New Approach to Designing Reliable Circuit for Acquiring Impedance Cardiography Signal (ICG)

Nguyen Minh Duc, Nguyen Tuan Linh, Nguyen Duc Thuan

Abstract - Impedance Cardiography (ICG) is a non-invasive method which can continuously monitoring cardiac activities. By measuring the changes of thoracic impedance (ΔZ) during a cardiac cycle, many Hemodynamic parameters can be calculated such as Stroke Volume (SV), Cardiac Output (CO), Total Fluid Content (TFC) and so on, that are impossibly derived using solely ECG. However, the accuracy of this method is still a problem. One of the major causes comes from the analog system that acquires and processes the input signals, in which ΔZ is too small compared to noises. Some common drawbacks were found in some ICG measurement devices, including non-optimal demodulation method, cut-off frequency being to low, DC noises and analog differentiation. Therefore, ICG signals often lose its higher frequency components, that are maybe related to a certain cardiac symptom. On this paper, we propose a new design that can ensure acquiring completely thoracic impedance signals (including the impedance change - ΔZ and the base impedance - Z_0), that has a bandwidth of being up to 50Hz, without distorting the signal. This new design was experimented with simulated signals to evaluate the system's accuracy and efficiency. Specifically, amplitude-modulated sine, containing both AC and DC components with various amplitudes corresponding with ΔZ and Z_0, were used in the first experiment. Then, simulated Electrocardiography (ECG) and Electromyography (EMG) were used for evaluating the capability of this new system when dealing with a small physiological signal. The frequency spectrums of the input and output signals were measured consecutively 20 times and compared to each other after each measurement. The positive results, which were expressed by average value and the corresponding standard deviation (SD) of Pearson correlation and a predefined parameter called Spectrum Amplitude Error (SAD), exhibited the high accuracy and reliability of this new system.

Keyword: Impedance Cardiography (ICG), Impedances changes (ΔZ), Base impedance (Z_0), simulated signals, demodulation, Biomedical toolkit.

IV. INTRODUCTION

Impedance Cardiography is a non-invasive method used for measuring Cardiac Output, a vital Hemodynamics used for diagnosing and treating cardiac diseases, especially related to heart failure.

Basically, this method measures patients' thoracic impedance signals in a cardiac cycle, including the base impedance (Z_0) and the impedance change (ΔZ), whose changes have a relationship with the blood flow. By differentiating ΔZ, another signal is derived called Impedance Cardiography (ICG), expressing the speed of change in thoracic impedance (dZ/dt).

This is the most important signal, along with Z_0, taking part in equations of calculating many Hemodynamics. However, ΔZ has a very low amplitude (0,2Ω - 1Ω) [1], while the bandwidth of ΔZ and ICG are from 0.02Hz to 50Hz [2]. Therefore, in order to reduce noises in ΔZ signal, many designs use a low pass filter with a relative low cut-off frequency [3]–[5]. As a result, ICG signal, derived from ΔZ, lost its higher frequency components and the signal is normally attenuated or distorted in frequency spectrum and phase.

Most of previous Impedance Cardiography measurement systems [3], [4], [5], [8] the most prevalent demodulation method is the amplitude synchronous demodulation, including a multiplier and a followed low pass filter. However, this method has a problem that contributes to the reduction of accuracy. If the modulated signal has a phase shift of α radian compared to the carrier signal, the amplitude of the demodulated signal will be decreased by 1 / cos(α) times. This amount of phase lag is partially caused by the cell membrane, which is acted as a capacitor when a high frequency current flow through the body. Therefore, this phase shift will vary according to physiological characteristics of each individual patient and it cannot be predicted.

The second reason related to the accuracy's reduction is the filter type. Commonly, the Butterworth topology is often used, but it is not the best type producing linear-phase respond. Combining with a low cut-off frequency, it will distort the signal in frequency spectrum.

Besides that, in many ICG measurement systems, the base impedance (Z_0), which contains static impedance components located in the thorax, is also received with errors. This signal is often separated from the demodulated signal by a low pass filter with a cut-off frequency of below 1Hz. Because the modulation module is complicated and the DC error is inevitable, the received Z_0 component will also vary and cause error to the calculations.

Finally, the ICC signal, the most important signal used for calculating Hemodynamics such as CO or SV,

Manuscript Received on March 2015.

Nguyen Minh Duc, M.Sc. Department of Electronics technology and Biomedical Engineering, Hanoi University of Science and Technology, Hanoi, Vietnam.

Mr. Nguyen Tuan Linh, Department of Electronics technology and Biomedical Engineering, Hanoi University of Science and Technology, Hanoi, Vietnam.

Prof. Nguyen Duc Thuan, Department of Electronics technology and Biomedical Engineering, Hanoi University of Science and Technology, Hanoi, Vietnam.

is derived from an analog differential circuit in some systems. Thus, the received signal is distorted and noise interference becomes more serious.

On this paper, we propose some analog processing solutions in order to overcome those drawbacks and improve the accuracy of acquiring thoracic impedance signals. Those solutions will be adapted into a new design of an ICG measurement system. In order to evaluate the new system, various kinds of simulated signals are used for experiments. Firstly, amplitude modulated sine signals, containing both AC and DC components, are used for evaluating the system's gain respond when dealing with one-frequency component and with DC component. Then, in order to test the efficiency of the system when dealing with physiological signals having bandwidth of DC-50Hz, the simulated ECG and EMG are used. The frequency spectrums of input signals are compared to the corresponding output signals' to evaluate the error of this new system.

II. THE OBJECT OF STUDY

Our desire in the future is to analyze and extract the abnormal features of ICG curve related to symptoms or cardiac diseases. Hence, it is necessary to *receive the ICG signal with full bandwidth without distortions.* On this paper, we focus on receiving Z_0 and ΔZ signals at the best, that is the starting point of getting a full ICG sginal and calculating acurrately Hemodynamics, by proposing improved analog processing solutions.

A. The Amplitude And Frequency Of The Current Injecting Into The Thorax

In order to ensure a good SNR of the input signal, the current's frequency is still selected to be 100kHz and the amplitude is 4mA (2.8mA rms), according to our previous study [6]. This value is absolutely meet the safety standard IEC 60601-1.

B. Input Impedance Signal

According to [1], the base impedance Z_0 of Western people ranges from 20Ω to 35Ω, and the impedance change ΔZ is 0.2 - 1Ω. These components also vary denpending on a certain cardiac disease or symptom, like patients suffering from Hypertension often have lower Z_0 [7].

Up to now, there are not any studies about thoracic impedance values of the Southeast Asian, especially Vietnamese. However, based on the facts that having a smaller body, also meaning less total amount of blood, as well as a weaker heart, we suppose that Vietnamese' ΔZ would be lesser and Z_0 would be bigger than the Western people. Therefore, this system is aimed for measuring a larger range of ΔZ (0.05 - 1Ω) and Z_0 (10 - 50Ω).

III. PROPOSED PROCESSING ANALOG SOLUTIONS

In many ICG measurement systems, there are four major issues needed to be improved: *the demodulation method, filter selection, method of acquiring* Z_0 and *method of acquiring ICG*.

A. Demodulation Method

Amplitude synchronous demodulation method with a multiplier and a followed low pass filter exhibits its problem when existing phase shift between the modulated signal and the carrier. Specifically, an alternate current can be expressed as: $I.\sin(2\pi f)$, with I, the amplitude and f, the frequency (here is 100kHz). The thoracic impedance signals being modulated by the current can be represented as: $I.\sin(2\pi f + \alpha).(Z_0 - \Delta Z)$, with α is the dephasing, Z_0 and ΔZ are the two impedance components. When this modulated signal is made through a synchronous demodulation, the acquired output signals can be expressed as below:

$$U_o = I.\sin(2\pi f).I.\sin(2\pi f + \alpha).(Z_0 - \Delta Z)$$

$$\downarrow \textit{Low pass filter}$$

$$U_o = \frac{1}{2} I^2.\cos(\alpha).(Z_0 - \Delta Z)\ (\mathrm{V}) \qquad (2)$$

The output signal is attenuated by $1/\cos(\alpha)$ times and this attenuation partially depends on physiological characteristics of individual body, leading to the error of the output signal.

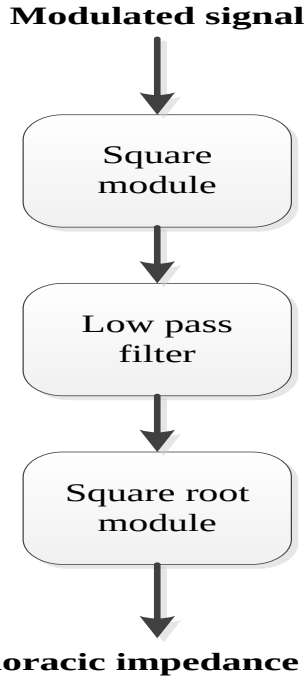

Figure 1. New demodulation method

Because the dephasing in the equation (2) cannot be eliminated by any means, we propose a new solution (Fig. 1), in which the appearance of the dephasing α does not affect the result and the ICG signal will be received completely. Specifically, the output signal of the square module will be:

$$U_{o(\mathrm{square})} = I^2.\sin(2\pi f + \alpha)^2.(Z_0 - \Delta Z)^2$$

In the next step, the signal after the low pass filter will be as presented in equation (4):

$$U_{o(LPF)} = \frac{1}{2} I^2 . (Z_0 - \Delta Z)^2 \quad (4)$$

Finally, the original signal will be restored after the square root module without any attenuations caused by the dephasing:

$$U_{o(\text{square root})} = \frac{1}{\sqrt{2}} . I . (Z_0 - \Delta Z) \quad (5)$$

Equation (5) theoretically and mathematically proves the higher efficiency of this new method when it eliminates the appearance of the dephasing α.

B. Analog Filters

In vast majority of cases, Butterworth topology is often used for analog filters when processing a physiological signal. Furthermore, the cut-off frequency for ΔZ in some ICG measurement systems are lower or close to the signal's frequencies [3], [4], [5]. Therefore, the output signal is distorted and not fully restored because the phase respond is not quite linear in the passband and the frequency spectrum is shorter than the desired bandwidth.

Thus, a proper topology for the analog filters would be Bessel, which can produce a nearly absolute linear phase respond. The passband of the analog filters would also be expanded, so that the gain respond is flat enough for the thoracic impedance signal's bandwidth (up to 50Hz).

C. Method of Acquiring Z_0

Because the base impedance (Z_0) also participates in calculating CO, the error in receiving Z_0 will lead to the error of CO parameter.

Therefore, in order to avoid inevitable and complicated DC noises caused by demodulation module, we decide to start with the modulated signal right before the modulation (figure 2). This selected signal are already free of DC components because of the high pass filter before with a cut-off frequency of above 1kHz. This signal then will be demodulated using the simplest method including diode, capacitor and resistor. Finally, a 0.2Hz low pass filter is followed to remove AC components including ΔZ and acquire Z_0.

The advantage of this method is the DC offset is predictable and highly stable, while the DC noises interfere at much lower level due to its simplicity. Therefore, it is easier to calculate the accurate output Z_0 value.

D. Method of Acquiring ICG Signal

In some designs, ΔZ signal is made through an analog differentiation to receive ICG signal. However, the analog differentiation is, in fact, a high pass filter with the ΔZ's bandwidth being within the transition between passband and stopband.

As a result, the gain of ICG signal is very small while the noises' level are significant. Therefore, we decide to implement a digital differentiation instead of the analog module.

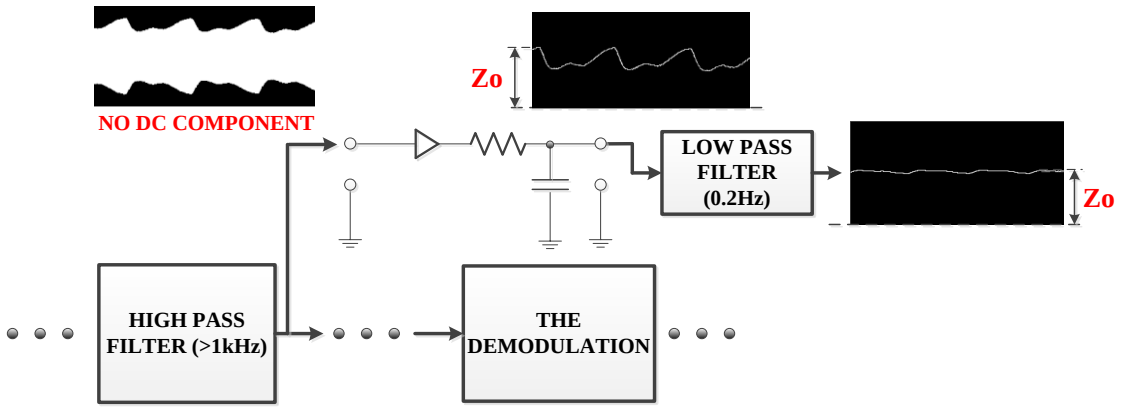

Figure 2. New method of acquiring Z_0

V. NEW APPROACH TO A SYSTEM DESIGN

This new ICG measurement system (figure 3) was and processing thoracic impedance signals. designed based on one of our previous study [5], [6]. There are two major parts: *the current source*, generating an alternate current injecting into the body, and *the analog processing*.

Current Source, *Pre-amplifier* and the *0.02Hz high pass filter* are inherited from our previous study [5], [6]. Proposed Modules For Improving The Quality Of Processing are:

10kHz – 300kHz Band-pass filter: this module contains a 10kHz high pass filter and a 300kHz low pass filter. Both of them are second order Bessel filters. The pass band is expanded to 290kHz to ensure the flatness in the gain respond.

Demodulation: contains a square and a square root using IC MPY634, a specialized IC for amplitude modulation and demodulation, and a low pass filter between those two modules, which is also a second order Bessel filter. The cut-off frequency is 350Hz for fully covering the signal's bandwidth.

350Hz low pass filter: the square root module can produce high-frequency noises with amplitudes of around 15mV, that is large enough compared to ΔZ (around 40mV). Hence, a 350Hz low pass filter is used for continuously removing noises.

Amplifier: after *the 0.02Hz high pass filter*, the received signal will be (-ΔZ). The amplifier module is designed as an inverter amplifier with a gain of 50 times,

so that ΔZ signal can be acquired with a amplitude of above 100mV before sampling.

Envelop detection: using a diode, a capacitor and a resistor for demodulating the signal. Shottky diode 1N5819 is selected due to its low forward voltage (0.34V at 0.1A), high speed (10000 V/µs) and low barrier capacitor (60pF at the frequency of above 100kHz). These characteristics are suitable for rectifying a signal with the frequency of 100kHz. Besides that, the selected discharge time of capacitor is 100µs, so that the signal's envelop can be detected smoothly enough without attenuation.

0.2Hz low pass filter: with the desire of only restoring DC component, a passive low pass filter is selected.

In addition, IC OPA211 / OPA2211 with low voltage noise (< 2 nV/$\sqrt{Hz}$), low current noise (< 3.2 pA/$\sqrt{Hz}$), wide bandwidth (45MHz at K = 1), low gain's error and wide power supply (±18V) is used for filters and amplifier in the system.

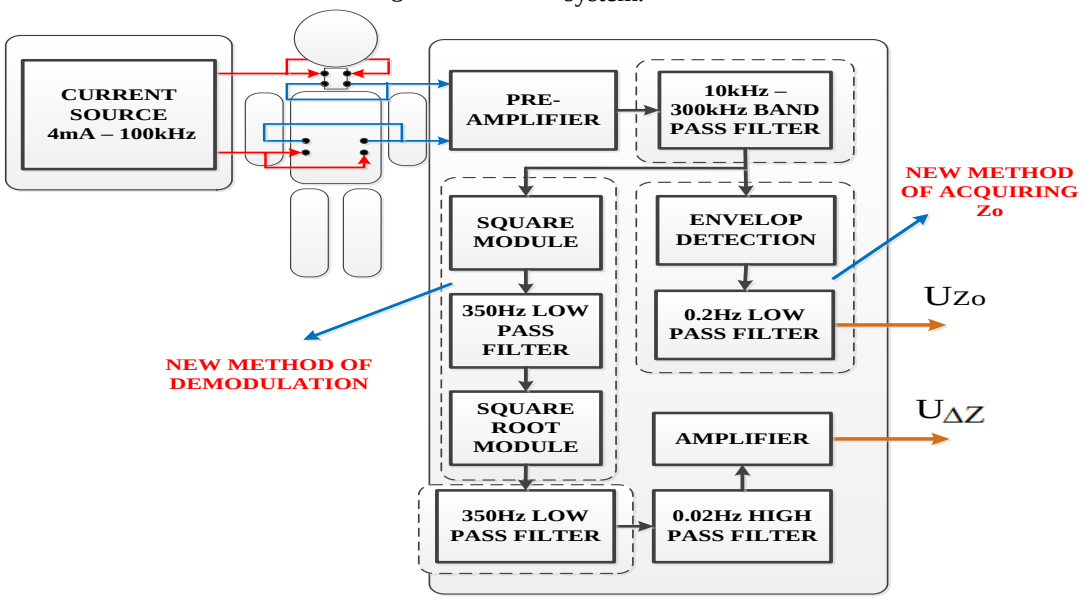

Figure 3. The analog processing design

VI. EXPERIMENTS AND RESULTS

At first, the system was experimented with sine signals to test its basic functions. Then, simulated ECG and EMG were used for evaluating the efficiency of system when dealing with physiological signals.

During the experiment, we had used some tools presented below:

- Sampling module of Board NI Elvis II with the resolution of 12 bits, input voltage noise of 0.05mV. The selected sampling frequency is 1000 samples/second.
- Labview 2012 for simulating ECG, EMG signals, receving processed signals and displaying.
- A digital oscilloscope HM1507-3 and a function generator FG-32.
- An amplitude modulation module for modulating simulated signals.

A. Experiment With Sine Signals Methodology

At the first stage, simple simulated sine signals was created, modulated and made through the new system, the output signals then were recorded at the ΔZ's and Z_0's output terminations to evaluate the system's gain respond.

Specifically, sine signals with varied frequencies and amplitudes corresponding with ΔZ's bandwidth (0.02Hz - 50Hz) and amplitude's range (impedance of 0.05Ω - 1Ω ↔ voltage of 0.2mV - 4mV with the current of 4mA) were generated. Then, these signals, containing DC components corresponding with Z_0 (10Ω - 50Ω ↔ 40mV - 200mV), were modulated with a 100kHz carrier and made through the system. The output signals then will be measured to evaluate the system's gain respond.

Result

Table I below demonstrated the output results recorded at the ΔZ termination, and table II demonstrated the results of receiving DC components by new method of acquiring Z_0 component.Those results illustrated that there were very little errors in the system's gain. For DC component, the error of merely 1% proved the high accuracy of the new analog solution for acquiring Z_0. In terms of AC component, the system also exhibited a flat gain respond within the bandwidth of 50Hz with the average error of 1%. Those positive results were very important for us to implement the second experiment.

Table I. The output results at the □Z termination

input sine (corresponding with ΔZ)	theoretical output result	output result at ΔZ termination					
		1Hz	10Hz	20Hz	30Hz	40Hz	50Hz
0.2mV	**0.2V**	~0.195V	~0.195V	~0.195V	~0.195V	~0.195V	~0.19V

0.5mV	**0.5V**	~0.49V	~0.49V	~0.49V	~0.49V	~0.49V	~0.485V
1mV	**1V**	~0.98V	~0.98V	~0.98V	~0.985V	~0.985V	~0.98V
2mV	**2V**	~1.96V	~1.96V	~1.96V	~1.95V	~1.95 V	~1.94V
4mV	**4V**	~3.93V	~3.93V	~3.92V	~3.92V	~3.91V	~3.9V

Table II. The output results at the Z_0 termination

input DC component (corresponding with Zo)	theoretical output result	output result at Zo termination
40mV	0.8V	~0.79V
100mV	2V	~1.99V
150mV	3V	~2.98V
200mV	4V	~3.98V

B. Experiment with ECG and EMG (simulated physiological signals)

Methodology

In the second experiment, simulated physiological signals, that had bandwidth corresponding with the thoracic impedance signals, were used to evaluate the system's accuracy.

Because a simulated ΔZ (0.02Hz - 50Hz) was still absent, we decided to use Biomedical toolkit of Labview 2012 to simulate ECG (bandwidth of 20Hz) and EMG (bandwidth of 50Hz) with amplitudes ranging from 0.2mV - 4mV. These simulated signals were then modulated with a 100kHz carrier and made through the system. Thereafter, the input and output signals were acquired simultaneously and their frequency spectrums were compared to each other. In order to evaluate effectively, the comparisons were expressed by *four values* of *two parameters*:

Pearson correlation:

- avr_r_{sbw}: the average of all Pearson correlations in a sub-bandwidth, which is explained below.
- SD_r_{sbw}: the corresponding standard deviation of avr_r_{sbw}.

Spectrum Amplitude Deviation (SAD):

- avr_SAD_{sbw}: the average of all SAD values, which are also explained later in this section, in a sub-bandwidth.
- SD_SAD_{sbw}: the corresponding standard deviation of avr_SAD_{sbw}.

Specifically, after each interval of 20ms, the frequency spectrums of unprocessed and processed signals were calculated in order to achieve a good frequency resolution (0.05Hz) in spectrum. We divided the bandwidth of 50Hz into five sub-bandwidths of 10Hz. In each sub-bandwidth, we chose the same set of consecutive frequency components, whose distances were 0.02Hz, e.g. 0.02Hz, 0.04Hz, etc., from both signals to compare and convert to Pearson correlation. After calculations, each sub-bandwidth would have a set of Pearson correlations: $r = \{r_1, r_2, \dots, r_n\}$ with n is the number of chosen frequency components in that sub-bandwidth (n = 50). However, in order to present an overall characteristic of a sub-bandwidth, we selected the average value of that set to represent its sub-bandwidth:

$$r_{sbw} = \frac{\sum_1^n r_k}{n} \quad (6)$$

r_{sbw} is the representative Pearson correlation of a sub-bandwidth, and r_k is the Pearson correlation of the kth frequency component.

We repeated that process 20 times for more detailed evaluation. After 20 measurements, each sub-bandwidth continued to possess a set of 20 representative Pearson correlations: $r_{sbw\_measured} = \{r_{sbw_{1th}}, r_{sbw_{2th}}, \dots, r_{sbw_{20th}}\}$. In order to get a general view of the outcomes, the average value of that set and the corresponding SD were calculated. That are avr_r_{sbw} (equation (7)) and SD_r_{sbw} (equation (8)) mentioned above.

$$avr\_r_{sbw} = \frac{\sum_1^{20} r_{sbw_{kth}}}{20} \quad (7)$$

$$SD\_r_{sbw} = \sqrt{\frac{\sum_1^{20}\left(r_{sbw_{kth}} - avr\_r_{sbw}\right)^2}{20}} \quad (8)$$

avr_r_{sbw} is the average of a sub-bandwidth's Pearson correlation after 20 measurements, SD_r_{sbw} is the corresponding standard deviations, and $r_{sbw_{kth}}$ is the Pearson correlation recorded in the mth measurement.

Beside assessing the system using Pearson correlation, in order to evaluate more precisely and deeply, we proposed another parameter called *Spectrum Amplitude Deviation (SAD)*, which demonstrates the error in amplitude of output signal's spectrum compared to input signal's:

$$SAD_k = \frac{A_{ik} - A_{ok}}{A_{ik}}.100\% \quad (9)$$

Err_k is Error value of the kth frequency component, A_{ik} and A_{ok} are the amplitudes of the kth frequency component of the input and output signal.

The process of calculating SAD values was the same as calculating Pearson correlation. After an interval of 20ms, the frequency spectrums of the input and output signals were compared and SAD value of each frequency component (0.02Hz, 0.04Hz, etc.)

was calculated. Therefore, each sub-bandwidth would have a set of SAD values: $SAD = \{SAD_1, SAD_2, \dots, SAD_n\}$. We continued to select the average value of that set to represent its sub-bandwidth:

$$SAD_{sbw} = \frac{\sum_1^n SAD_k}{n} \quad (10)$$

SAD_{sbw} is the representative SAD value of a sub-bandwidth.

After 20 consecutive measurements, there was a set of 20 representative SAD values corresponding to each sub-bandwidth:$SAD_{sbw\_measured} = \{SAD_{sbw_{1th}}, SAD_{sbw_{2th}}, \dots, SAD_{sbw_{20th}}\}$. From these sets, avr_SAD_{sbw} and SD_SAD_{sbw} were calculated as below:

$$avr\_SAD_{sbw} = \frac{\sum_1^{20} SAD_{sbw_{kth}}}{20} \quad (11)$$

$$SD\_SAD_{sbw} = \sqrt{\frac{\sum_1^{20}\left(SAD_{sbw_{kth}} - avr\_SAD_{sbw}\right)^2}{20}} \quad (12)$$

avr_SAD_{sbw} is the average of a sub-bandwidth's SAD values after 20 measurements, SD_SAD_{sbw} is the corresponding standard deviations, and $SAD_{sbw_{kth}}$ is the SAD value recorded in the m[th] measurement.

Those four values including avr_r_{sbw}, SD_r_{sbw}, avr_SAD_{sbw}, SD_SAD_{sbw} were used in this experiment to evaluate how distorted the received output signals were when the system worked with small physiological signals.

Result

Table III and IV are the results of comparisons (through Pearson correlations) when processing ECG (2 sub-bandwidths) and EMG (5 sub-bandwidths), while Table V (ECG) and VI (EMG) express the SAD values.

In terms of simulated ECG, the outcomes were very positive, while the average Pearson correlations were always above 0.9 with small signal and even reached 0.98 with higher input amplitude. The SD values were also very low, only 0.015 in average. Referring to table V, the errors in amplitude between the input and output signals' spectrum were insignificant, accounting for only 1.5% in average with SD of around merely 0.5%. It proved that, with bandwidth of below 20Hz, the system worked very accurately.

Table III. The correlation between two frequency spectrums of input and output signals (ECG)

	0.2Hz – 10Hz (sub-bandwidth 1)	10Hz – 20Hz (sub-bandwidth 2)
4mV	$avr\_r_1 = 0.98$ $SD\_r_1 = 0.005$	$avr\_r_2 = 0.98$ $SD\_r_2 = 0.005$
1mV	$avr\_r_1 = 0.97$ $SD\_r_1 = 0.005$	$avr\_r_2 = 0.96$ $SD\_r_2 = 0.005$
0.2mV	$avr\_r_1 = 0.96$ $SD\_r_1 = 0.01$	$avr\_r_2 = 0.96$ $SD\_r_2 = 0.01$

Table IV. The correlation between two frequency spectrums of input and output signals (EMG)

	0.2Hz – 10Hz (sub-bandwidth 1)	10Hz – 20Hz (sub-bandwidth 2)	20Hz – 30Hz (sub-bandwidth 3)	30Hz – 40Hz (sub-bandwidth 4)	40Hz – 50Hz (sub-bandwidth 5)
4mV	$avr\_r_1 = 0.99$ $SD\_r_1 = 0.005$	$avr\_r_2 = 0.98$ $SD\_r_2 = 0.005$	$avr\_r_3 = 0.97$ $SD\_r_3 = 0.005$	$avr\_r_4 = 0.97$ $SD\_r_4 = 0.01$	$avr\_r_5 = 0.96$ $SD\_r_5 = 0.01$
1mV	$avr\_r_1 = 0.97$ $SD\_r_1 = 0.005$	$avr\_r_2 = 0.96$ $SD\_r_2 = 0.005$	$avr\_r_3 = 0.95$ $SD\_r_3 = 0.01$	$avr\_r_4 = 0.95$ $SD\_r_4 = 0.02$	$avr\_r_5 = 0.95$ $SD\_r_5 = 0.02$
0.2mV	$avr\_r_1 = 0.95$ $SD\_r_1 = 0.005$	$avr\_r_2 = 0.95$ $SD\_r_2 = 0.01$	$avr\_r_3 = 0.93$ $SD\_r_3 = 0.03$	$avr\_r_4 = 0.9$ $SD\_r_4 = 0.03$	$avr\_r_5 = 0.9$ $SD\_r_5 = 0.03$

Table V. Calculated SAD values of ECG

	0.2Hz – 10Hz (sub-bandwidth 1)	10Hz – 20Hz (sub-bandwidth 2)
4mV	$avr\_SAD_1 = 1.1\%$ $SD\_SAD_1 = 0.34\%$	$avr\_SAD_2 = 1.32\%$ $SD\_SAD_2 = 0.39\%$
1mV	$avr\_SAD_1 = 1.17\%$ $SD\_SAD_1 = 0.35\%$	$avr\_SAD_2 = 1.38\%$ $SD\_SAD_2 = 0.42\%$
0.2mV	$avr\_SAD_1 = 2.4\%$ $SD\_SAD_1 = 0.51\%$	$avr\_SAD_2 = 2.81\%$ $SD\_SAD_2 = 0.57\%$

Regarding EMG with larger bandwidth, this new system still showed its good quality of processing, while the avr_r values were always around 0.95 with SD of 0.02 in average. Even in the sub-bandwidth of 40 - 50Hz where the power-line noise (50Hz) and low frequency noises could be significant, SD of the Pearson correlation are only 0.02 to 0.03. About the error in amplitudes, due to the complexity of EMG, the differences between input and output signals were bigger compared to ECG, accounting for 1% - 2% for the frequencies of below 30Hz and 2% - 5% for the frequencies of above 30Hz to 50Hz (with EMG's amplitude of 1mV). However, small SD of only 1% in maximum showed a very stable gain respond in the bandwidth of 50Hz. These results were still promising for further developments.

Table VI. Calculated SAD values of EMG

	0.2Hz – 10Hz (sub-bandwidth 1)	10Hz – 20Hz (sub-bandwidth 2)	20Hz – 30Hz (sub-bandwidth 3)	30Hz – 40Hz (sub-bandwidth 4)	40Hz – 50Hz (sub-bandwidth 5)
4mV	$avr\_SAD_1 = 1.2\%$ $SD\_SAD_1 = 0.29\%$	$avr\_SAD_2 = 1.31\%$ $SD\_SAD_2 = 0.37\%$	$avr\_SAD_3 = 1.8\%$ $SD\_SAD_3 = 0.39\%$	$avr\_SAD_4 = 2.36\%$ $SD\_SAD_4 = 0.59\%$	$avr\_SAD_5 = 3.52\%$ $SD\_SAD_5 = 0.61\%$
1mV	$avr\_SAD_1 = 1.4\%$ $SD\_SAD_1 = 0.33\%$	$avr\_SAD_2 = 1.67\%$ $SD\_SAD_2 = 0.4\%$	$avr\_SAD_3 = 1.97\%$ $SD\_SAD_3 = 0.58\%$	$avr\_SAD_4 = 2.76\%$ $SD\_SAD_4 = 0.65\%$	$avr\_SAD_4 = 3.97\%$ $SD\_SAD_4 = 1.04\%$
0.2mV	$avr\_SAD_1 = 2.5\%$ $SD\_SAD_1 = 0.39\%$	$avr\_SAD_2 = 2.82\%$ $SD\_SAD_2 = 0.45\%$	$avr\_SAD_3 = 4.29\%$ $SD\_SAD_3 = 0.58\%$	$avr\_SAD_4 = 6.3\%$ $SD\_SAD_4 = 1.09\%$	$avr\_SAD_4 = 7.94\%$ $SD\_SAD_4 = 1.28\%$

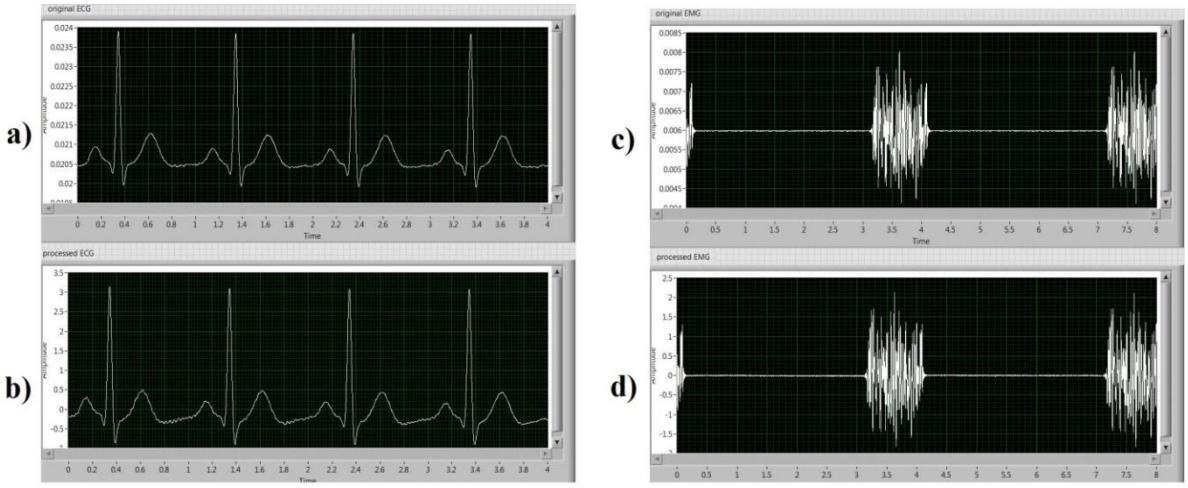

Figure 4. Original simulated ECG and EMG: (a) and (c), and ECG and EMG after being processed: (b) and (d)

Figure 6 displays the images of unprocessed and processed ECG and EMG at the amplitude of 1mV. About the signals' curve, the new system had been successful in preserving the shape and keeping the gain stably.

In general, the above results had proved the high accuracy and stable of the system when working with a physiological signal, especially a very small signal. It is very necessary in case of measuring patients suffering from heart failure (ΔZ is very low) in the future.

VII. DISCUSSION AND CONCLUSION

After understanding and analyzing some common drawbacks appearing in ICG measurement systems, we had proposed new analog processing solutions for improving the capability of acquiring and processing two thoracic impedance signals: ΔZ and Z_0. The two experiments indicated that the new system had achieved a high accuracy in processing, with the error of below 1% regarding DC component and around 3% in average regarding AC component with the frequency of below 50Hz. Therefore, the new system had showed its high potential when working with thoracic impedance signals with very low amplitude and bandwidth expanded to 50Hz. These results are important and promising for further desire of analyzing ICG signals related to symptoms or cardiac diseases, and calculating Hemodynamics.

In the next study, we desire to use this system to measure thoracic impedance signals of both healthy volunteers and voluntary patients in order to gather data, investigate and evaluate them for supporting doctors and medical services.

REFERENCE

1. M.C.Ipate, A.A.Dobre, A.M.Morega (2012), "The stroke volume and the cardiac output by the impedance cardiography", *U.P.C. Sci. Bull.*, Series C, Vol.74, Iss.3, pp. 193-201.
2. Barry E. Hurwitz, Liang-Yu Shyu, Chih-Cheng Lu, Sridhar P. Reddy, Neil Schneiderman and Joachim H. Nagel (1993), "Signal fidelity requirements for deriving impedance cardiographic measures of cardiac function over a broad heart rate range", *Biological Psychology*, 36, pp. 3-21.
3. Gerard Cybulski (2011), Ambulatory Impedance Cardiography – The systems and their applications, Springer.
4. Liang-Yu Shyu, Chia-Yin Chiang, Chun-Peng Liu, Wei-Chih Hu (2000), " Portable Impedance Cardiography System for Real-Time Noninvasive Cardiac Output Measurement", *Journal of Medical and Biological Engineering*, Vol. 20, Iss.4, pp. 193-202.
5. Nguyen Minh Duc, Nguyen Thai Ha, Vu Duy Hai (2013); "Design of the circuit for acquiring and processing Impedance Cardiography signals", *Journal of Science and Technology for technical universities* (ISSN 0868 – 3980), No.97, pp. 57-62.
6. Nguyen Minh Duc, Nguyen Thai Ha (2013); "Design of the continuous Cardiac Output measurement system using Impedance Cardiography and building the alternative current source applying for patient's thorax", *Journal of Science and Technology for technical universities* (ISSN 0868 – 3980), No.96, pp. 33-39.
7. David Prutchi, Michael Norris 2005, Design and development of medical electronic instrumentation, Wiley – Interscience.
8. Webster J.G (2010), Medical Instrumentation: Application and design: Measurement of flow and Volume of blood, NewYork J.Wiley Publisher.

AUTHORS PROFILE

M.Sc. Nguyen Minh Duc received a Master degree of Electronics Engineering at Hanoi University of Science and Technology in 2014. He is currently a lecturer at Department of Electronics technology and Biomedical Electronics Engineering, Hanoi University of Science and Technology (HUST), Vietnam. His current research activities are in Biomedical Instrumentation, Medical Information System, Biomedical Signal Processing, eHealth.

Mr. Nguyen Tuan Linh. He is currently final-year undergraduate student of Talent Engineer Program at School of Electronics and Telecommunications, Hanoi University of Science and Technology (HUST), Vietnam. His current research activities are in Biomedical Instrumentation, Telecommunications, Biomedical Signal Processing.

Prof. Nguyen Duc Thuan received a Doctorate of Electronics technology at Saint-Petersburg University, Russia in 1986. Currently, he is working at Department of Electronics technology and Biomedical Electronics Engineering, Hanoi University of Science and Technology (HUST), Vietnam. Now he is a full Professor and the Chair of Biomedical Engineering Research Program at HUST. His current research activities are in Biomedical Equipment Design, Medical Information System, Biomedical Signal Processing.