Power Supply and Signal Filtering for 24-Hour ECG

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***Abstract*—The goal of this research is to present signals filtering for the high resolution Holter monitoring system. The first part of this paper concentrates on working of human heart and description of architecture of biopotentials recorder. This part also includes proper amplification and filtration of interferences of bioelectric signals. The second part of this research presents practical realization of Holter monitoring system. Special attention is put on proper selection of components of the recorder and joining them into optimal structure.**

***Index Terms*—Holter monitoring system, electrocardiograph (ECG), baseline wandering, muscle noise, power line interference**

I. Introduction

NOWADAYS, heart diseases are to be considered as the symbol of progress of our civilization. More and more young people suffer from diseases resulting from incorrect heart action as well as the whole blood circulation system. Life of these people depends on fast and correct diagnosis.

Device presented in this article is designed for proper recording and archiving of biopotentials. Occurrence of these signals is indirect result of heart action and can be used for making a correct diagnosis. Simultaneously this examination is one of the most non-invasive types of diagnostic investigation.

A Holter monitoring system, also called an ambulatory electrocardiogram, is a small portable device that is used to continuously record heart's rhythm during daily activities (usually for 24 hours). The monitor records heart's rhythm through electrodes that are placed on the chest. Electrodes are small adhesive patches attached by wires to a monitor. This test helps show how the heart responds to normal and abnormal (possible during daily activities) condition that may occur during usual day. This experiment also helps observe how people react to certain medications.

II. Principle of Operation and Electrical Activity of Heart

Heart's electrical system (also called cardiac conduction system) controls all the events that occur when heart pumps blood. ECG (electrocardiogram) is a graphical representation (waveform) of the heart's electrical activity (Figure 1).

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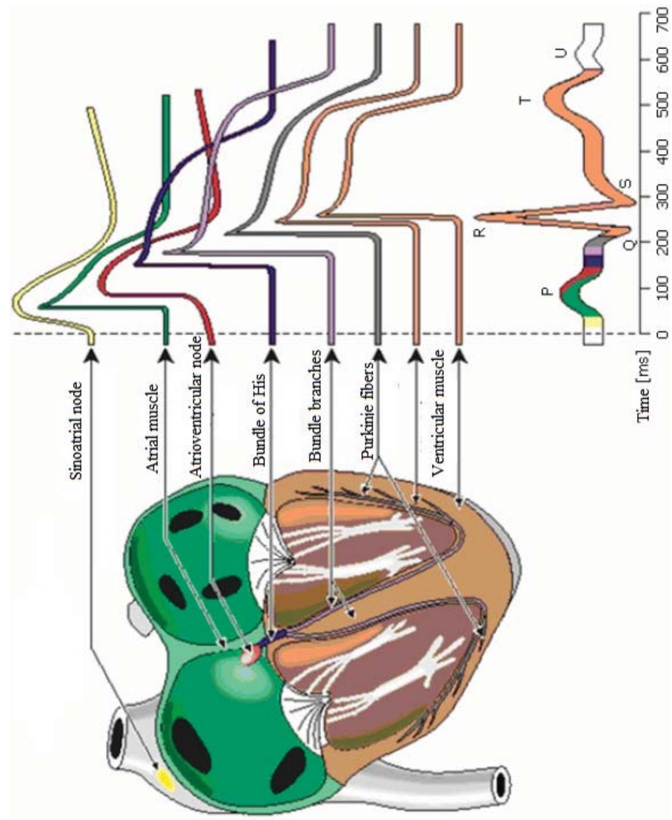

Fig. 1. Longitudinal section of a heart with a shape of heart stimulation.

This system is compound of three main parts:

- the sinoatrial (SA) node, located in the right atrium of heart - SA node depolarizes and the resulting impulse spreads across the atrial myocardium and through the internodal fibers to the AV node. The atrial myocardium contracts in response (a physical event). The fastest autorhythmic area is called the sinoatrial node and it acts as the heart's "pacemaker". This electrical activity can be recorded from the surface of the body as a "P wave" on the patient's ECG (electrocardiogram);
- the atrioventricular (AV) node, located on the interatrial septum close to the tricuspid valve (just above the ventricles) - AV node lifts the impulse and transfers it to the AV Bundle (Bundle of His). Here, the electrical impulse is held up for a brief period. This delay allows the right and left atrium to continue emptying its blood contents into the two

ventricles. The delay is recorded as a „PR interval." The AV node thus acts as a „relay station" delaying stimulation of the ventricles. long enough to allow the two atria to finish emptying. This produces a 0.1 second delay in the cardiac cycle. It takes approximately 0.03 sec from SA node depolarization to the impulse reaching the AV node, and 0.13 seconds for the impulse to get through the AV node and reach the Bundle of His;

- the His-Purkinje system, located along the walls of heart's ventricles. The impulse travels from the AV node through the bundle branches and through the Purkinje fibers to the ventricular myocardium, causing ventricular depolarization and ventricular contraction (a physical event). The electrically stimulated ventricles contract and blood is pumped into the pulmonary artery and aorta. This electrical activity is recorded from the surface of the body as a „QRS complex". Ventricles then recover from this electrical stimulation and generate an „ST segment" and T wave on the ECG.

A heartbeat is a complex series of events that take place in the heart. The heartbeat is a single cycle in which heart's chambers relax and contract to pump blood. This cycle includes the opening and closing of the inlet and outlet valves of the right and left ventricles of the heart.

Each heartbeat has two basic parts:

- diastole - during diastole, the atria and ventricles of the heart relax and begin to fill with blood;
- systole - at the end of diastole, heart's atria contract (atrial systole) and pump blood into the ventricles. The atria then begin to relax. Heart's ventricles then contract (ventricular systole), pumping blood out of the heart.

III. Requirements for an Electrocardiograph

The front end of an ECG must be able to deal with extremely weak signals ranging from 0.5 mV to 5.0 mV, combined with a dc component of up to ±300 mV—resulting from the electrode-skin contact—plus a common-mode component of up to 1.5 V, resulting from the potential between the electrodes and ground. The useful bandwidth of an ECG signal, depending on the application, can range from 0.5 Hz to 50 Hz—for a monitoring application in intensive care units—up to 1 kHz for late-potential measurements (pacemaker detection). A standard clinical ECG application has a bandwidth of 0.05 Hz up to 100 Hz.

ECG signals may be corrupted by various kinds of noise. The main sources of noise are:

- power-line interference: 50–60 Hz pickup and harmonics from the power mains;
- electrode contact noise: variable contact between the electrode and the skin, causing baseline drift;
- motion artifacts: shifts in the baseline caused by changes in the electrode-skin impedance;
- muscle contraction: electromyogram-type signals (EMG) are generated and mixed with the ECG signals;
- respiration, causing drift in the baseline;
- electromagnetic interference from other electronic devices, with the electrode wires serving as antennas, and
- noise coupled from other electronic devices, usually at high frequencies.

For meaningful and accurate detection, steps have to be taken to filter out or discard all these noise sources. Typical ECG Signal Chain Figure 2 shows a block diagram of a standard single-channel electrocardiograph. In that chain it is apparent that all filtering is done in the analog domain, while the microprocessor, microcontroller, or DSP is used principally for communication and other downstream purposes. Thus the powerful computational properties of the digital core are not readily available to deal with the signal in its essentially raw state. In addition, sophisticated analog filters can be costly to the overall design due to their inflexibility—and the space, cost, and power they require.

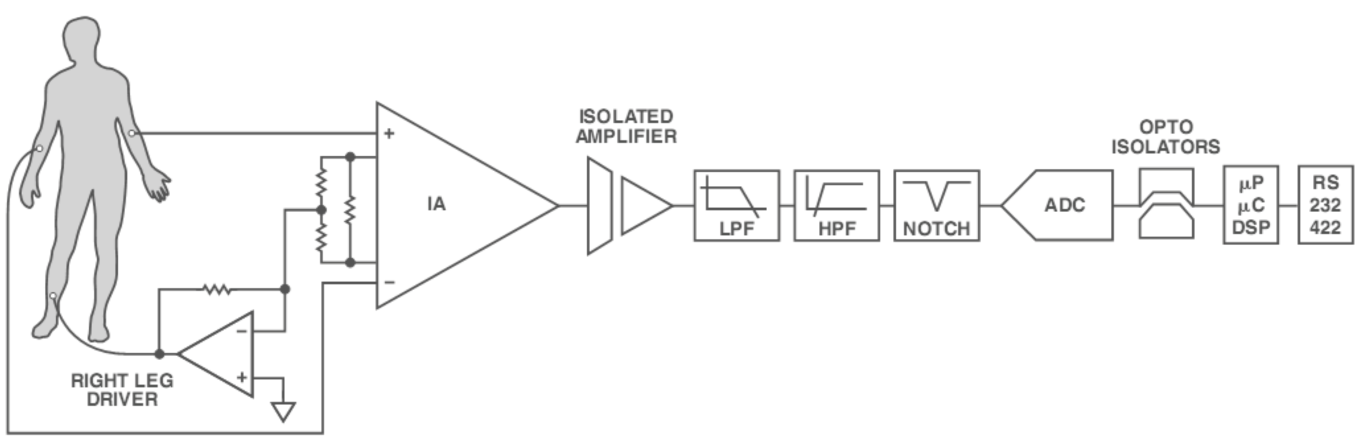

Fig. 2. Typical single-channel electrocardiograph.

IV. Power Supply System

Holter monitoring system has to be battery powered, and its time of operation should exceed 24 hours. Author decided to use 4 high capacity Ni-MH (nickel–metal hydride) rechargeable AA cells with a capacity of 2700 mAh.

The series circuit of batteries would result in increase of single cell voltage from 1.2 V to 4.8 V, enough to power an analog circuit. However, capacity of such a connection would remain at the same level of 2700 mAh, which at the assumed maximum current consumption of 150 mA, is not enough. Moreover, a failure of one of the cells would cause a drastic drop in both voltage and capacity of the entire system.

Parallel battery wiring would quadruple power supply system capacity, without changing the value of generated voltage. Due to the possibility of spontaneous discharge of unevenly loaded cells and the low availability of integrated converters that transform the voltage from 1.2 V to 5 V this approach was also abandoned.

Author decided to combine the two approaches outlined above by wiring in parallel two pairs of serially connected cells. The voltage and capacity obtained in this way are twice the values for a single battery.

In order to obtain the highest voltage present in the system - the analog circuit supply voltage, which is 5 V, step-up dc-dc converter was used. This type of converters has a very high efficiency and thus very low power consumption. It stabilizes the output voltage for a wide range of input voltages.

Step-up converter is a class of switching-mode power supply (SMPS). One of its disadvantages is ripple voltage, which could affect the measurement results. It was necessary to connect an external choke and capacitor. The choke might emit strong electromagnetic interference. Therefore, the power supply system was placed on a separate board containing a digital circuit. The digital section of the system is less sensitive to interference in comparison with the analog circuit.

The used converter comes with a build-in damping switch to minimize "ringing" at LX input. The damping switch allows to quickly dissipate energy accumulated in the choke, minimizing ringing (Figures 3, 4).

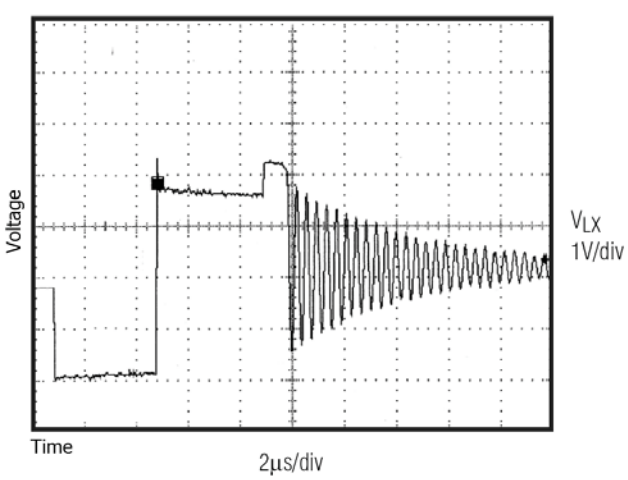

Fig. 3. LX input voltage – ringing without damping switch.

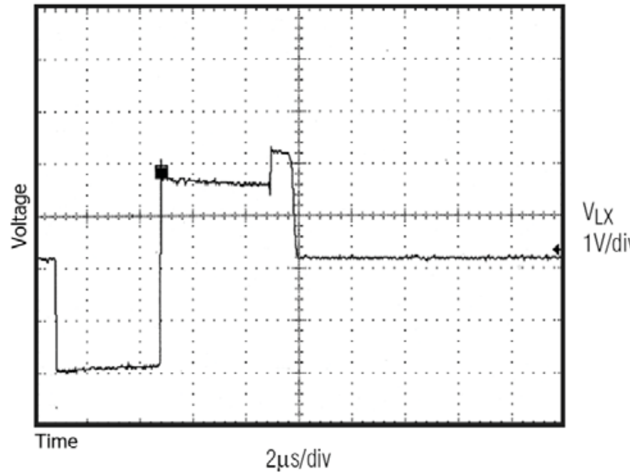

Fig. 4. LX input voltage – damping switch used, no ringing.

According to the initial expectations, the output voltage is around 5 V (Figure 5). This voltage is not constant, but features a ripple frequency of 40 Hz and amplitude of 20 mV.

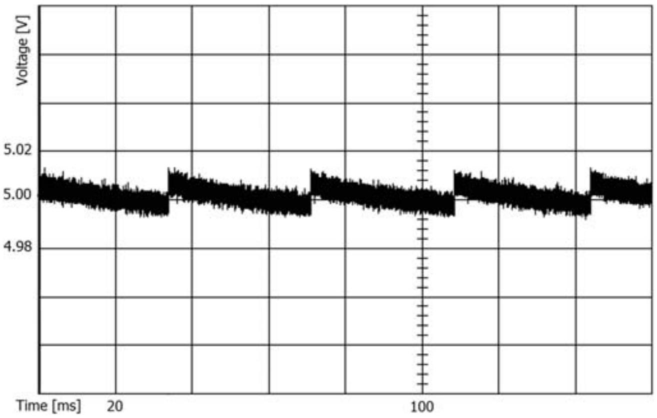

Fig. 5. Output voltage of the step-up converter.

The amplitude of ripple is only 0.4% of the determined value, although, it should be remembered that signals in the analog circuit also have small amplitude and are particularly sensitive to interference. Therefore, all the active components, which are supplied with 5 V, are characterized by a large Power Supply Rejection Ratio (PSRR). The average value of PSRR at the frequency of 40 Hz is about 110 dB for the pre amp system and almost 130 dB for op amps systems.

Additionally all the active elements of the analog circuit received a pair of capacitors that filter power supply voltage changes.

A voltage of 3.3 V, obtained from a low-dropout (LDO) voltage regulator, was used for powering the digital part of the recorder. The input voltage of the regulator corresponds to the output voltage of the converter system. This eliminates the issue of changes in the output due to a change in the input of the voltage regulator. The applied element is characterized by extremely low supply current and a high output current, which meets the requirements, as digital part of the project consumes significantly more power than the analog circuit.

Voltage reference in this project is a value of 2.5 V. It is provided by an integrated circuit featuring ultralow noise and high accuracy. The peak value of noise level is only 1.2 microvolts and the output source current is 10 mA.

V. Filters

A. The Anti-Aliasing Filter

In order to avoid the undesirable effect of aliasing, an anti-aliasing filter is employed before the signal is sampled. It is necessary to know the bandwidth of the signal of interest in order to perform this task. The use of an anti-aliasing filter is of paramount importance when sampling any signal. The effects of aliasing cannot be undone, nor can their presence always be detected. In any A/D acquisition system, the cutoff frequency of the anti-aliasing filter must always be less than one half the sampling frequency. This guarantees that the no aliasing will occur.

B. Filter Types

Some features of well-established and commonly-used filter types will be presented. Each type has specific parameters which can be optimized to approach ideal filter features at the expense of some other characteristics. Unfortunately, there is no readily available filter which can approach all the attributes of the ideal filter.

1) The Butterworth Filter

This filter is best used for its maximally flat response in the transmission passband, minimizing passband ripple. Specifying the maximum overshoot in the passband allows the determination of the minimum necessary order to achieve the desired response. This filter is best suited for applications requiring preservation of amplitude linearity in the passband region. This feature makes the Butterworth filter an ideal candidate for conditioning the ECG signal. The corner frequency, fc, of this filter is defined as the 3-dB point, as described in the previous sections. Note that the phase response of this filter is not particularly linear. This filter is completely specified by the maximum passband gain, the cutoff frequency and the filter order.

2) The Chebyshev Filter

Similar to the Butterworth filter, this filter can achieve steep rolloffs with high order designs. The Chebyshev outperforms the Butterworth's attenuation in the transition band for the same order design. This advantage, however, comes at the expense of noticeable ripple in the passband regions. The total number of maxima and minima in the passband region is equal to the filter order. Unlike the Butterworth filter, the cutoff frequency for this filter is not specified at the 3-dB point, but rather at the frequency where the specified maximum passband ripple is exceeded. Like the Butterworth filter, this filter is completely specified by the maximum passband gain, the cutoff frequency and the filter order.

3) The Thompson or Bessel Filter

The magnitude response of the Bessel filter is monotonic and smooth – there are no ripples in the transmission band or the stop band. The rolloff of this filter, however, is much less steep than the filters presented above. The main advantage of this filter is its exceptional phase linearity. This preservation of phase also minimizes "ringing" caused by sharp inputs (known as step or impulse responses) which are commonly found in the other filters.

C. Analog vs. Digital Filters

The previous section described the behavior and the specification of several classic filter types. These filters (along with many others) can be implemented in either the analog signal domain (where the signals are continuously varying voltages) or in the digital domain (where the analog signals have been sampled and are represented by an array of numbers).

Author decided to use analog filters because of some of their features. Analog filters are usually implemented with electronic circuits, making use of three fundamental components: resistors, capacitors and inductors. By arranging these components in a variety of configurations, it is possible to customize filter performance to very specific needs. In addition, operational amplifiers are commonly used to increase the performance of these filters. It is important to note that these filters are commonly used in "signal conditioning stages" before any digitization takes place. Signal conditioning generally refers to the modification of a signal for the purpose of facilitating its interaction with other components, circuits or systems. This may involve the removal of unwanted noise or the reduction of bandwidth to simplify further signal analysis or processing.

The most notable application of this kind is low-pass filtering for anti-aliasing purposes, as described above. It is critical for the anti-aliasing filter to be applied on the analog signal before any digitization occurs since the effects of incorrectly-sampled data cannot be undone.

The performance of analog filters is directly related to the quality of the components used and the circuit design. Things such as component tolerances, power consumption, design techniques and often the physical size of components – all play important roles in establishing the practical limits of analog filters.

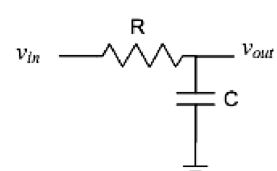

Fig. 6. Single-pole low pass filter.

Single-pole low pass filter is the simplest analog filter possible with one resistor and one capacitor. Rolloffs of this basic design is 20 dB/decade, with a corner frequency given as 1/RC (Figure 6).

Double-pole low pass filter design essentially cascades two single pole filters, facilitated by the use of an op amp. The rolloff of this filter is 40 dB/decade. The correct combination of R and C values can tailor this generic filter to have different responses, such as the Butterworth and Chebyshev filters previously described.

Higher order filters are obtained by cascading more RC single pole filters (Figure 7).

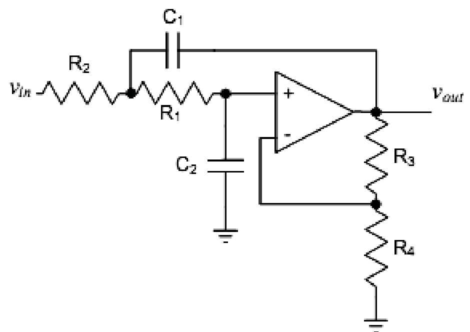

Fig. 7. Double-pole low pass filter.

VI. Considerations for Analog-to-Digital Converters

The digitization process of an analog signal is performed by a device known as an Analog-to-Digital Converter (ADC). These devices are a common component of modern electronic products, and their use is highly varied and widespread; it is important that each application is assessed by considering the advantages and limitations of the specified ADC.

When choosing an appropriate ADC for digitizing ECG signals, it is important to consider three interacting factors:

- the gain of the system,
- the input noise of the system,
- the maximum voltage output of the system.

It is important to understand how these three factors interrelate in order to determine the necessary specifications of the analog-to-digital conversion system.

A. System Gain

Knowledge of the overall system amplification is necessary for relating the output signals to the true input signals detected. For example, if the overall system gain is 1000 V/V, the occurrence of a 1 V spike at the output corresponds to a 1 mV (1/1000 V) disturbance at the input. A similar 1 V output recorded with a gain of 10 000 would be caused by a 100 µV (1/10 000 V) input signal.

When the amplitude of output signals are divided by the overall system gain to obtain the input amplitude, it's said that they are referred to input, abbreviated as "r.t.i.". This is a useful procedure for modeling and comparing signal and noise characteristics, regardless of a signal's origin.

B. System Noise

Noise can be described as any portion or aspect of the output signal which is undesirable and may possibly mask the true signal of interest. Noise can have many sources and interpretations; it can come from external radiated sources (such as "line interference"), it can be caused by electrical disturbances intrinsic to the recording environment (such as motion or stimulus artifact), and it can be caused by the nature of the recording devices themselves.

C. Signal Range

The signal range of a system is defined as the maximum voltage output the device is capable of sustaining. For example, if the voltage output of the designed system is specified to be within the range of ±5 V, it means that even in the most adverse circumstances, when the amplifiers are saturated or in the presence of excessively large noise artifacts, the system output will never exceed ±5 V.

VII. Realization of Analog Signal Filtering in Holter Monitoring System.

To lower the cost of the designed Holter monitoring system authors decided to use standard passive electrodes and dedicated wires.

There are three common noise sources which interfere with ECG signal:

- baseline wander,
- power line interference,
- muscle noise.

Baseline wander, or extragenoeous low-frequency high-bandwidth components, can be caused by perspiration (effects electrode impedance), respiration, body movements.

ECG signal recordings are often contaminated by residual power-line interference. Traditional analogue and digital filters are known to suppress ECG components near to the power-line frequency.

Muscle noise is a big problem according to recording ECG because even the slightest movement will overwhelm the waveform. The main purpose of this test is to record how a heart works under various actual conditions over an extended period. Patients being monitored do their daily activities normally so Holter monitoring system should be resistant to shocks and muscles interference.

The correctly designed filtering system should have following features:

- Elimination of frequencies associated with electromagnetic interference (radio frequency). Capacitors C1x, C2x, C3x with resistance play a part responsible for elimination of radio frequency – low-pass filter with cutoff frequency 1500 Hz.
- Removing frequencies connected to slowly fluctuating electrode voltage – electric circuit composed of resistor R5x and capacitor C4x are used to filter (high-pass filter) this interference (cutoff frequency equals 0.05 Hz).
- Elimination of 50 Hz Power Line Interference. The following elements are responsible for removing Power Line frequency: R2x, R6x, R7x, R8x, C5x, C6x, C7x, C8x, U2xA and U2xB. These elements

are composed of double T-type band-stop filter. Double T configuration has two main virtues – it is possible to achieve high quality factor.

- Elimination of muscle noise interference (frequency above 30 Hz). 30 Hz cutoff frequency which is often used in Holter monitoring systems causes signal deformation, particularly P wave and T wave. According to this fact and cardiologist's knowledge low-pass cutoff frequency was set to 125 Hz.

Figure 8 presents analog signal conditioning and filtering system which was designed by author of master's thesis entitled "High Definition Vectocardiograph in 24-Hour Monitoring System".

Presented diagram shows one of the three channels (axis X). Signal conditioning system for other two axes Y and Z is analogical. One common channel named Body was designed for this system. Every channel's signal is based on two electrodes V- and V+ which is a result of Frank's vectocardiographic lead system usage.

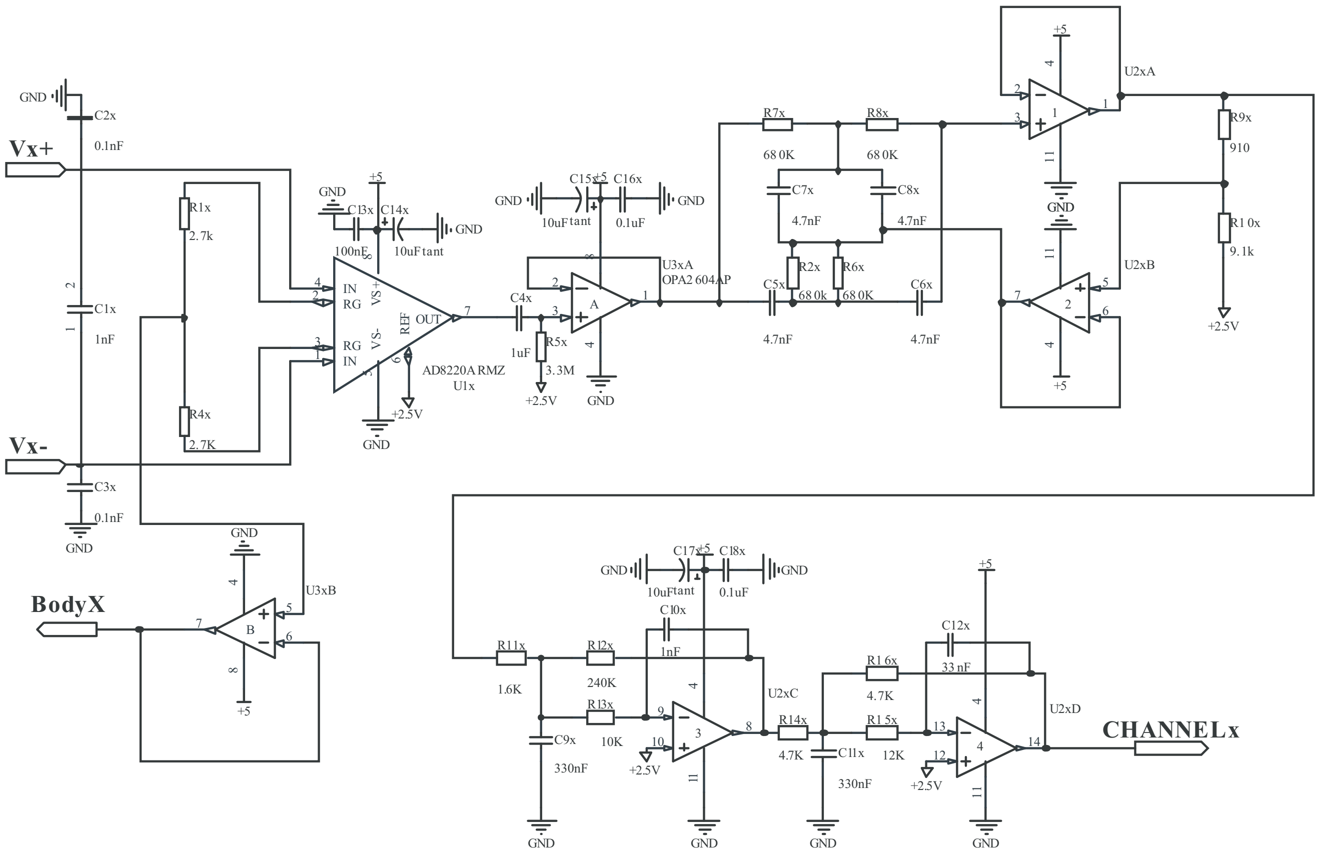

Fig. 8. Block diagram of the analog signal conditioning and filtering system for Holter monitor.

Figure 9 shows the amplitude response and phase response of the filtering system. These two characteristics show that signal conditioning and filtering system was designed correctly and it could be used in Holter monitoring system.

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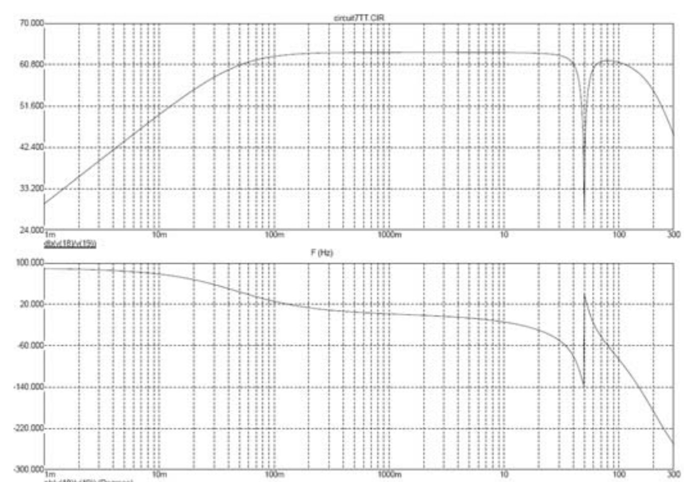

Fig. 9. The amplitude response and phase response of the filtering system.

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