Two-electrode Impedance-sensing Cardiac Rhythm Monitor for Charge-Aware Shock Delivery in Cardiac Arrest

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***Abstract* — This paper presents a two-electrode cardiac rhythm-monitoring amplifier that also senses the impedance between the same electrodes. Online monitoring of the body and electrodes impedances during high-voltage (HV) energy delivery or defibrillating, is needed to precisely estimate the amount of charge being delivered to the cardiac muscle. By adjusting the HV stimulus waveform, the amount of charge and thus energy delivered to the cardiac muscle can be precisely controlled for each patient. Additionally, online monitoring of the cardiac impedance ensures proper electrodes connectivity during both electrocardiogram (ECG) signal monitoring and HV energy delivery. The proposed amplifier employs two current sources of the same value and opposite direction (one at each electrode) to adaptively source through the chest an out-of-band (1 kHz) sinusoidal current and measure the resulting voltage to infer the cardiac impedance. This is done in superposition with ECG monitoring due to the same modality of the signal (i.e. voltage). A functional prototype with low-cost off-the-shelf components was implemented and tested demonstrating the feasibility of the proposed technique.**

I. Introduction

The principle of operation of most commercial defibrillators (Fig. 1(a)) is based on the delivered energy, meaning that the device charges a capacitor to a predefined voltage and then delivers a specified amount of energy to the body. The amount of energy, which arrives at the myocardium (Fig. 1(b)), is dependent on the selected voltage and the transthoracic impedance (which varies patient-to-patient). The optimum amount of energy delivered to the myocardium has been suggested as a major factor for successful defibrillation [1].

In critical situations, it is important to deliver the right amount of energy without having to deliver the second shock. The probability of the first shock success depends on how effectively a defibrillator delivers the right amount of current. This defibrillation current must be large enough to be therapeutic but not so large as to be harmful. A low defibrillation current may be ineffective; however a high current has the risks of an overdose. Given a specified amount of energy, the impedance yields the amount of current, or the amount of the charge passing through the body per unit time. Hence, it is important to monitor the impedance of the body and of the in-path electrodes during the energy delivery. As Fig. 1(c) shows, generating a high-voltage (HV) biphasic waveform, combined with automatic measurement of the body impedance is the basis for modern defibrillators.

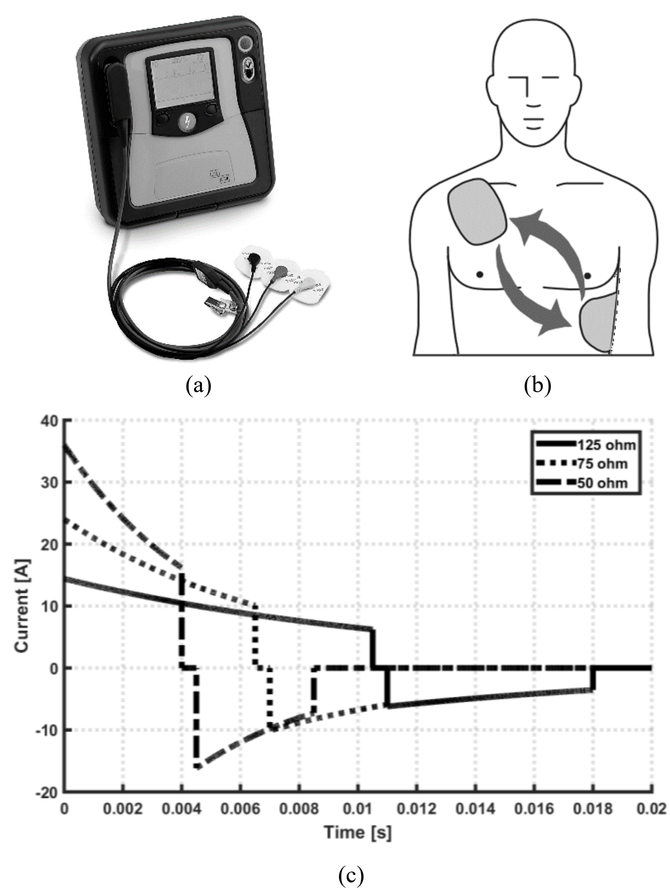

Figure 1(a): An example of a commercial external defibrillator. (b): The defibrillation current flows through the body by means of two electrodes. (c): The defibrillation current varies with the impedance between the electrodes [2].

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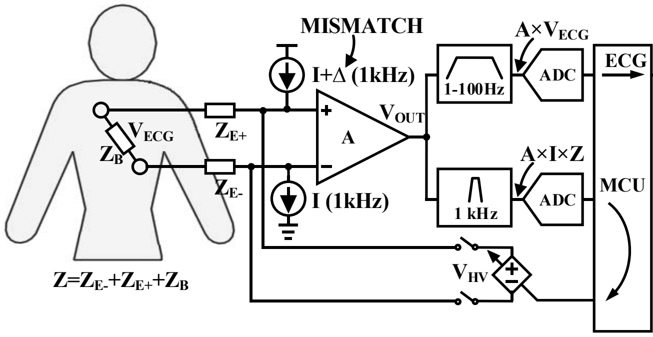

Figure 2: Block diagram of a two-electrode cardiac rhythm-monitoring defibrillator with online impedance sensing for precise charge delivery during high-voltage shock in cardiac arrest. The intrinsic mismatch between the two current sources leads to an impedance measurement error in this open-loop circuit.

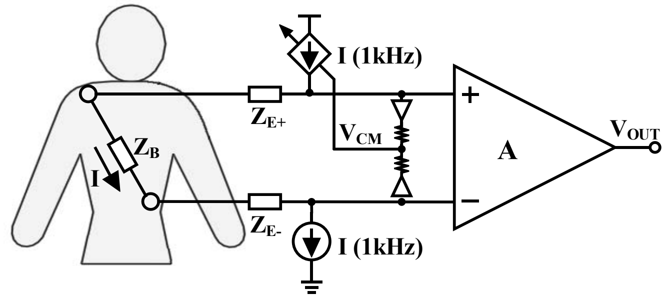

Figure 3: A CMFB loop senses the current mismatch and adjusts one current source to match the other for an accurate impedance measurement.

Measuring the impedance of the body and of the two electrodes without either using extra electrodes or degrading the electrocardiogram (ECG) signal is very challenging. Conventional ECG amplifiers employ at least three electrodes in various applications. Some diagnostic ECG monitoring applications require as many as twelve leads, but for most other monitoring applications like ECG monitors in intensive care and operating rooms, Holter and arrhythmia monitors, automatic defibrillators, and biotelemetry, a single extra lead (three electrodes) is sufficient – two for differential signal amplification and one for the reference [3]. Many monitoring applications can benefit from having as few electrodes attached to patient as possible in order to minimize the cost and preparation time. In addition, eliminating the reference electrode removes the hazardous leakage currents but reduces the common-mode rejection (CMR) of the front-end amplifier. This is mostly because a common-mode voltage (CMV) induced by power line or by an electrostatic charge is transformed into a differential voltage due to an imbalance of electrode-body impedance and could possibly saturate the low-voltage biopotential amplifier front-end.

ECG signal monitoring done simultaneously with cardiac impedance measurement has the advantage of using the impedance information between the electrodes. This information is well-correlated with motion artifacts and can be utilized in post-processing adaptive ECG filters and in automatic defibrillators (Fig. 2). Also, this ensures system reliability by continuously checking the electrode connectivity to the patient, lead-off detection and electrodes quality assessment.

Generally, to obtain the front-end impedance, a current is applied to the body and the corresponding voltage drop at a fixed frequency is measured. Since the cardiac activity is monitored in the voltage domain, this method has the benefit of using the same signal modality. However, placing two high-impedance serially-connected current sources at the two-electrode amplifier input without setting the body CMV (i.e. no third electrode) saturates one of the current sources due to the intrinsic mismatch between the current sources. Hence, some designers use three or higher electrode-count amplifiers to set the reference [4] or use programmable current sources with a reset capability when the CMV of the front-end amplifier reaches either of the two supply rails [5]. Time multiplex rather than real-time impedance monitoring is another obvious solution that has the drawback of less accuracy and recording interruption.

Two-electrode amplifiers have also been extensively studied. Some two-electrode techniques use a bootstrapped input stage to obtain high input impedance [6], [7]. Making use of phase-sensitive detection to balance the input impedances of one of the two differential input terminals of the two-electrode biopotential amplifier was introduced to control the input impedance [8]. A direct interference-canceling scheme was introduced to reject interference for a two-electrode biopotential amplifier to cancel differential-mode interference directly by adding a feedback signal to the interference [9]. A circuit using chopper-stabilized DC current sources rather than sinusoidal current sources to measure the signal source impedance in a ECG readout channel was also introduced [8]. Making use of band-stop or digital filters can also reduce interference, however the shape of the ECG signal is distorted due to the attenuation of important frequency components in the QRS complex.

II. Circuit Architecture

To measure the ECG signal source impedance, two 1kHz-tone current sources are utilized as shown in Fig. 2 – one for sourcing the current $(I + \Delta)$ and one for sinking the current (I). The frequency of the interrogating current sources is set at 1kHz in order to obtain the required accuracy in the impedance measurement and to avoid higher-frequency band-pass filtering. The current passing through the body and electrodes impedance $(Z = Z_B + Z_{E+} + Z_{E-})$ induces a voltage which is filtered and amplified by a band-pass filter and a programmable gain amplifier (the latter is not shown in Fig. 2), respectively. However, the two current sources are not intrinsically matched (the mismatch is shown by Δ). Since a charge builds up in the tissue, as a result one of the current sources will saturate the other one. In other words, the floating

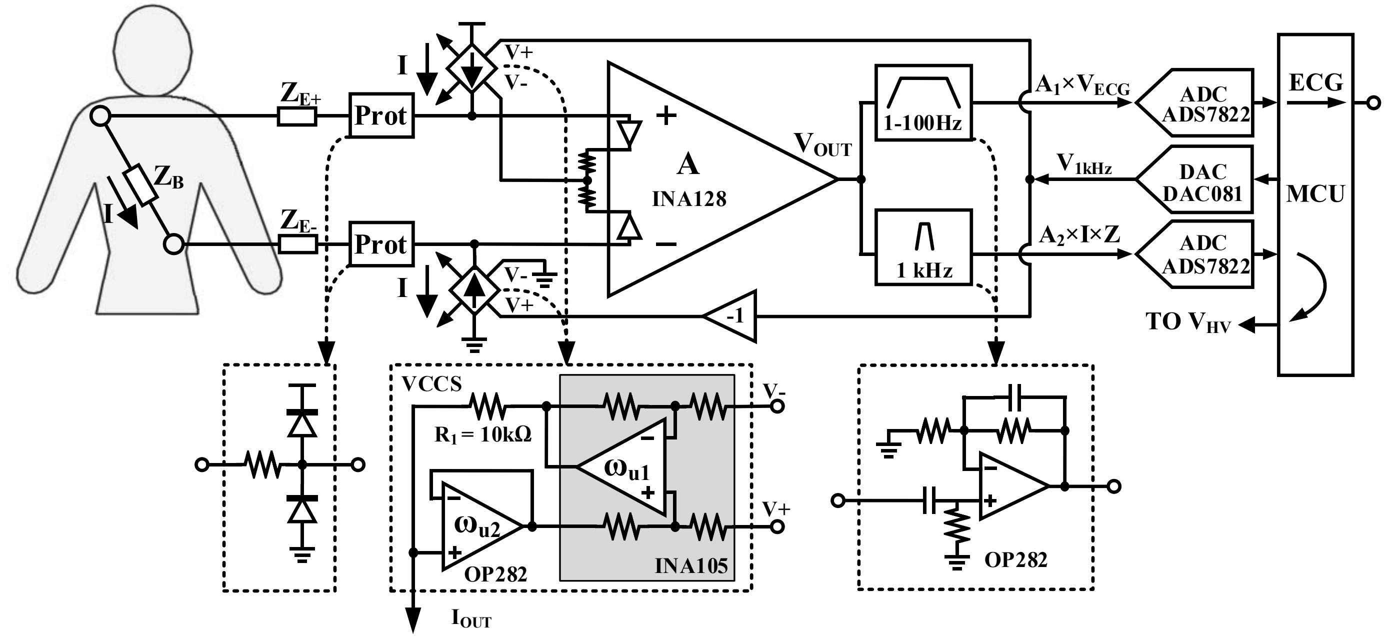

Figure 4: Block diagram of the proposed two-electrode cardiac rhythm-monitoring amplifier with online impedance sensing.

CMV at the interrogating current frequency band will drift to one of the supply rails.

To overcome this problem a common-mode feedback (CMFB) loop senses the CMV of the body impedance at the impedance measurement frequency (1kHz) and adjusts the current sources as shown in Fig. 3. The CMV is measured by taking the average of the buffered electrode voltages with two equal resistors. To ensure that the CMFB loop has enough inverting gain, an amplifier/filter is placed after the resistors (not shown in Fig. 3).

III. Circuit Implementation

Fig. 4 depicts a detailed implementation of the proposed circuit using low-cost microamp-power off-the-shelf components. The main part of the circuit consists of a classical instrumentation amplifier (INA128) to provide high CMR and high input impedance for the amplifier. A programmable digital-to-analog converter (DAC) generates a 1kHz sinusoidal voltage. Two voltage-controlled current sources (VCCS) are used to convert the voltage to the 1kHz interrogating currents (for simplicity, one is shown in Fig. 3).

A. *Current Sources*

The simplest technique to obtain a constant current is to use a VCCS, which can be defined as a combination of a positive and negative feedback around a high gain operational amplifier. This VCCS is implemented as Howland current source [5], which is a versatile bipolar VCCS capable of sourcing and sinking a current. This circuit requires six highly matched resistors to obtain a high output impedance. If all the resistors are accurately matched the output impedance of the VCCS becomes theoretically infinite, however the intrinsic mismatch and parasitic capacitance at the output node limit the output impedance, especially at the interference frequency. In addition, since the two current sources are directly connected to the input terminals of the amplifier, their output impedance is in parallel with the input impedance of the amplifier. Therefore, any impedance mismatch limits the amplifier's CMR due to the common-mode to differential conversion.

In Fig. 4, one possible solution to keep the output impedance of the VCCS as high as possible rather than employing precisely matched discrete or integrated components is shown [11]. Equations (1) to (4) represent the output current, the transconductance, the output equivalent resistance and parasitic capcitance of the VCCS, respectively. The output impedance of the current sources at the interference frequency is substantially reduced due to the output parasitic capacitance of the VCCS. The parasitic capacitance is limited by one of the two op-amps bandwidths. However by selecting a wider-bandwidth external op-amp compared to INA105, this parasitic capacitance becomes equal to 100 pF and yields the output resistance of approximately 35 MΩ at the frequency of 50 Hz.

$$I_L = \left(\frac{1}{R_1}\right)(V_+ - V_-) \tag{1}$$

$$g_m = \left(\frac{1}{R_1}\right) \tag{2}$$

$$Z_{out} = R_{out} + \frac{1}{sC_{out}} \approx R_1 + \frac{R_1 \omega_{u1} \omega_{u2}}{(2\omega_{u1} + \omega_{u2})s} \tag{3}$$

$$\Rightarrow R_{out} = R_1 \quad and \quad C_{out} = \frac{2}{R_1 (2\omega_{u1} \parallel \omega_{u2})} \tag{4}$$

where ω_u is the unity-gain bandwidth of the opamp and C_{out} is the equivalent output parasitic capacitance.

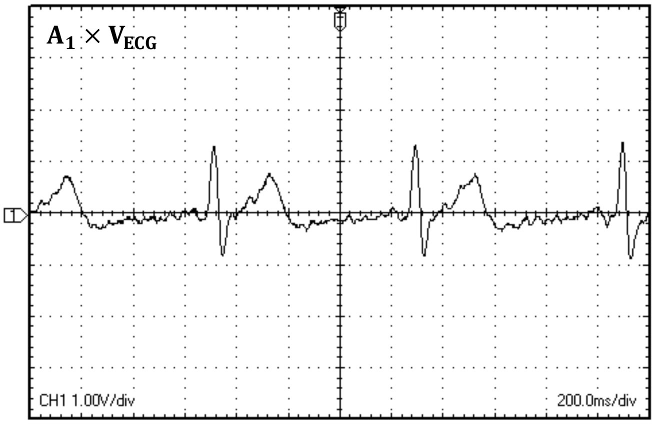

Figure 5: ECG signal experimentally measured by the implemented two-electrode amplifier.

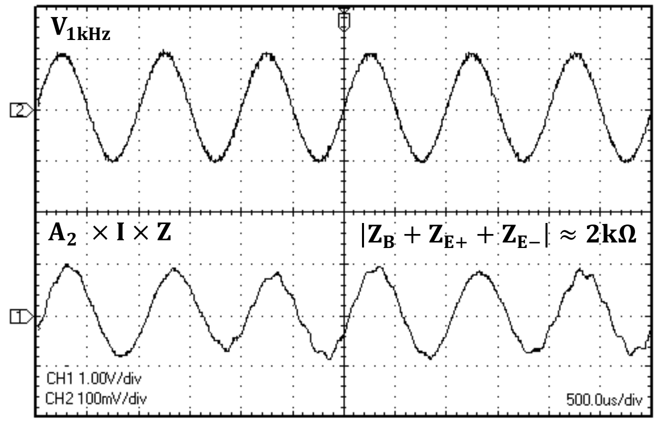

Figure 6 (bottom): Body-electrode impedance-dependent output voltage experimentally measured by the implemented two-electrode amplifier. (top): The VCCS input sinusoidal voltage.

B. *Other Circuits*

To protect the amplifier from transients and the HV defibrillation signal, a protection circuit is placed at the two input nodes as shown in Fig. 4. The in-path resistance of the protection circuit is known and is subtracted from the body/electrodes impedance by the microcontroller. Each filtering block in Fig. 4 models an analog band-pass filter and a programmable gain amplifier (not shown) to adjust the amplitude of the signal to match the analog-to-digital converter (ADC) input range.

IV. Experimental Results

The inputs of the implemented amplifier were connected with a pair of non-screened wires and commercial ECG electrodes (not defibrillator electrodes) to a subject. Fig. 5 shows the measured ECG signal. The obtained ECG signal is slightly contaminated with 50/60Hz interference mainly due to the input terminal impedances mismatch that transforms the CMV to differential voltage. A higher-order analog or digital filter can easily remove this interference signal.

Fig. 6 shows a response of the amplifier to the injected 1kHz currents with an amplitude of 10μA. The obtained impedance signal shows no trace of a floating CMV.

V. Conclusion

This paper presents a two-electrode amplifier with real-time body and electrodes impedance measuring capability. The amplifier employs two sinusoidal current sources to interrogate the electrodes/body impedance. Conventional two-electrode amplifiers are highly susceptible to CMV saturation. A CMFB keeps the CMV of the body constant and avoids saturating amplifier by any mismatch in the injected currents. The current sources are implemented with the popular Howland VCCS structure with one integrated circuit and a buffer in a positive feedback path to operate with only one external resistor. To minimize the sensitivity of the output impedance of the VCCS to intrinsic mismatch of components, which lead to a CMV division in the input terminals, the parasitic capacitance of the VCCS is kept as low as possible. The proposed circuit was experimentally demonstrated for use in defibrillator systems, but also finds many applications in the field of high-accuracy physiological monitoring.

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