A New Method of Measuring Impedance Cardiography for Cardiac Output Estimation by Directly Digitizing the High Frequency Modulated Signal at Lower Sampling Rate

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Abstract

Cardiac output (CO) is an important hemodynamic index to assess the heart condition of the patients. Impedance cardiography (ICG) is an advanced method that can noninvasively and continuously monitor CO based on the variation of thorax impedance. Since the variation is very small compared to the base impedance, the acquisition solutions generally require complicated analog processing circuits. This makes the output influenced by external noise, temperature, and component tolerances. This study presents a new method of measuring the changes in bioimpedance with high reliability by directly digitizing the modulated thoracic impedance signal. The proposed method has a notable advantage that allows to be implemented with low-performance hardware. The experimental results showed that the extracted data is not only similar to the reference one, but also stable over long working time. The early digitization solution makes the processing steps could be highly flexible and easy to be upgraded in further research.

Keywords: *Cardiac output*, CO, *Impedance cardiography*, ICG, Hemodynamics

1. Introduction

Cardiac output (CO) is the blood volume that the heart pumps into the aorta each minute. The assessment of CO is an important criterion in the diagnosis and treatment of diseases related to heart functioning. Numerous techniques have been investigated over the past decades to evaluate the cardiac output. These can be classified into two categories: invasive and non-invasive [1]. Impedance cardiography (ICG) has been developed strongly among of non-invasive techniques since the 1940s and becomes more and more popular because of its outstanding advantages over the others [2].

The basic principle of ICG measurement is based on the changes in thoracic impedance corresponding to blood volume changes in the thorax region during each cardiac cycle. In order to record these changes, a low intensity current source is applied to the thorax region through skin electrodes; the voltage across the thorax is simultaneously measured on other electrodes. The position of electrodes used in measuring ICG signal is followed the 8-spot electrode configuration proposed in [3], as illustrated in Fig. 1. For the reason of safety and reduction of the skin impedance, the current source applied to the human chest is normally sinusoidal with the amplitude of 1–5 mA and the frequency of 20–100 kHz [4]. Thus, the essence of the acquisition system is to extract the thoracic impedance signal from the sensed voltage for further processing.

The thoracic impedance can be divided into two components: (i) base impedance Z_0 due to the fat tissues, muscles, bones, etc.; (ii) variable impedance ΔZ due to the circulation of blood in the thorax. The spectrum of the thoracic impedance signal can spread over a range of about 0–50 Hz [5]. Normally, the base impedance range of the thorax for an adult is about 20–48 Ω over the 50–100 kHz frequency range of the current source. The variable impedance component accounts for a relatively small ratio, about 0.5%, of overall thoracic impedance [6]. This causes great challenges in designing the system that can acquire ΔZ with high reliability.

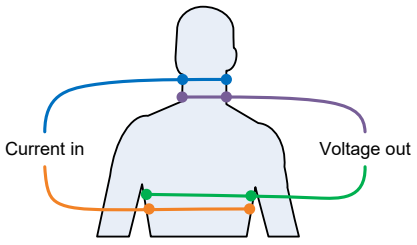

Fig. 1. Positions of the ICG electrodes.

The most informative data to calculate CO is dZ/dt, the first derivative of ΔZ, called ICG signal. The calculation of hemodynamic indices related to ICG signal is based on the identification of robust points on the ICG waveform [3, 7, 8]. Therefore, constructing an acquisition circuit able to reflect the

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impedance signal with high fidelity plays a critical role in entire systems for CO monitoring.

Currently, there are two tendencies to design the ICG acquisition systems: (i) processing signal mostly by analog circuits [9-11]; (ii) digitizing the modulated signal with a high speed analog to digital converter (ADC) and processing data with field-programmable gate array (FPGA) platforms [12]. In the last decades, the trend of using analog circuits was more prevalent. However, in recent years, digital systems have made great strides in terms of performance; the trend to apply powerful digital systems such as the FPGA into processing biomedical signals has been investigated and widely deployed [13-15].

The ICG measurement systems constructed by analog circuits typically comprise of the following modules: amplifiers, filters, amplitude demodulator, Z_0 and ΔZ separator, and analog differentiator. Hence, ensuring the stability is a great challenge when designing the acquisition systems. The operation stability strongly depends on the accuracy of electronic components, ambient temperature, external noises, and other factors. In addition, these systems could induce significant signal distortion because of using a series of analog filters such as band-pass filters in the pre-processing block, low-pass filters in the amplitude demodulator, low-pass filters and high-pass filters in the Z_0 and ΔZ separator.

An FPGA-based high performance system in combination with a high speed ADC can overcome the above-mentioned drawbacks. This scheme offers an ability to directly digitize the modulated signal without passing amplitude demodulator and post-processing stages; hence, minimize influences of the analog circuits. However, this requires the ADC able to operate with high sampling rate due to the high frequency of the carrier, and high resolution due to the low $\Delta Z/Z_0$ ratio. Thus, the ADC generates a really huge data flow, causing trouble in data transmission and signal processing.

This paper presents a new method of measuring the ICG signal for CO estimation by directly digitizing the high frequency modulated signal at much lower sampling rates. The proposed system contains a dedicated triggering module for a 16-bit ADC that only samples and holds the peak values of the modulated wave for quantization. By using the new method of envelope detection, the maximum required conversion rate of the ADC is equal to the frequency of the carrier. The actual sampling rate is even much lower because the system processes non-consecutive peaks. The lowest speed should be two times higher than the maximum frequency in the ICG spectrum. Thus, the processing load of subsequent stages could be minimized. Experimental results confirmed the proper operation of the designed circuit and the accuracy of the measured waveform, compared with the reference data. This study contributes a possibility of monitoring the advanced hemodynamic parameters with low-cost systems.

2. Method

2.1 Terminologies

In order to present the work clearly, following terminologies are defined and used in the whole paper:

- *Thoracic impedance*: represented by Z. At any time, Z can be separated into an unchanged base (Z_0) and its variation (ΔZ). In actual calculation, Z is equal to $Z_0 - \Delta Z$.
- *Carrier wave*: the high frequency current source applied to the human body for measuring Z.
- *Original signal* or *modulating signal*: the baseband wanted signal for ICG calculation.
- *Modulated wave*: the high frequency signal measured from the human body. This is also the product of Z and the applied current.

2.2 Proposed extraction mechanism

The basic idea of the proposed method is that the original signal could be recovered by digitizing the modulated signal at the peaks of the periodic wave. The output data, therefore, represent the envelope of the modulated wave and reflect the changes in the baseband signal.

This extraction mechanism is practicable and relatively optimal. First, as mentioned in Sec. 1, the frequency of the carrier wave must be high for reason of safety and reduction of skin impedance. Because of the large difference between the carrier frequency and the ICG spectrum, the frequency of the modulated wave is almost unchanged. Thus, the period of the measured signal is stable and can be precisely calculated. By utilizing a zero crossing detector and a suitable timer for time delay, a pulse can be exactly generated at each peak of the modulated wave to trigger an AD conversion, as illustrated in Fig 2. Second, the large difference between the frequencies of the carrier and baseband signal generates huge redundant data. Hence, instead of processing all peaks of the modulated wave, the proposed system samples non-consecutive peaks as long as the sampling rate is at least two times higher than the maximum frequency in the ICG spectrum. Actually, the chosen ratio is a greater value to maintain good results, with a trade-off between the signal quality and cost of the whole system.

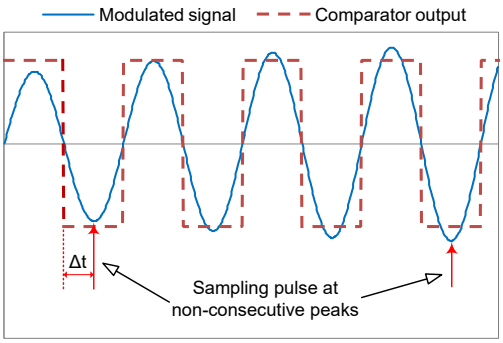

Fig. 2. Modulated signal and sampled points.

2.3 System hardware

On the basis of the proposed extraction mechanism, the system hardware consists of three major portions, as shown in Fig. 3:

- *The analog module*: includes an instrumentation amplifier for signal amplification and a high-pass filter for noise rejection. The gain of the amplifier was adjusted to get the output voltage swing of about 1.5–2 V, for the best linearity. The cutoff frequency of the filter should be low enough to reject the unwanted spectrum (e.g., DC offset, power line noise, and ECG signal) without attenuating the modulated signal.
- *The envelope detection module*: is the main contribution of this work. This module has two key components: an analog comparator and a 16-bit ADC. Here, the analog comparator is actually a zero crossing detector because its threshold voltage is set to 0 V. As mentioned in Sec. 1, hemodynamic parameters are almost calculated from ΔZ, which is hundred times smaller than Z_0. Therefore, even a small difference in measuring the modulated signal could cause a significant change in the final results. Hence, a 16-bit (or higher) ADC is required. Quantizing the modulated signal at the resolution of 16-bit is equivalent to digitize ΔZ with several hundred of quantization levels.
- *The digital processing module*: could be a commercial 32-bit microcontroller or a low-cost signal processing unit. This module has two major tasks: triggering the ADC according to the analog comparator output; processing ADC read-out to calculate the values of Z, Z_0, and ΔZ. An integrated timer is used to exactly delay 1/4 period, Δt in Fig. 2, of the modulated signal. Then, if enabled, the timer triggers the AD conversions at the peaks of the incoming signal. The processor enables AD triggering non-consecutively and steadily, as above-mentioned mechanism. Finally, the hemodynamic parameters can be mathematically estimated from the ICG signal by the microcontroller or computer software. The calculation algorithms could be found in [3, 7, 8].

3. Experimental setup

3.1 Component selections

In actual implementation, the authors chose the following configuration of the processing circuit:

- *The instrumentation amplifier*: INA129 (Texas Instruments), working in the differential mode with a gain of about 40.
- *The high-pass filter*: a second order Butterworth filter with cutoff frequency of 1 kHz. This value is high enough compared to the spectrum of the unwanted noises and low enough compared to the carrier frequency.
- *The analog comparator*: NE521 (On Semiconductor) with 12 ns propagation delay and maximum operating frequency of 55 MHz.
- *The ADC*: ADS8411 (Texas Instruments) with 16-bit resolution, zero latency, parallel interface, and inherent sample and hold.
- *The 32-bit microcontroller*: Tiva TM4C123GH6PM microcontroller (Texas Instruments) with integrated timers and an 80 MHz clock source.

In order to generate the modulated signal, the authors connected a sinusoidal current source to an ICG simulator (Niccomo ICG simulator, Medis), as shown in Fig. 4. The carrier frequency is set at 85 kHz to match the frequency of the reference ICG measurement device (Niccomo, Medis).

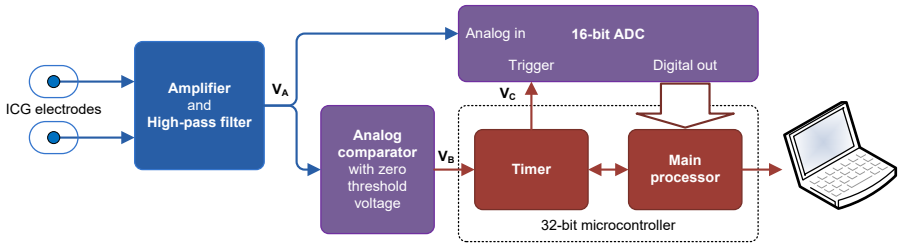

Fig. 3. System hardware with key blocks.

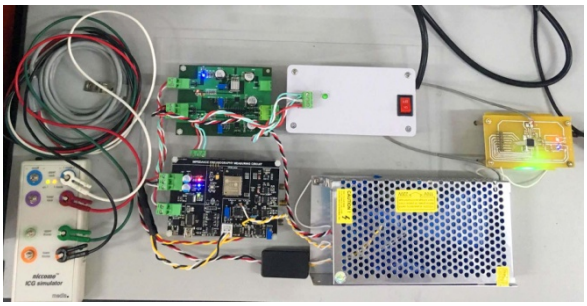

Fig. 4. Implementation of the proposed method for experiments with ICG simulator.

3.2 Experimental steps

The authors performed experiments with both the proposed system and the reference device to evaluate the application capability. Because the hemodynamic parameters are estimated from the changes in thoracic impedance, the similarity between two measured waveforms is the most important consideration.

First, with the proposed processing circuit, the current source was connected to current-electrodes of the ICG simulator. Immediately, a weak 85 kHz modulated signal was generated between the voltage-electrodes. This modulated wave represents the signal that can be measured from a healthy human body. Then, the proposed circuit was used to amplify and demodulate the signal to extract ΔZ. The impedance is the result of the division of the measured voltage by the applied current.

Second, the reference device was also used to process the simulated bioimpedance signal. The measured values were exported into Excel files. After that, all data were normalized for comparison; the results are presented in the next section.

4. Results

First, the proper operation of the designed system was confirmed by waveforms at some intermediate nodes. Figure 5 shows the strip charts on an oscilloscope of three measured signals: the 85 kHz modulated wave, V_A (see Fig. 3); the square wave at the output of the comparator, V_B; and the trigger signal that has a falling edge whenever the modulated wave reaches the top value, V_C. Here, the trigger signal is temporally enabled at all peaks for the best illustration. This experiment was performed many times and the circuit was carefully calibrated to make sure that the timing is perfectly same as the desired charts in Fig. 2.

Second, the demodulated signals at the outputs of the proposed system and the reference device were compared. Figure 6 presents the raw value of ΔZ and the filtered one after normalization. Here, the raw signal is filtered by a simple digital low-pass filter to smooth the changes in ΔZ. Details of the filter and other auxiliary blocks are not the aim of this work, therefore, they would be shown in another study. On the other hand, the output waveform of impedance measured by the reference device is reported in Fig. 7. The similarities between the strip charts in the two figures confirmed that the proposed system has a capability to measure the ICG signal for estimating the hemodynamic parameters.

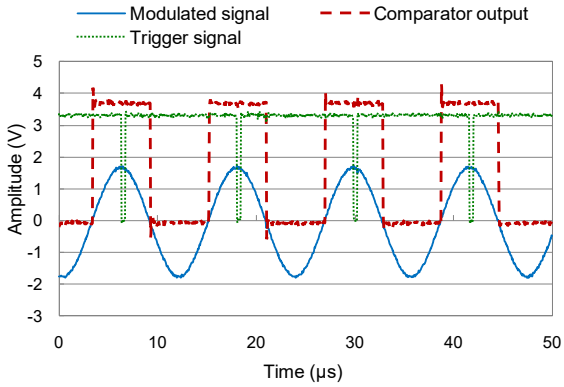

Fig. 5. Intermediate waveforms on oscilloscope.

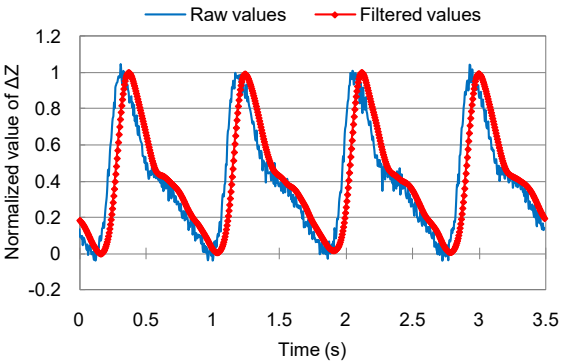

Fig. 6. Normalized values of ΔZ measured by the proposed system.

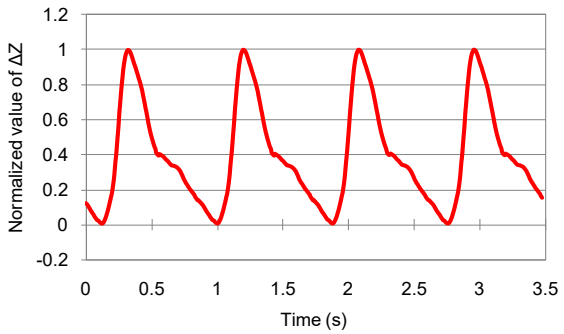

Fig. 7. Output waveform of impedance measured by the reference device.

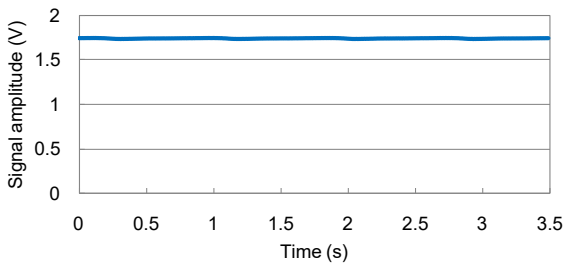

Fig. 8. Waveform of the digital signal at the output of the 16-bit ADC.

4. Discussion

The experimental results have already proven the rationality of the new idea and the application capability of the proposed system. After measuring the changes in impedance, the hemodynamic parameters could be easily calculated by the algorithms in [3, 7, 8]. In fact, there may be many ways to estimate the hemodynamic indices from ΔZ. Hence, finding the best way is still the goal of many studies. The authors may contribute to addressing this issue in their next publications.

The advantage of the proposed method is that the good results have been achieved with low-performance hardware. Before this work, the authors had faced great challenges when trying to capture the ICG signal by both analog circuits and high speed ADCs. In the first scheme, the Butterworth filters were used due to the maximal flatness of amplitude response in the pass band. However, the phase response of this filter is not linear; therefore, this can cause a significant distortion [16]. Furthermore, a nonlinear operation on any signal is inevitable with analog amplitude demodulator [17]. Hence, with many things considered, the authors concluded that the first solution could not ensure the fidelity of the acquired bioimpedance signal for CO estimation. In the second scheme, the 85 kHz carrier wave requires a sampling rate of at least two times greater. In fact, the rate should not be lower than 5 MHz to ensure the accuracy of the signal amplitude. The high speed and the high resolution make the output data from the ADC could be up to 80 Mbps. This data flow cannot be completely handled without the use of a dedicated digital signal processing circuit. Here, there is a contradiction between the high data rate and the low frequency spectrum of the ICG signal. On the contrary, the proposed system can precisely acquire the ICG signal at much lower sampling rates with relatively optimal hardware. Although the acquisition system does not require low-pass filtering, there is no aliasing introduced in the reconstructed signal. This is due to the peak sampling mechanism allows the ADC to detect and digitize the envelope of the modulated wave concurrently [18]. Regarding the ADC resolution, Fig. 8 confirmed that the use of 16-bit types is necessary. After digitization, the DC level (for Z_0 calculation) of the achieved signal is much greater than its AC component (for computing CO).

The proposed system has its requirement and limitation. First, the difference between the carrier frequency and the ICG spectrum must be very large. This is to make sure that the frequency of the modulated wave is almost constant. The stability of frequency is essential to trigger the ADC at exact time points. The next issue is that the sampling noise. Because the sample capacitor inside the ADC is directly charged by the input signal, a high speed sampling process may induce noise and affect the input signal. However, this problem could be overcome by buffering the input signal with a good voltage follower.

5. Conclusion

In this study, the authors have successfully proposed a new scheme of measuring impedance cardiography for cardiac output estimation. The proposed design allows directly digitizing the high frequency modulated signal at much lower sampling rates. The early digitization solution makes subsequent processing steps could be highly flexible and easy to be upgraded in further studies. The experimental results have already proven the rationality of the idea and confirmed the capability of the proposed system. Although more optimized configurations and better processing algorithms need to be discovered in further works. The achieved results could be a noticeable reference for future designs.

Acknowledgments

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