused by invasive techniques. During the past three decades, there are several non-invasive techniques that can be used for monitoring cardiac mechanical activity, including Magnetic Resonance Imaging (MRI), ultrasound Doppler and Impedance Cardiography (ICG). Nevertheless, compared to ICG technique the other two non-invasive techniques of cardiac mechanical activity are very inconvenient and expensive, so they are not suitable for long-term portable monitoring [1]. Different from ECG (cardiac electrical activity), ICG is a novel technique capable of quantifying cardiac mechanical activity. ICG technique mainly includes direct measurements of base impedance, varying impedance, velocity index, acceleration index, pre-ejection period, left ventricular ejection time. These direct measurement parameters can be used for estimating hemodynamic parameters, such as Stroke Volume (SV) and Cardiac Output (CO) [2]. Non-invasive hemodynamic parameters monitoring by Impedance Cardiography has been applied to more than four million patients till now. Cardiac output measured by Impedance Cardiography technique has been demonstrated to correlate well with CO obtained by thermodilution with correlation coefficients ranging from 0.76 to 0.89 [3].

Bluetooth Low Energy (BLE) has many advantages over Classic Bluetooth and other wireless communication technologies. Compared to Classic Bluetooth BLE is a significant improvement, as it has low energy consumption which provides a new technical support for portable and wearable applications. The improvement of low energy consumption allows two different types of implementation modes, single mode communication and dual mode communication. Single mode implementation only supports BLE and cannot communicate directly to Classic Bluetooth. While dual mode implementation can support both Classic Bluetooth and BLE. Just like Classic Bluetooth BLE works on the same frequency range (2.402-2.480 GHz), one of the differences between them is the amount of radio channels. Different from 79 radio channels 1 MHz each of them of Classic Bluetooth, BLE only has 40 radio channels, but each of them is 2 MHz [4]. BLE's controller is also more intelligent, which can sleep for longer periods to save power if without any tasks and can be woken up only if it needs to perform an instruction. There is no doubt that BLE will be the perfect choice for portable monitoring device.

Monitoring Electrical Activity (ECG) and mechanical activity (ICG) of the heart simultaneously can help us evaluate cardiovascular disease comprehensively. Acceleration and angular acceleration also can provide movement status which is especially important for elderly people. The major accomplishments of this paper is the design of a small-sized, light-weighted, integrated and low power portable device which can provide a continuous long-term non-invasive portable monitoring for cardiac electrical activity and mechanical activity.

The remainder of this paper are divided as follows: in Section 2 we analyzed the electrical circuit modeling of tissues and then we introduced the overall design of the health monitoring device in Section 3. In Section 4 we applied filtering algorithm to remove baseline drift from ECG and ICG signal. Finally, we analyzed power consumption of the device and impact of postural change to ICG signal in Section 5 and Section 6. Then we obtained our conclusions in Section 7.

2 Electrical Circuit Modeling of Tissues

ICG technology is based on bioimpedance, so we need to know that structure of tissues with different content may have different electrical characteristics. There are several models to describe

these tissues features. Electric models most often used to characterize tissues are those containing three circuit elements (Fig. 1).

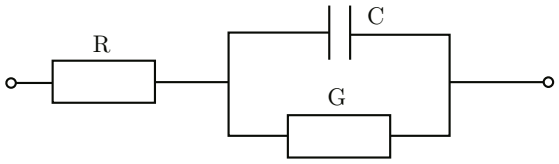


Fig. 1: Electrical circuit model of tissues

Due to the nature of tissues, impedance varies with the frequency of measurement signals; usually the impedance will decrease as frequency increases. The relationship between impedance and frequency is linear [5]. Both resistivity and permittivity are frequency dependent. The circuit is Alternating Current (AC) powered and ensures current limitation at high frequencies. It is possible to obtain the same values for all frequencies with one set of values of elements circuit components. The impedance is best characterized using a series circuit, as the time constant is uniquely defined (equations (1)-(7)). It is used as an electrical equivalent of human tissues, with a value of R representing the depth of tissue in series with the electrical representation of the superficial tissues (skin) composed of G and C in parallel.

The 2R-1C series circuit can be described by the following equation:

$$Z = R + \frac{G - j\omega C}{G^2 + \omega^2 C^2} \tag{1}$$

$$Z = R + \frac{1}{C(1 + j\omega\tau_Z)} \tag{2}$$

$$\tau_Z = \frac{C}{G}$$

$$\varphi = \arctan\left(\frac{\omega C}{G(1 + RG) + \omega^2 C^2 R}\right) \tag{3}$$

$$\varphi = \arctan\left(\frac{\omega C}{G(1 + RG + \omega^2 \tau_Z \tau_2)}\right) \tag{4}$$

$$Y = \frac{G(1 + RG) + \omega^2 C^2 R + j\omega C}{(1 + RG)^2 + \omega^2 C^2 R^2}$$
$$Y = \frac{G(1 + RG + \omega^2 \tau_Z \tau_2 + j\omega\tau_Z)}{(1 + RG)^2 + (\omega\tau_2)^2} \tag{5}$$

$$\tau_2 = CR$$

$$C_{ext} = \varepsilon' = \frac{C}{(1 + RG)^2 + (\omega\tau_2)^2} \tag{6}$$

$$\varepsilon'' = \frac{G(1 + RG + \omega^2 \tau_Z \tau_2)}{\omega[(1 + RG)^2 + (\omega\tau_2)^2]} \tag{7}$$

Impedance is an important parameter that characterizes this circuit, as defined by a single time constant τ_Z. This time constant is independent of R if the circuit is carrying current. Measured impedance parameter is in close correlation with characteristic frequency. Determination of τ_Z is directly related to capacity and a conductance (such as effects of membrane tissues) connected in parallel. The same is not valid for admittance, which is dependent on τ_Z and τ_2, and therefore

of R and G [6]. Equation 6 is of particular interest. C_{ext} terminal capacity is measured, for example, a deck or an amplifier. At high frequencies the susceptible $B = Y$ is small and C_{ext} is strongly dependent on frequency $\frac{1}{\omega^2}$. In this frequency range, high capacity increases by decreasing frequency. Human tissues may be seen as parallel conductor model containing muscle and blood. In this case, the bioimpedance relationship can be expressed by:

$$Z = \frac{L}{\sigma_m S_m + \sigma_b S_b} = \frac{L}{\sigma_m S_m + \sigma_b \frac{V_b}{L}} \tag{8}$$

where σ_m-muscle conductivity; σ_b-blood conductivity, S_m-sectional area of muscle tissue, S_b-blood vessel section area, L-measured length, V_b-blood volume. The method we use in determining the impedance is the four electrodes method (Fig. 2). This method consists of applying an alternating current I_o with high frequency between two electrodes and detected voltage V_o between the other two electrodes.

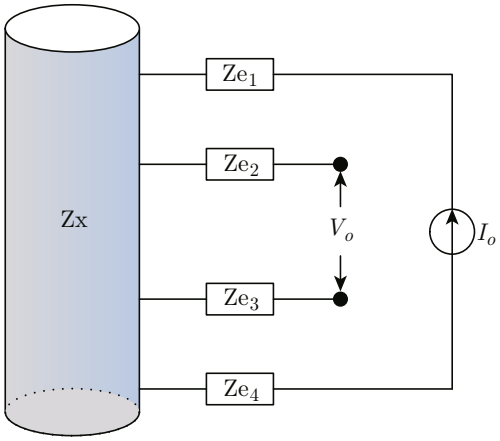


Fig. 2: Four electrodes measurement method

This tetrapolar configuration has been used for over a century and it largely removes the interference of the electrode-electrolyte interface impedance.

3 Overall Design of the Device

In order to make the device operate for a longer time, every part of the circuit should consume power as little as possible. In the design of the portable monitoring device, small size and low power consumption should be considered as two important factors. The architecture of the device is shown in Fig. 3, which mainly includes four modules: microcontroller and wireless communication module, signal acquisition module, power management module and local storage module. Power supply of each module can be turned on or off independently. This electrical isolation provides a certain degree of robustness in the event of a failure. The PCB of the health monitoring device is shown in Fig. 4. In this section, we will analyze four modules respectively.

3.1 Microcontroller and Wireless Communication Module

Each device is equipped with a Microcontroller (MCU) which acts as a local control centre for receiving and processing data, arbitrating sensor behavior, maintaining communication with the

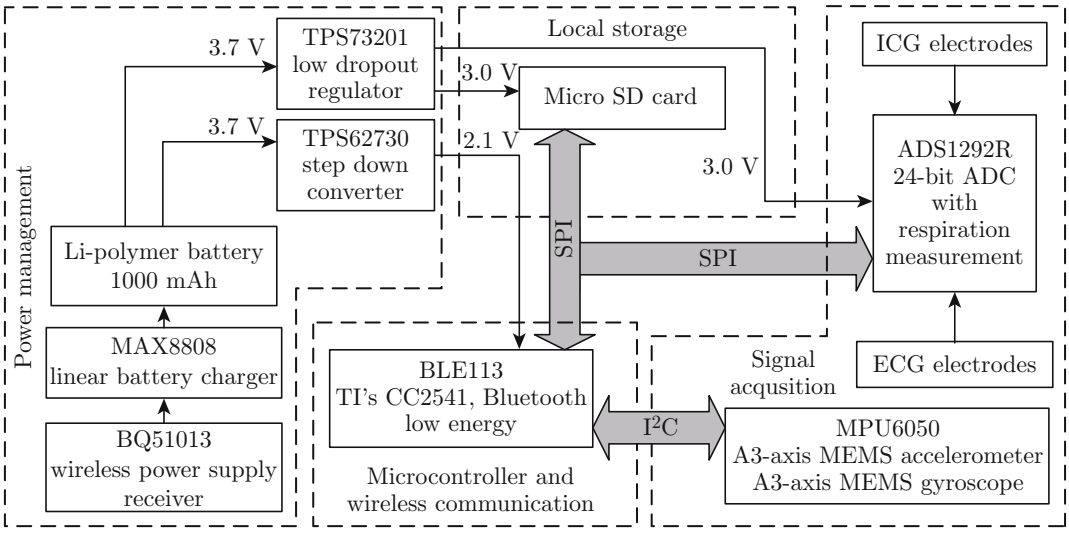


Fig. 3: The architecture of the device

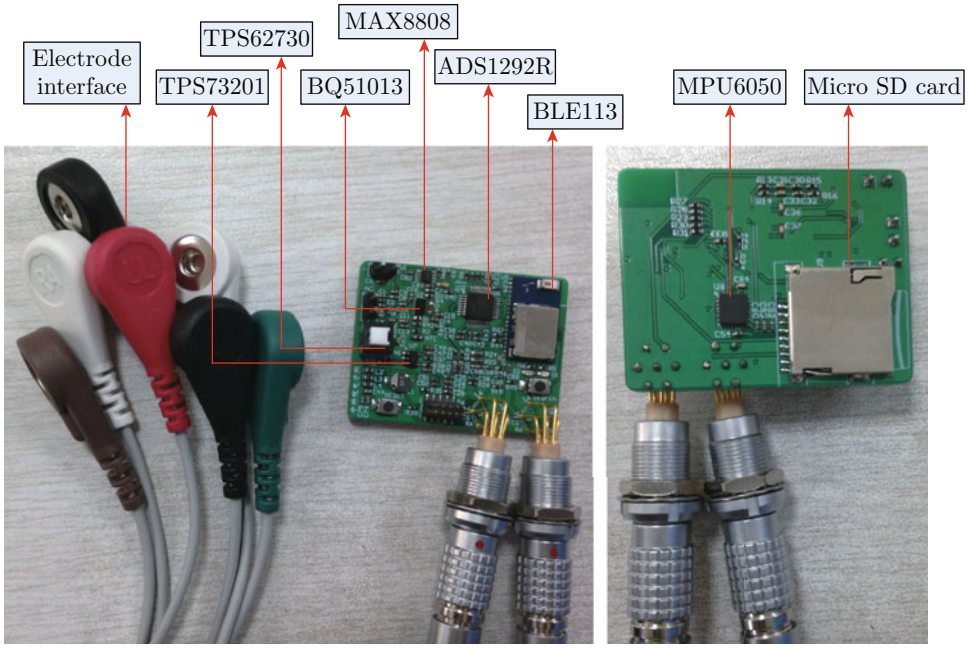


Fig. 4: PCB of the health monitoring device

wireless module, and timing events. To accomplish the lower power consumption, design principle should focus on sleep current, wakeup time of the device and energy-efficiency. It is critical to distinguish low power consumption with energy-efficiency. Low power consumption is a quality of a device that consumes low power per clock, while energy-efficiency is a ability of a device that consumes low energy per instruction. So the selection of the microcontroller is a very important for the design of a low power and miniaturized circuit. For the sake of higher integration, we also need the microcontroller to include wireless communication module.

Based on the above consideration, we chose Bluegiga's Bluetooth Low Energy module-BLE113 as the microcontroller as well as the wireless communication module in our design. BLE113 is a BLE single mode module based on TI's Bluetooth 4.0 chip CC2541 that is targeted for low

power applications. BLE113 provides all Bluetooth Low Energy characteristics: radio, stack, profiles and application space for our own design. BLE113 also offers flexible hardware interfaces to connect different sensors which make it possible to act as the microcontroller as well in our design.

BLE113 has many new features compared to classic Bluetooth technology. It supports master and slave modes, and can connect 3 slaves simultaneously in master mode. In our design, the USB dongle works in master mode and the sensor nodes work in slave mode, the USB dongle can send and receive data from three sensor nodes simultaneously, so it is possible for us to monitor 3 subjects at the same time with only one personal computer or a smart mobile phone. Another important feature is its ultra-low current consumption: 27 mA (Tx power at 0 dBm) in transmitting and only 0.4 μA in sleep mode [7].

3.2 Signal Acquisition Module

ECG and ICG are both important diagnostic parameters for a health monitoring device. ECG is an index of the cardiac electrical activity and could show the overall rhythm of the heart. ICG is an index of the mechanical activity of the heart, cardiac output and stroke volume which are very important hemodynamic parameters that can be calculated by measured ICG.

Design of the signal acquisition module is very critical and difficult to accomplish the accurate and stable measurements of ECG and ICG. The Analog to Digital Converter (ADC) is a significant component for design of the signal acquisition module whose sampling rate, resolution, power consumption, functions and package size could largely influence the overall performances of monitoring device.

We chose ADC for the circuit based on the following principles: 1) Low power consumption; 2) High resolution; 3) High integration; 4) Small package. According to the above principles, eventually we chose Texas Instruments' ADS1292R as the ADC of monitoring device. ADS1292R has two channels, simultaneous sampling, 24-bit delta-sigma ($\Delta\Sigma$) ADC with integrated programmable gain amplifiers and internal reference [8]. Moreover, ADS1292R also integrates bioimpedance measurement which can be used to monitor ICG. ADS1292R incorporates most of the characteristics that are required in the design of a small and low power portable health monitoring device. The outstanding performance and high integration of ADS1292R make the signal acquisition circuit very simple and compact, which significantly decreased the physical size and power consumption of the whole device.

The health monitoring device is designed to be hung on pants or belt. Location of the electrodes is shown in Fig. 5. Two drive electrodes (No. 4 and No. 5) on the back inject a high-frequency (64 kHz, 1.3 mA) current and two receive electrodes (No. 6 and No. 7) on the chest sense the voltage over the thorax segments. The lower current drive electrode is located at least 3 cm below the horizontal plane of the lower receive electrode, placed at the xiphoid process. The upper current drive electrode is located 3 cm above the horizontal plane of the upper receive electrode. The upper receive electrode is placed at the jugular notch of the sternum between the collar bones [9]. This electrodes placement can reduce the impact of respiratory artifacts to ICG signal to the maximum extent. Electrodes No. 1, No. 2 and No. 3 are used to monitor ECG (Lead II), and Electrode No. 2 is Right Leg Drive (RLD) electrode. The health monitoring device can acquire ECG and ICG by the seven electrodes simultaneously at the sampling rate of 250 Hz.

MPU6050 has an inner 3-axis MEMS accelerometer, a 3-axis MEMS gyroscope, and a Digital

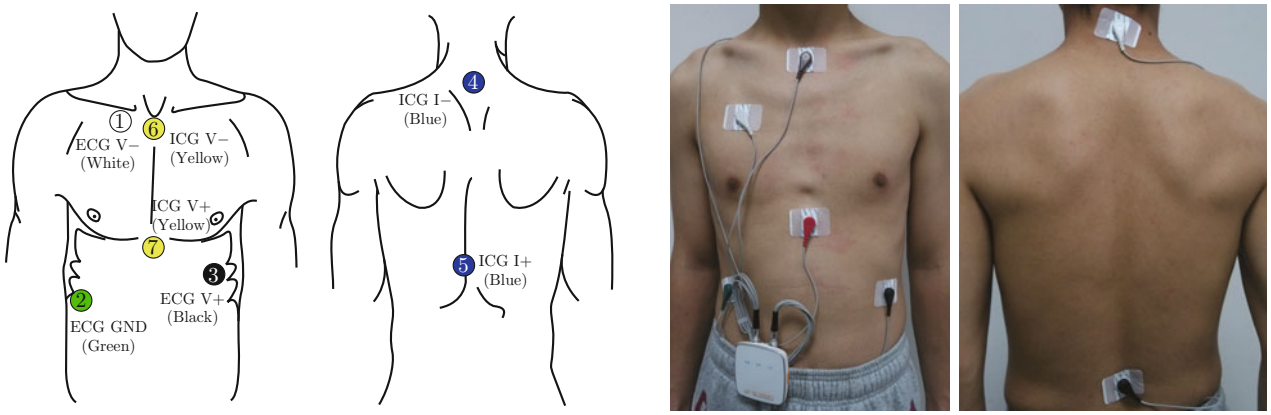


Fig. 5: Location of the electrodes

Motion Processor hardware accelerator engine with an auxiliary I^2C port that accesses to third party digital sensors. MPU6050 has three 16-bit ADCs for digitizing the accelerometer outputs and three 16-bit ADCs for digitizing the gyroscope outputs. MPU6050 can track both fast and slow motions, and it also has user-programmable full-scale range. Package size of 4 mm×4 mm improves integration for the portable health monitoring device [10].

3.3 Power Management Module

The major task of power management module is to ensure the battery voltage at a certain stable value and supply enough energy for other three modules of the monitoring device. Because the voltage conversion circuit costs some energy, we should be deliberate when choosing the voltage regulators. In this design, the power is supplied by a 3.7 V lithium polymer battery which is lightweight, compact and rechargeable. The microcontroller module is powered by 2.1 V, while the signal acquisition module is powered by 3 V. In order to decrease the power consumption and prolong the work time of the device, we decided to take the following solutions: 1) Selecting the low-power power regulators: We selected TI's TPS73201 and TPS62730 as the voltage regulators to provide suitable and stable voltage for different modules, TPS62730 has an ultra low power bypass mode with typical 30 nA current consumption to support sleep and low power modes of CC2541; 2) Incorporating a wireless charging circuit in the power management module: We selected Maxim's linear charger-MAX8808 with programmable fast-charge current (1A max) and low dropout voltage (300 mV at 500 mA) as the charging module for lithium polymer battery, the device can be wirelessly recharged via TI's BQ51013 conveniently when it runs out of energy. BQ51013 is a flexible, advanced, secondary-side devices for wireless power transfer in portable and wearable applications. The BQ51013 provides the AC/DC power conversion and regulation while integrating the digital control required to comply with the Qiv1.1 communication protocol.

3.4 Local Storage Module

Micro SD card is selected as the local storage module because of its massive storage, small size, non-volatile nature and low power. MCU can read or write micro SD card via SPI interface. The micro SD cards not only provides many gigabytes of data storage, but also allows rapid data communication to a personal computer by physically removing the card and connecting it to a

USB reader. This process is facilitated by saving the data on a FAT file system, which has become a standard file format. The micro SD card with physiological information can be taken to the hospital for professional diagnosis.

4 Filtering Algorithm for Removing the Baseline Drift and Power Line Interference

Power line interference and baseline drift are two main factors degrading the ECG and ICG. More often than not, they can even affect the accurate diagnosis of heart disease such as arrhythmia and ischemia. In this section, we adopted a practical, fast and simple digital filtering algorithm to remove the baseline drift and power line interference without degrading original ECG and ICG. For power line interference and white noise, it is easy to be filtered out by a five orders Finite Impulse Response (FIR) filter. Its cut-off frequency is 15 Hz (−3 dB) and 21 Hz (−6 dB) with 50 Hz notch frequency, and delay is 2 samples.

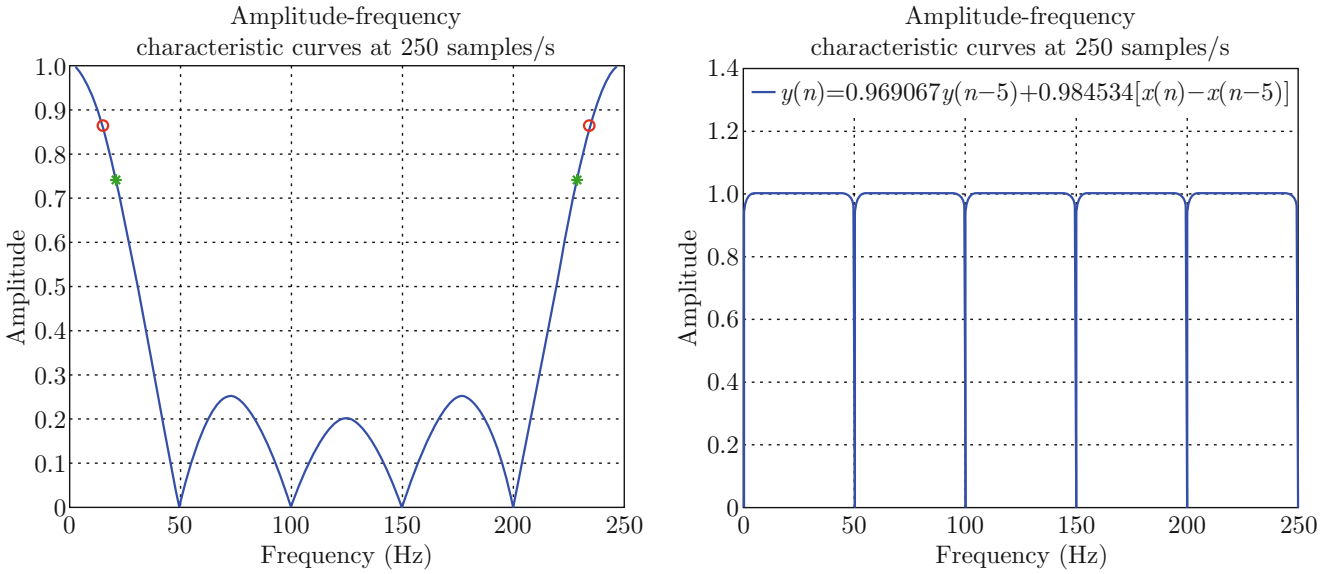
Transfer function of the FIR filter is:

$$H(z) = \frac{1}{5}(1 + z^{-1} + z^{-2} + z^{-3} + z^{-4}) \tag{9}$$

Differential equation is:

$$y(n) = \frac{1}{5}[x(n) + x(n-1) + x(n-2) + x(n-3) + x(n-4)] \tag{10}$$

Fig. 6 (a) shows amplitude-frequency characteristic curve of the five orders FIR filter. The positions in red circle represent the −3 dB point, and the positions in green star represent the −6 dB point.



(a) The five orders FIR filter

(b) The comb filter

Fig. 6: Frequency response curve

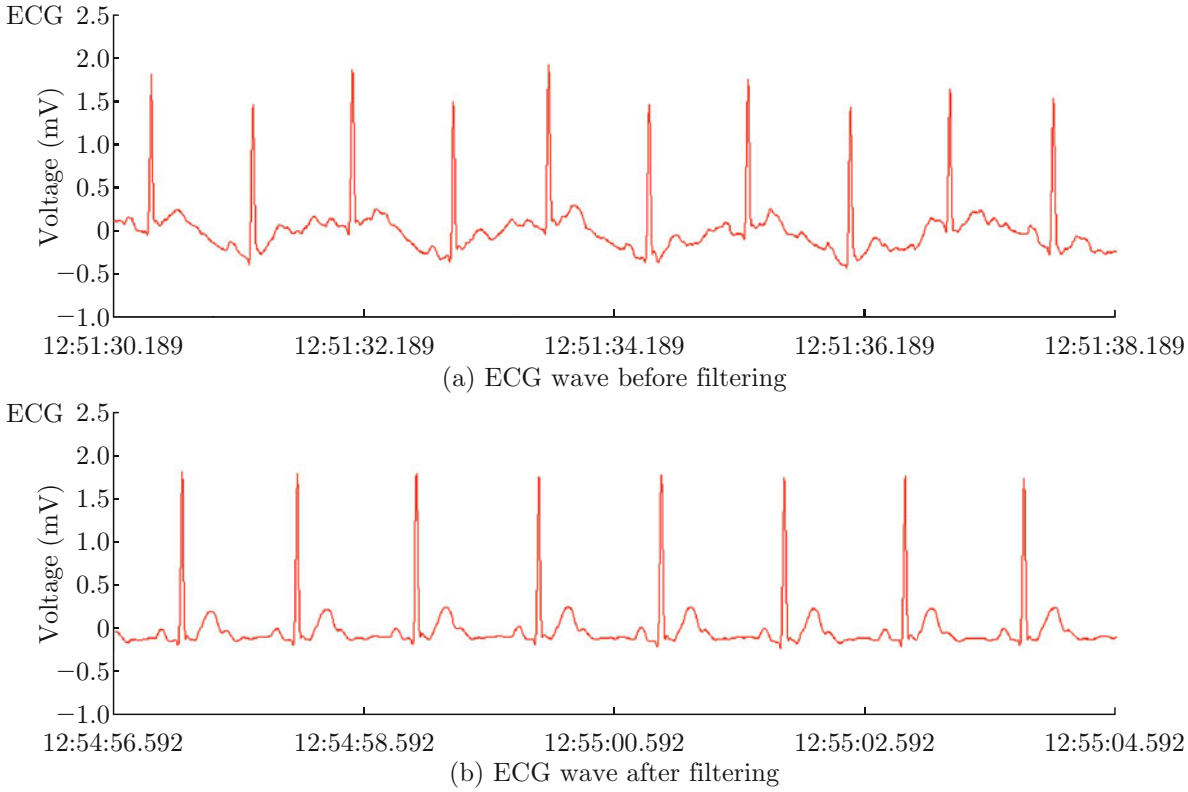
A comb filter with 0.5 Hz (−3 dB) cut-off frequency and 50 Hz notch frequency is used to remove the baseline drift, and it also can suppress power line noise. Transfer function of the comb filter is:

$$H(z) = \frac{0.984534(1 - z^{-5})}{1 - 0.969067z^{-5}} \tag{11}$$

Differential equation is:

$$y(n) = 0.969067y(n - 5) + 0.984534[x(n) - x(n - 5)] \tag{12}$$

Frequency response of the comb filter is illustrated in Fig. 6 (b). Effect of the five orders FIR filter and comb filter is illustrated in Fig. 7.



(a) ECG wave before filtering

(b) ECG wave after filtering

Fig. 7: Effect of FIR filter and comb filter

5 Power Consumption Analysis

In order to testify whether the device is low power and can continuously monitor ECG and ICG for a long term, we practically tested the current consumption of the device as it works at different modes. The current consumption is measured by a digital multimeter. The current consumption varies when the device works at different modes, and the test results of the current consumption is provided in Table 1.

According to the test results, the average current consumption of the device is 12.05 mA when operated normally, so we can draw a conclusion that the device can monitor for over 83 hours

continuously without recharging when powered by a lithium polymer battery with 1000 mAh. This completely meets the application of family health monitoring. However, there are still more problems need to be solved involving long-term stable monitoring compared with clinical health monitoring device.

Table 1: The current consumption in different working modes

Working mode					Current consumption (on average)
Standby	Advertising	Connected	Sampling	Transmit	
√					0.52 mA
√	√				1.56 mA
			√		10.08 mA
	√		√		10.67 mA
		√			10.34 mA
		√	√		11.25 mA
		√	√	√	12.05 mA

6 Impact of Postural Change to ICG Signal

The body's physiological responses to postural change are primarily a reaction to the change of gravity, because a change of gravity can cause changes of blood flow distribution in organs and tissues [11]. In order to verify that change of posture can lead to change of hemodynamic parameters, we measured ICG in supine and upright posture by using the health monitoring device.

Before monitoring ICG signal, we need to clean the skin and keep it dry to get a high-quality signal. Then we attached Ag/AgCl electrodes to the chest and back by pressing the sticky plastic brim of the electrodes on the skin and then pushing the metal stud at the center of the electrode firmly to properly spread the conductive gel. Finally, we obtained two different ICG signals in supine and upright posture. As is shown in Fig. 8, amplitude of C wave-$(dZ/dt)_{max}$ of both postures are almost in the same level, but C wave of upright posture is a little sharper than C wave of supine posture, which means Left Ventricular Ejection Time (LVET) of supine posture is a little larger than LVET of upright posture.

When body is in an upright posture, blood accumulates on the lower body and the blood volume in the brain and thorax decreases due to gravity. When the upright posture of the body is changed into the supine posture, redistribution of blood flow occurs. More venous blood of the lower body flow to the heart, which leads to atrial wall tension increasing, then results in stroke volume and cardiac output increasing. On the contrary, the redistribution of blood inside body causes a prominent reduction in the intrathoracic blood volume returning to the heart when supine posture is changed into upright posture, which leads to the reduction of stroke volume and cardiac output.

Accelerometer and gyroscope of the health monitoring devices are capable of measuring motion state of body (resting, walking or running), if body is resting, we can determine body posture by

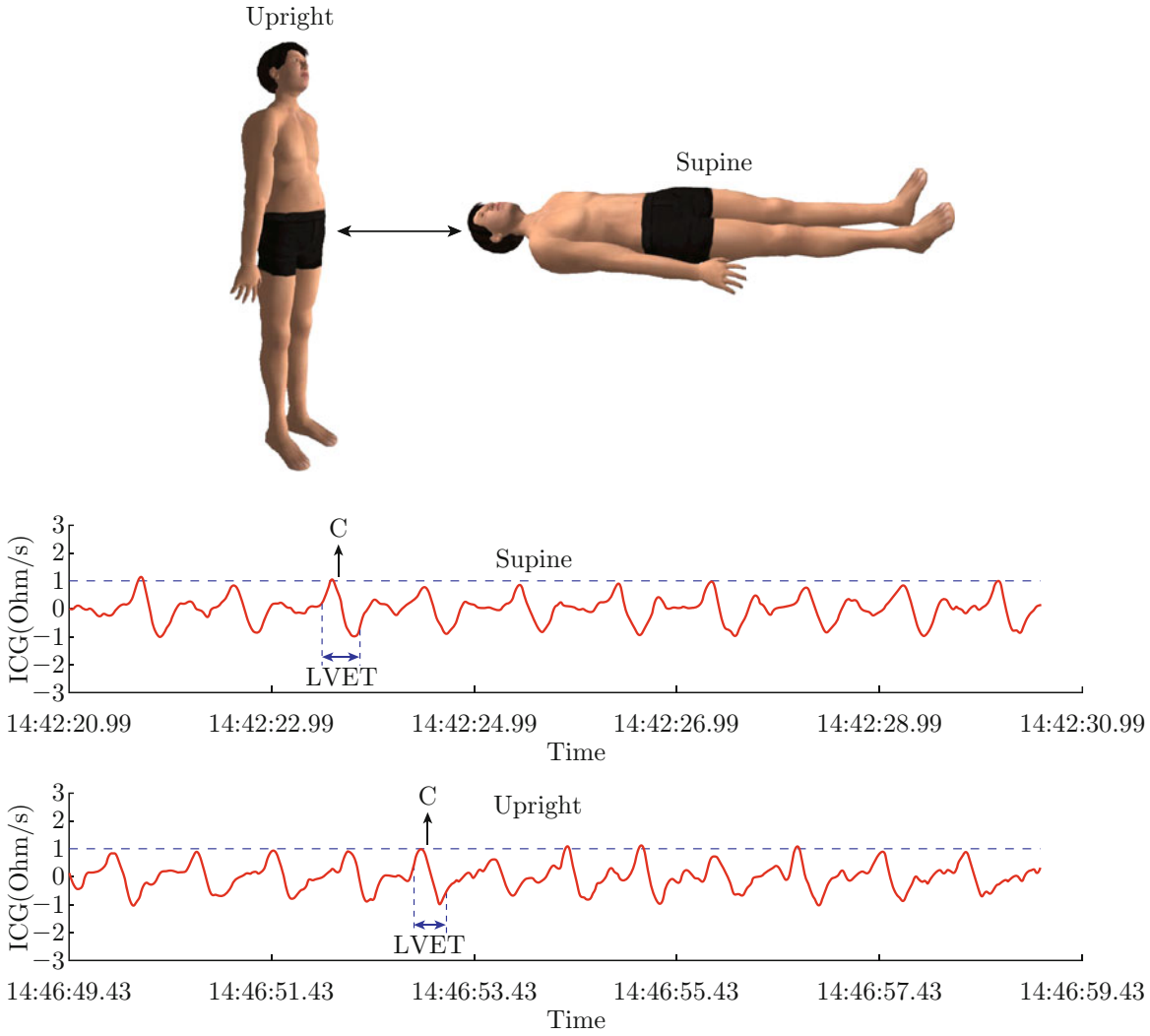


Fig. 8: The ICG waveform of healthy body in different postures

estimating cardiac output, which is very important to patients with cardiovascular disease, especially for those elderly patients who are ill in bed in hospital or at home. Patient's hemodynamic parameters can be adjusted by postural change at a certain extent.

7 Conclusions

In this paper, a low power portable device for non-invasive ECG and ICG monitoring was designed and evaluated. We mainly discussed electrical circuit model of tissues, overall design of the device and filtering algorithm applied to signal processing, analyzed the current consumption of the device when it works at different modes. Postural change can cause variation of ICG signal accordingly, which can be used for changing hemodynamic parameters by adjusting body posture. The experimental results proved that the portable health monitoring device is practical, accurate, and reliable. In future work, we plan to conduct further improvement for power consumption and apply it to different medical occasions.

Acknowledgement

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