A High Sensitivity Bioimpedance Detector

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***Abstract*-- A bioimpedance detector is developed as part of instrumentation for impedance cardiography. It uses slicing amplifier for increasing the sensitivity for the impedance variation and synchronous sampling for a ripple-free output. The circuit provides digital control of excitation current and frequency used for the measurement. Its operation has been verified using a thorax simulator for detecting the impedance variations well below 2%.**

1. Introduction

Sensing of the variation in the bioimpedance across a body segment can be used for noninvasive monitoring of the changes in the fluid volume or the underlying physiological events. The bioimpedance is generally measured by passing an alternating current through a pair of surface electrodes. The resulting amplitude modulated voltage, sensed using the same or another pair of electrodes, is demodulated to get the impedance signal. In order to avoid any physiological effects, the measurement is carried out using a low-level current (<5 mA) in the frequency range of 20 kHz to 1 MHz [1] - [3]. Higher the frequency used; easier it is to filter out the carrier ripple from the demodulated output. The sensed impedance consists of basal impedance, a time-varying component related to the physiological phenomenon of interest, and various physiological and non-physiological artifacts.

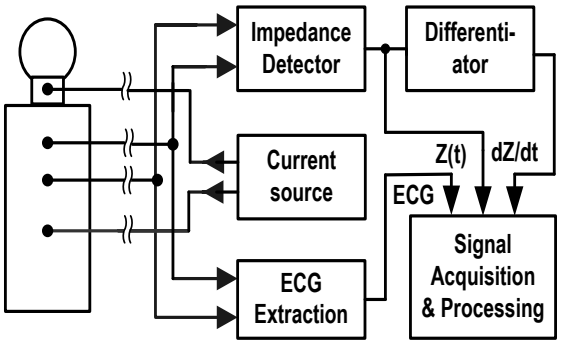

Fig. 1 Block diagram of instrumentation for impedance cardiograph

Impedance cardiography [2] - [6] is a noninvasive method for estimating the stroke volume and some other cardiovascular indices by sensing the variation in the thoracic impedance during the cardiac cycle. An instrument for impedance cardiography generally consists of current injecting electrodes, voltage sensing electrodes, a current source, an impedance detector, and an ECG extraction module, as shown in Fig. 1. A voltage controlled current source can be used for injecting the low amplitude (< 5 mA) current in the thorax. The impedance detector circuit includes a difference amplifier and a demodulator. The time varying component of interest in the impedance is generally less than 2 % of the basal impedance, and therefore the impedance detector should have high sensitivity, external noise suppression, and carrier ripple rejection. A peak detector circuit has low carrier ripple but it is prone to noise. Precision rectifier based circuits [5] [7] give better noise rejection but at the cost of higher carrier ripple. Synchronous detection can be employed for improving the noise rejection but it does not reduce the carrier ripple. Ripple in the output can be reduced by using a lowpass filter, but it introduces phase distortion. For increasing the sensitivity of the detector for input voltage with very low modulation index, Fourcin [8] introduced a slicing amplifier which amplifies the signal above a settable reference. But amplitude demodulation using this circuit increases the carrier ripple in the demodulated output. A sample-and-hold circuit for sampling the output of the slicing amplifier at the peaks of the sinusoidal excitation can be used for realizing a detector with high sensitivity and very low ripple. Use of synchronous sampling also reduces external noise. Summation of the signals obtained by sampling the positive and negative peaks can be used for further reduction of the external noise.

Here we present a bioimpedance detector circuit developed as part of an instrument for impedance cardiography. It uses a slicing amplifier for increasing the detection sensitivity for amplitude modulated signals with low modulation index, and synchronous sample-and-hold for suppression of external noise. Further the carrier ripple is rejected without lowpass filtering. The current level of excitation and the reference voltage for slicing amplifier can be digitally set for selecting the sensitivity for sensing the impedance variation. The frequency for measurement can also be digitally controlled. The circuit provides the variation in the bioimpedance as an analog signal which can be acquired along with other biosignals by a multichannel signal acquisition system for further processing.

2. Bioimpedance Detector Circuit

A block diagram of the instrumentation for impedance cardiography is shown in Fig. 1. The high frequency current is injected through the outer pair of electrodes and the resulting amplitude modulated voltage is sensed across the inner pair of electrodes. The impedance detector consists of a difference amplifier and demodulator. The demodulation is carried out by two channels of slicing amplifier with synchronous sample-and-hold, one channel for the positive peaks and the other for the negative peaks. The outputs of the two channels are added together for giving the demodulated output.

The slicing amplifier is realized using the voltage clamp amplifier IC AD8037 with internal high speed analog switches, as shown in Fig. 2. A dc reference voltage V_X is obtained using digital potentiometer IC1. The lower clamp level V_L of the voltage clamp amplifier IC3 is set to

$$V_Y = \frac{R_7}{R_7 + R_6} V_X \tag{1}$$

For $V_I > V_Y$, the internal switching in the IC connects the input at terminal 3 as the non-inverting input, while for $V_I < V_Y$ the input at terminal 5 is connected as the non-inverting input. Thus the output is given as

$$V_{O1} = \begin{cases} V_I\left(1+\frac{R_3}{R_4}\right) - \frac{R_3}{R_4}V_X, & \text{for } V_I > V_Y \\ V_Y\left(1+\frac{R_3}{R_4}\right) - \frac{R_3}{R_4}V_X, & \text{for } V_I < V_Y \end{cases} \tag{2}$$

With $R_7 = R_3$ and $R_6 = R_4$, Eq. (2) can be written as

$$V_{O1} = \begin{cases} \left(V_I - V_Y\right)\left(1+\frac{R_3}{R_4}\right), & \text{for } V_I > V_Y \\ 0, & \text{for } V_I < V_Y \end{cases} \tag{3}$$

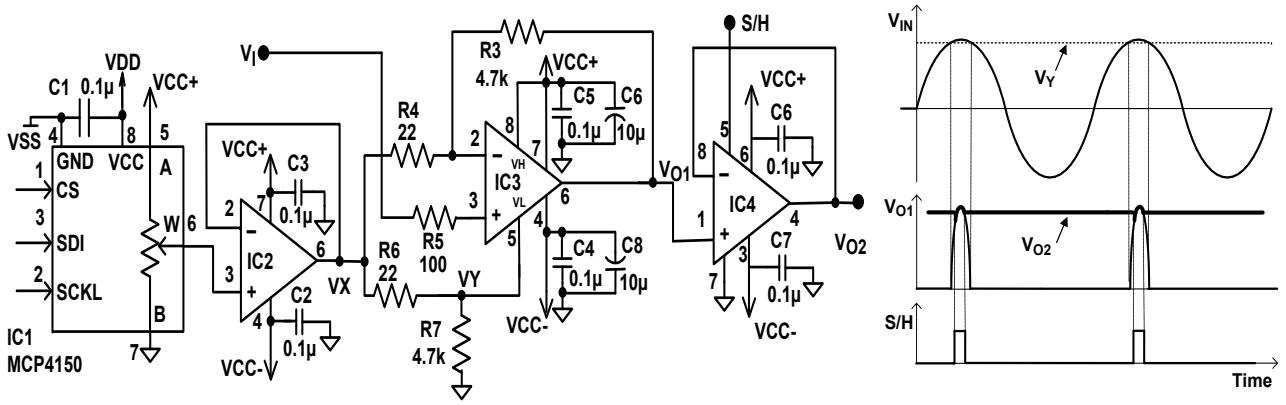

Fig. 2 Demodulator using slicing amplifier and sample-and-hold circuit (IC1: MCP4150, IC2: LF356, and IC3: AD8037, IC4: HA5351) and the output of the slicing amplifier.

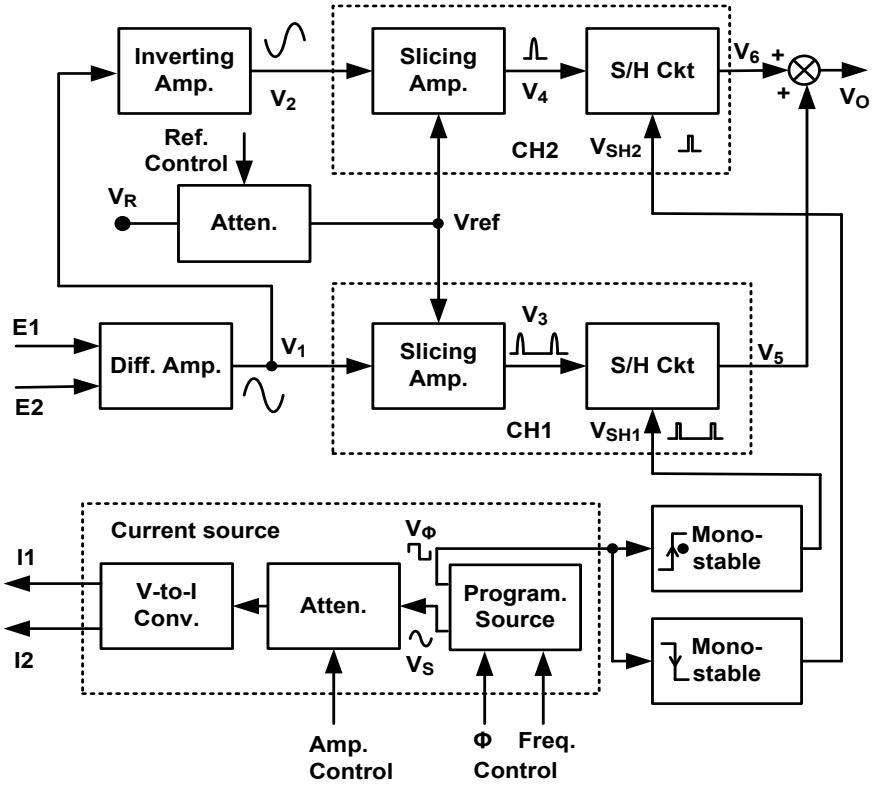

Fig. 3 Bioimpedance detector with two channel slicing amplifier and sample-and-hold circuit

Thus the circuit works as a slicing amplifier with a reference of V_Y and the gain of $1+R_3/R_4$. The figure also shows the waveforms at the input and the output of the slicing amplifier. With the resistance values as given in the figure, the circuit has a voltage gain of around 210 and it gives satisfactory operation up to 500 kHz. The output of the slicing amplifier is sampled near the peak of the excitation current, using the sample-and-hold IC4 (HA5351). The width of the S/H pulse should be very narrow to minimize the ripple and the hold edge should be closely aligned to the peak of the slicing amplifier output.

The amplitude demodulation of the sensed voltage is carried out by two channels of slicing amplifier with synchronous sample-and-hold as shown in Fig. 3. In the first channel, the positive peaks of the output of the difference amplifier above the set reference are amplified. In the second channel, the input signal is inverted and thus the output corresponds to the negative peaks. The sample-and-hold circuit in each channel samples the output of the slicing amplifier at the instant of the corresponding peak, giving a ripple-free output. The outputs of the two channels are added together for suppression of noise and low frequency drift in the input.

For obtaining the sampling pulses for the two channels, a sinusoidal source with two outputs with a

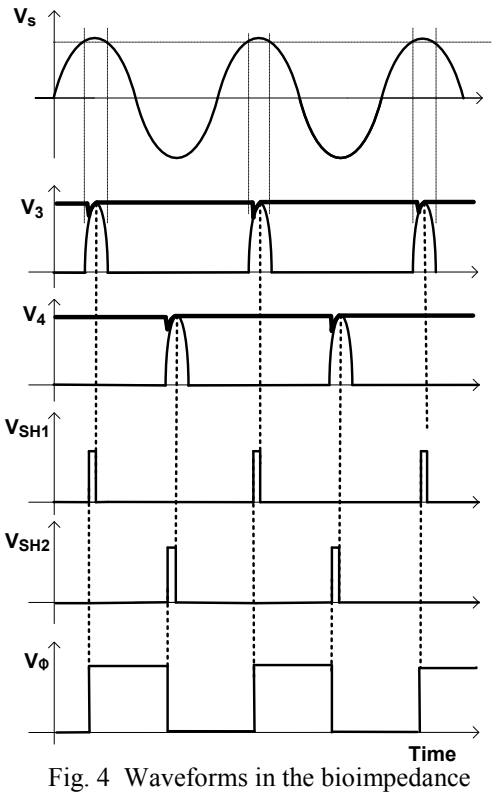

Fig. 4 Waveforms in the bioimpedance detector circuit of Fig. 3.

programmable phase shift is used [9]. It is realized using two direct digital synthesizer chips (AD 9834). The sinusoidal output of the first synthesizer is used as the input to the voltage-to-current converter for current excitation. The amplitude level is controlled by a digital attenuator. The square wave output of the second synthesizer is used for obtaining the sampling pulses for the two channels, by using monostables triggered at the positive and the negative edges. The phase shift between the outputs of the two synthesizers is set for aligning the holding edge of the sampling pulses to the peaks of the sensed voltage. The two synthesizers use the same input clock and after loading frequency and phase commands, are started at the same instants. The relationship between the various waveforms is shown in Fig. 4. A microcontroller is used to control the two digital synthesizers for setting the phase shift between the outputs and the desired frequency. It also controls the digital attenuator to set the excitation current level as well as the digital attenuator to set the appropriate reference voltage for the slicing amplifiers. Thus the circuit provides digital control of frequency and current level of excitation, and the demodulation parameters.

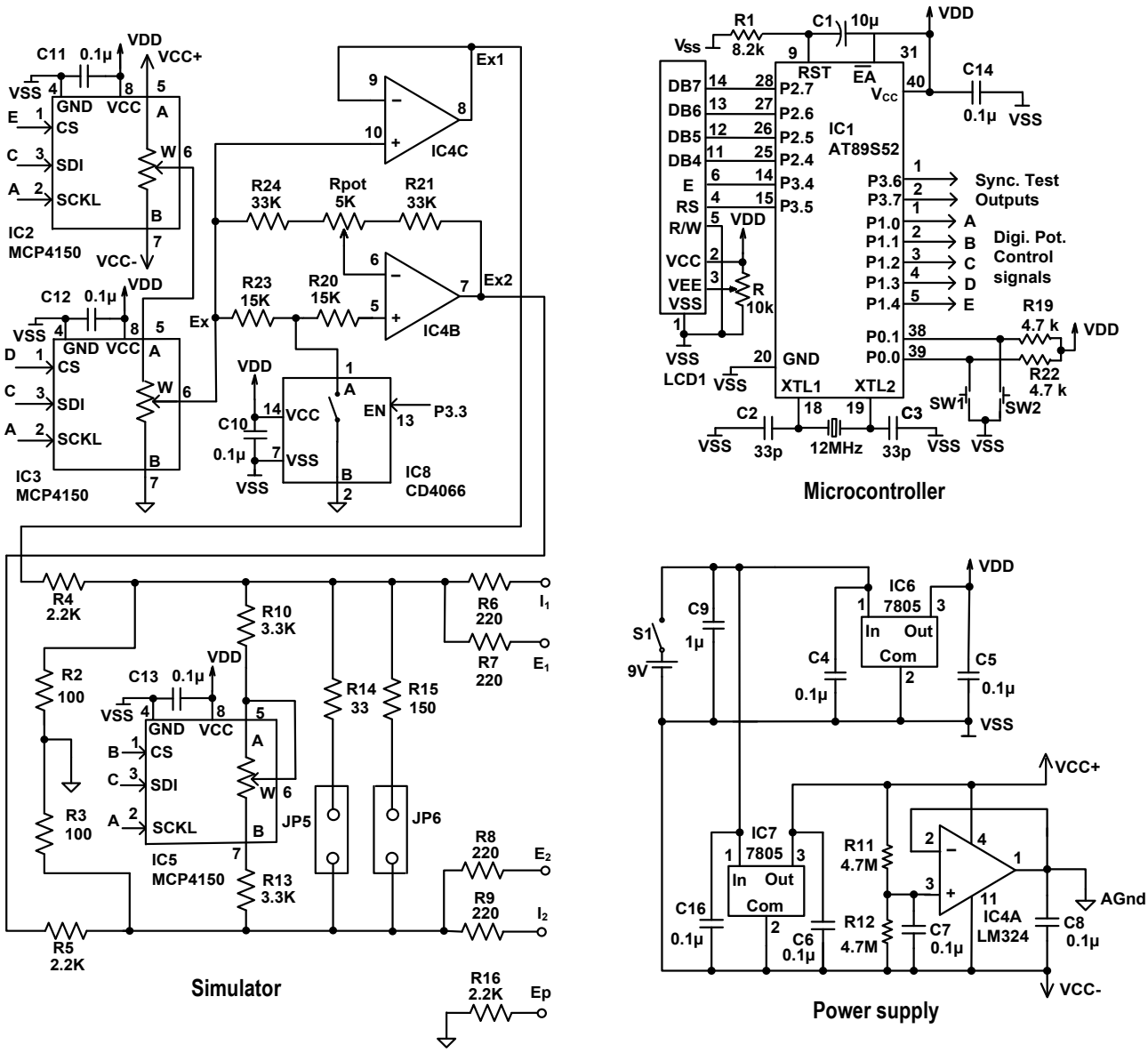

Fig.5 Thorax simulator for testing of the bioimpedance detector

3. Test Results

A bioimpedance simulator [10], [11] can be used for measuring various performance parameters of an impedance detector: the range of basal resistance, sensitivity, and frequency response. For this purpose, we have developed a thorax simulator which provides a basal resistance (settable : 20 Ω – 200 Ω) with a periodic resistance variation (settable: 0.1 – 1.2 %) [9] [11]. As shown in Fig.5, terminals I1 and I2 are the current injection terminals and the voltage drop is sensed across the terminals E1 and E2. The resistance variation is realized using the digital potentiometer IC5 (MCP4150-502E/P). The base resistance values can be selected through jumpers by connecting R14 and R15. Use of a microcontroller and a digital potentiometer permits the selection of the frequency (1 – 250 Hz) and waveshape (square, sinusoidal) of the variation in the resistance. The resistance is varied in accordance with the selected waveshape by providing the appropriate sequence of digital control words to the digital potentiometer. The microcontroller provides "Sync. Test" square wave output in synchronism to the impedance variation cycle.

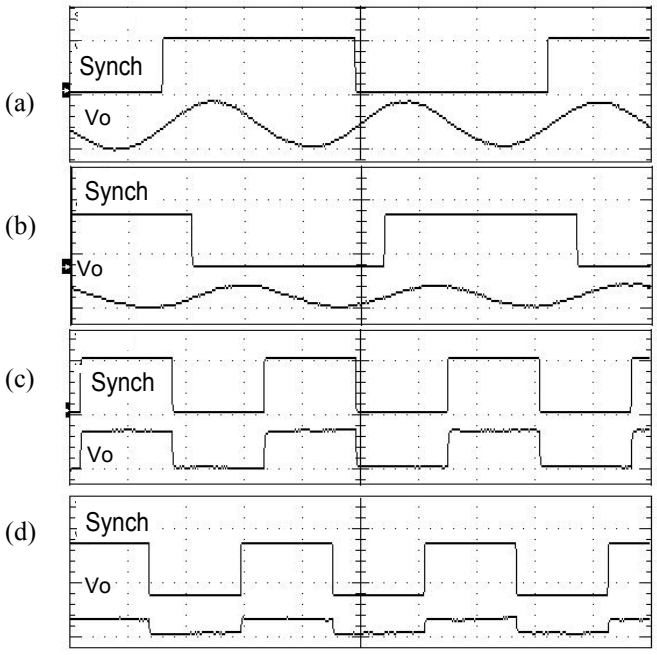

Fig. 6 Waveforms from measurements on the thorax simulator. Upper trace: synchronization test output of simulator, Lower trace: output of the impedance detector (excitation frequency = 100 kHz and current = approx. 1 mA) for f = 8 Hz, R = 196 Ω. and. resistance variations of (a) 1.2 % sinusoidal, (b) 0.6 % sinusoidal, (c) 1.2 % square, (d) 0.6 % square. Time scale: 40 ms/div, Ch1 scale: 5 V/div, Ch2 scale: 500 mV/div.

The bioimpedance detector presented in the previous section has been extensively tested using the thorax simulator, for frequencies of the impedance variation ranging from 1 Hz to 250 Hz and impedance variations well below 2%. Outputs for the excitation current of approximately 1 mA and frequency of 100 kHz as captured using a DSO are shown in Fig. 6, for basal resistance of 196 Ω and resistance variation in the range of 0.1 to 1.2%. The figure shows the detector output along with the "Sync. Test" output of the thorax simulator. We see that the output of the impedance detector tracks the variation in the resistance and the carrier ripple has been rejected without lowpass filtering the output.

4. Conclusion

A bioimpedance detector is developed as part of instrumentation for impedance cardiography. It uses slicing amplifier for detection of very low modulation index and synchronous sampling for ripple-free output. The circuit provides digital control of frequency and current level of excitation and sensitivity for measurement of the variation in the bioimpedance. The circuit operation has been verified using a thorax simulator for detecting variations well below 2%.and for the impedance variation frequencies from 1 Hz to 250 Hz. As the circuit does not require lowpass filtering for rejection of the carrier ripple, it faithfully detects the variation in the bioimpedance.

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