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A Modified Howland Current Source Design for Simultaneous EIT/ECG Data Acquisition

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Abstract

The current source is one of the most critical circuits in electrical impedance tomography (EIT) hardware systems. The simplicity and excellent performance of the Howland current source makes it a prime candidate for this role in EIT systems. Although the Howland source and its family may be the best option for the high-frequency EIT operation, its low frequency noise may also limit the implementation of a system to simultaneously collect electrocardiogram (ECG) and EIT signals from the electrodes. This paper proposes modifications to the conventional Howland source to make is suitable for simultaneous EIT and ECG. The preliminary experimental results of this modified Howland show significant improvement in the collected ECG signal quality in the presence of the EIT signal.

Keywords

EIT; ECG; Howland source

1 Introduction

A large set of important physiological parameters and vital signs can be measured from the chest. For example, chronic heart diseases can be diagnosed via ECG measurements (Bashi *et al.* 2017). Some other studies showed the ability to assess relevant vital signs of asthma or cystic fibrosis using EIT images (Frerichs *et al.* 2016). So, combining and relating the synchronized EIT and the ECG measurements may provide even a more profound insight into the relationship between lung ventilation events and cardiac activities. A robust, simultaneous EIT/ECG signal acquisition system is required for these applications. This system should be able to provide an acceptable resolution for both signals to be used for diagnostic purposes.

In this paper, we will be continuing and building upon the simultaneous EIT/ECG acquisition approach presented in (Abdelwahab *et al.* 2021). The approach integrates the combined EIT/ECG signal over an integral number of cycles of the EIT excitation frequency

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to produce the nearly complete suppression of the EIT signal. The experimental results shown in that paper validated the ability of the proposed approach to isolate synthesized ECG signals in the presence of EIT signals. Further investigation, however, showed that performance degraded significantly when the source of the ECG signal had a high output impedance, which is the case with actual ECG signals. Specifically, the recovered ECG signals became unacceptably noisy as the source impedance for the ECG signal was increased. This degradation was traced to the Howland current source (Franco 2003) that is used to supply the EIT excitation, leading to further exploration of the current source behaviour in the ECG band near DC. The scope of this paper will mainly focus on improving the Howland circuit design used for EIT imaging to enhance the recovered ECG waveforms. The proposed modification of the Howland circuit includes adding a series feedback capacitor in the positive feedback path of the Howland to reduce the injection of low frequency noise and the use of a series capacitor on the output to reduce the loading of the ECG signal. The sections below detail the modifications and show the performance improvement that they provide.

2 System Requirements

For excellent EIT performance, it is required that the current source maintain a very high output impedance over the EIT operation bandwidth. For a simultaneous EIT/ECG system, this high output impedance range must be expanded to cover the ECG bandwidth, which is from 100 mHz – 150 Hz (Bashi *et al.* 2017), orders of magnitude below the typical EIT excitation range of 10 kHz to 1 MHz, to avoid loading the ECG signal. Obtaining a high output impedance at low frequencies is not difficult with a Howland source; in fact, it is easier than at higher frequencies where output capacitance degrades performance. However, the output current noise of the Howland source, which is higher near DC due to the $1/f$ noise introduced by the op amp, will flow through the high low-frequency output impedance to produce a significant voltage noise. Given the small amplitude of the ECG signal, generally on the order of 1 mV or less, this noise voltage can severely degrade the recovered ECG signal. The challenge, then, is to modify the current source to provide high output impedance over the EIT and ECG frequency range while also maintaining low noise near DC.

3 Modifications to Howland circuit

The standard Howland current source, shown at the left in Figure 1, ideally can achieve an infinite output impedance by balancing its resistor bridge network (Franco 2003). The modified version of the Howland is shown at the right in the figure. The first modification introduced is moving the voltage source driving the Howland from R_1 to R_3, which makes the circuit perform as an inverting voltage-to-current converter. This change provides an extra degree of patient protection by electrically isolating the voltage source driving the Howland circuit from the Howland source output attached to the patient.

The second modification to the standard Howland circuit is adding the capacitor C_1 into the positive feedback path. The insertion of this capacitor has double benefits. First, it blocks any path for a DC signal to propagate to the Howland output attached to the patient. Second, it reduces the noise generated by the Howland in the ECG bandwidth by unbalancing the resistor bridge at low frequencies, reducing the source output impedance. The value of C_1 is large enough that, at high frequencies in range of the EIT excitation, it presents a very low impedance relative R_2 and has little effect on the resistor balance. At low frequencies, however, the impedance presented by C_1 becomes large and, acting in series with R_2, makes the circuit unbalanced and reduces the output impedance. A lower output impedance for the source results in the output noise current producing a smaller noise voltage.

The output impedance results shown in Figure 2 were obtained from LTSpice (Analog Devices) simulation of the Howland circuit implemented using the low noise Analog Devices AD8033 op amp, 2 $k\Omega$ resistors for R_1 and R_3, and 0.5 $k\Omega$ resistors for R_2 and R_4. The load is a $100\ k\Omega$ resistor and this load resistance is considered to be part of the output impedance since it will contribute to the conversion of the output current noise into a voltage noise. The results on the left do not include C_1 while those on the right use a $10\ \mu\mathrm{F}$ capacitor for C_1. Each figure shows the output resistance and output capacitance when the output impedance is modelled as a parallel RC network. The capacitor C_1 is nearly an open circuit at low frequencies, and the results show that it greatly reduces the both the output resistance and capacitance at frequencies below 1 kHz while having little effect at higher frequencies.

Although the reduction in output impedance at low frequencies obtained adding the capacitor C_1 is beneficial for reducing noise in the ECG band, this same low output impedance will make the Howland output load down the ECG signal. The IEC-60601 medical standards for ECG recording indicates that the input impedance of the ECG buffer should not be less than $10\ M\Omega$ to provide a diagnostic ECG recording (Young & Schmid 2021). This high value of input impedance would be very difficult to achieve in a practical implementation of simultaneous EIT/ECG, but a reasonably high value is needed to recover an acceptable ECG waveform.

The addition of C_2 as shown in the diagram on the right in Figure 1 is one solution to the problem. If C_2 is a sufficiently small capacitor, it will introduce a reasonably high series impedance at the ECG frequencies, helping to isolate the Howland output from the ECG signal. In our experiments we have used $C_2 = 10\ \mathrm{nF}$. With this configuration, the overall output impedance Z_{out} seen from the load side, i.e. looking back from the electrode, is given by

$$Z_{\mathrm{out}} = \left(\frac{1}{j2\pi \mathrm{f} C_2} + Z_{Howland}\right) \Big\| Z_{Load}$$

Where $Z_{Howland}$ is the intrinsic Howland output impedance, $Z_{Load}$is the load impedance shown on the right in Figure 1, and $\|$ indicates that the impedances are in parallel.

From Equation (1), the Howland output impedance $Z_{Howland}$ with capacitor C_1 in the ECG range is low (as shown in Figure 2 Right), which results in a smaller noise voltage. While the overall impedance seen from the load side Z_{out} has a much larger magnitude of about

108 $k\Omega$ at 150 Hz due to the 10 nF capacitor C_2. So, this configuration has provided a lower voltage noise while reducing the ECG signal loading.

4 Experiments and Results

A combination of simulation and experimental setups were used to verify the validity of the proposed modification to the Howland design. The aim is to prove how these modifications enhanced quality of the recovered ECG signal with the simultaneous system in terms of noise reduction and reducing the loading effect caused by the DC blocking capacitor C_1.

4.1 Noise analysis

The Howland configuration described in the previous section was simulated in LTspice to evaluate its noise performance with and without the DC blocking capacitor C_1 of 10 μF. The resistor values remain the same with 2 $k\Omega$ for R_1 and R_3, and 0.5 $k\Omega$ for R_2 and R_4. For these simulations, the AD8033 op amp has been replaced with a lower noise device, the ADA4896. The new op amp provides about one-fourth of the voltage noise density compared with the AD8033 used earlier, with $2.4\ nV/\sqrt{Hz}$ compared with $11\ nV/\sqrt{Hz}$ at 100 kHz. The total RMS noise at the Howland output for the ECG bandwidth [100 mHz – 150 Hz] was 15.536 μV without C_1, and dropped to 681.78 nV with the presence of C_1. These results make perfect sense, as the capacitor C_1 is nearly an open circuit at simulated range and this configuration provides a low Howland output impedance and hence a smaller noise voltage with the same load connected.

4.2 Recovered ECG signal

The same ECG separation approach discussed in (Abdelwahab *et al.* 2021), along with the EIT system described in (Saulnier *et al.* 2020), was used to experimentally validate the enhancement of the ECG recording with an active EIT signal. A diagram of the experimental setup is shown in Figure 3. A lab-generated ECG waveform with a repetition rate of 1.5 Hz and amplitude of $1\ mV_{pp}$ is fed into the system through a $100\ k\Omega$ series resistor. On the system side of the $100\ k\Omega$, the signal sees the impedance denoted by Z due to the Howland current source as well as an AC-coupled voltage buffer. The RC high-pass filter that precedes the voltage buffer input has a cut-off (−3 dB) frequency of 159 mHz. The impedance Z presented by the Howland can be either Z_{out} from equation (1) when the capacitor C_2 is present or simply $Z_{Howland} \| Z_{Load}$ when the capacitor is replaced by a short. In the experimental setup, $Z_{Load} = 100\ \mathrm{k}\Omega$. The output of the buffer is processed as described in (Abdelwahab *et al.* 2021) to produce one ECG sample per burst of the EIT signal, resulting in a final sampling rate of approximately 1 kHz. The digital signal at a 1 kHz sampling rate is processed by a second-order Butterworth 150 Hz low-pass filter to set the ECG bandwidth and a line frequency notch filter.

Figure 4 shows the impact of including the 10 nF C_2 capacitor. The waveforms show the digital signals after the filtering mentioned above. The waveform on the left is obtained when C_2 is replaced by a short. The signal is very small due to the low output impedance of the Howland source created by the presence of C_1 in its positive feedback path and dominated by noise. In contrast, the waveform on the right shows the improvement caused by using $C_2 = 10\ \mathrm{nF}$. The results confirmed that this capacitor had done its job of reducing the ECG signal loading and maintaining the low noise performance needed. The reconstructed ECG signal is less noisy with the proposed configuration of the Howland compared to the standard circuit and the results presented in (Abdelwahab *et al.* 2021).

5 Conclusions

In this paper, three modifications to the standard Howland circuit design were discussed. These modifications aim to enhance the simultaneous EIT/ECG data acquisition system to improve the ECG data recording. The initial experimental results were promising showing a clean reconstructed ECG signal which can be used for timing alignment with the collected EIT data. The added capacitor C_1 has provided a less noisy current source as it reduced the output impedance and hence, the total RMS voltage noise by about 96% for the same load. Also, it added an extra degree of patient protection by blocking DC from propagating to the patient side. Additionally, the capacitor C_2 has reduced the ECG loading while maintaining the noise performance improvement. In all, the two capacitors have introduced the DC blocking feature and the lower noise performance while reducing the ECG signal loading.

Acknowledgments

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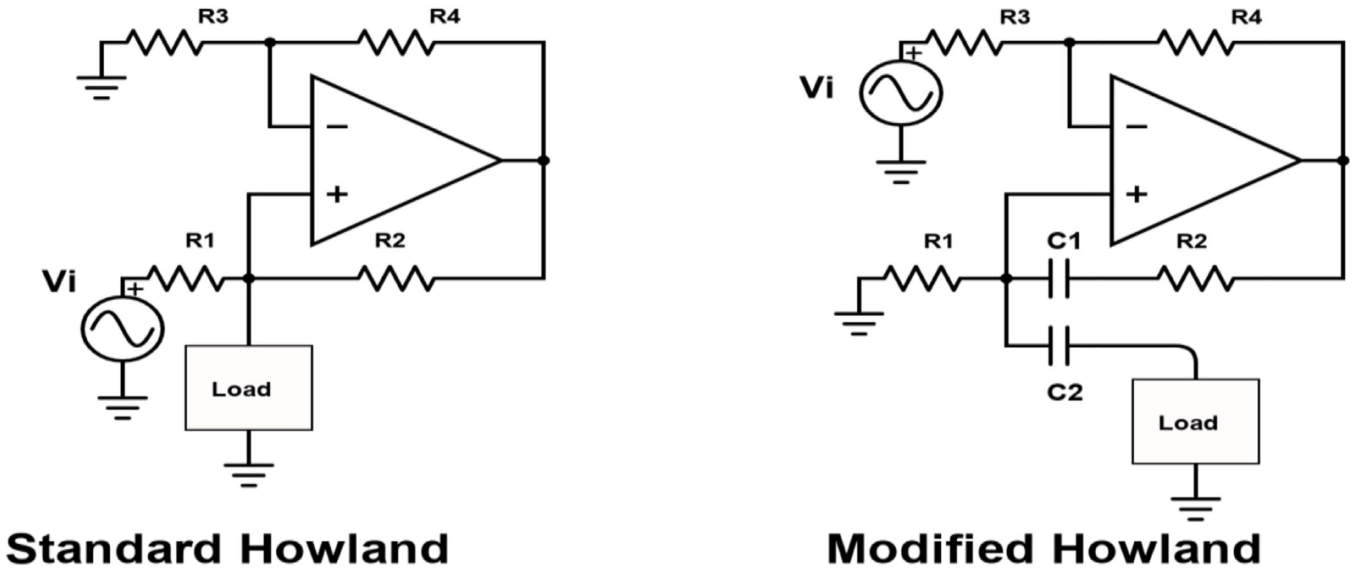
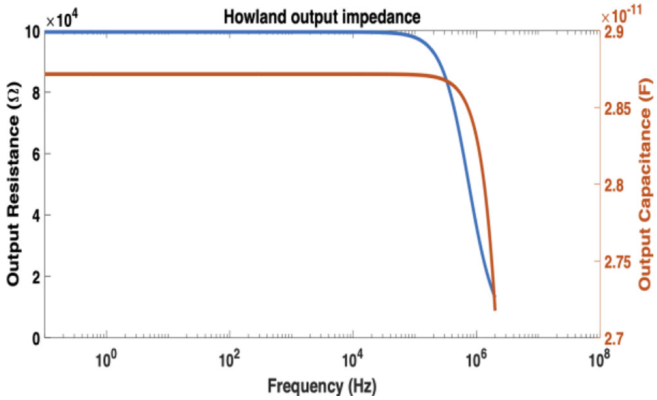


Figure 1.

The Standard textbook version of the Howland source (Franco 2003) is shown vs the proposed modified Howland version.



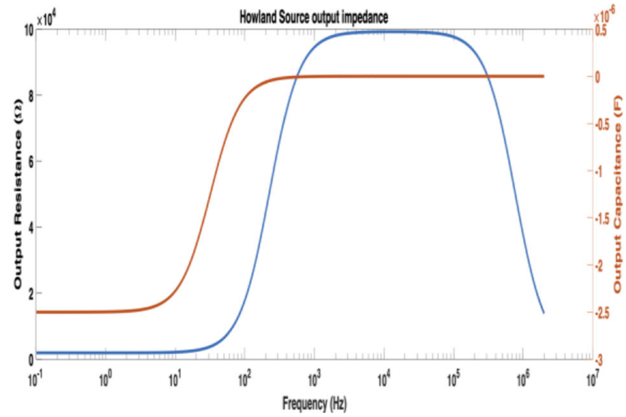


Figure 2.

The figure shows a simulation results of the Howland output impedance, without C_1 (Left), with C_1 (Right).

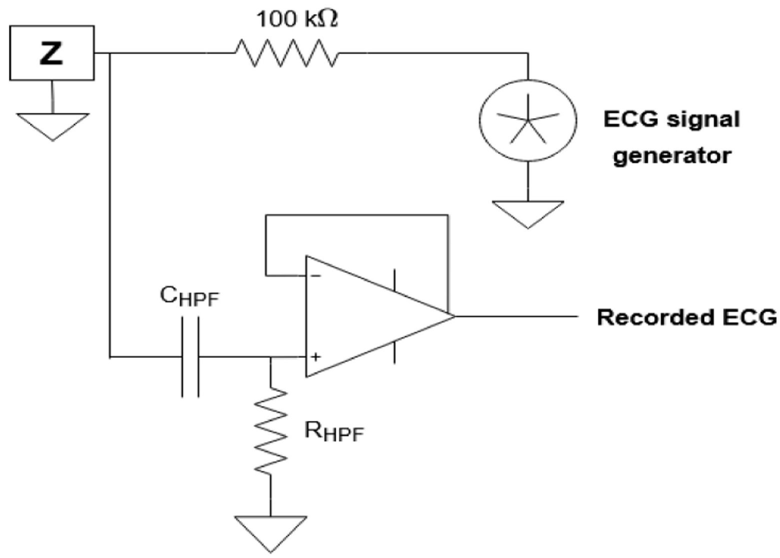


Figure 3.
Experimental setup for ECG recording.

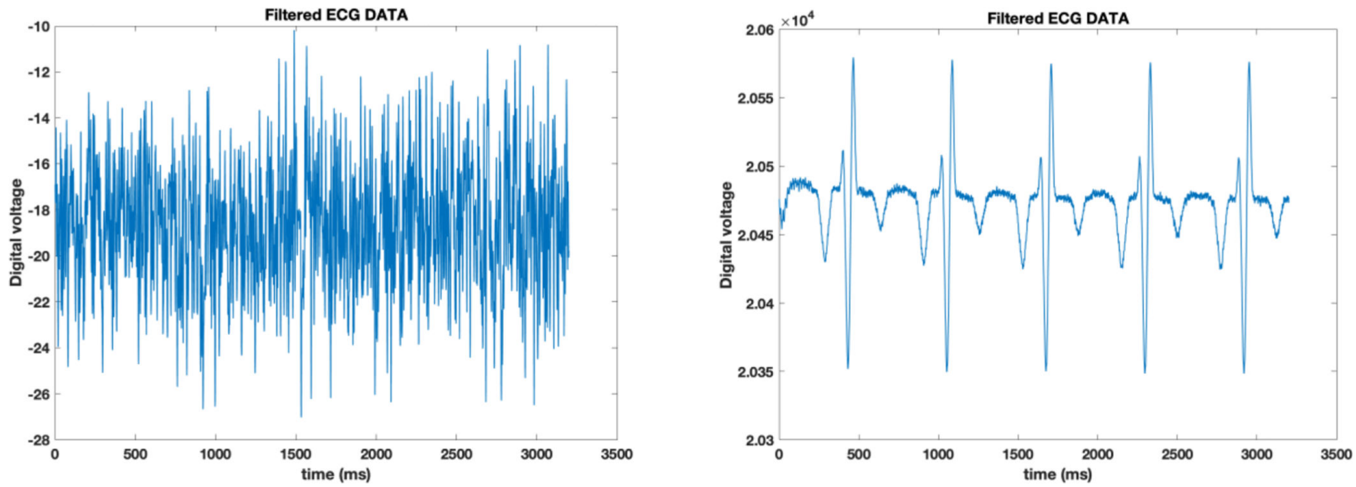


Figure 4.
The reconstruced ECG waveform with the 100 $k\Omega$ in seres without capacitor C_2 (left), with C_2 (right).