

An Improved Very Low-Power Electrode Contact Impedance Monitor

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Abstract. This paper reports the design improvement of a circuit for monitoring the quality of contact of un-gelled stainless steel electrodes with a patient's skin in the measurement of the ECG. The circuit is intended to be incorporated into a heart rate monitor used prior to and during childbirth to measure the mother's heart rate. The revised design uses commercially available low-power technology to implement the circuit with a reduction in the power consumption by a factor of 20 over a previous design [1]. The circuit measures the quality of contact of electrode impedances up to 100k Ω and gives a visual indication when the contact impedance of either electrode is above this level. The impedance is measured using a signal injected through the electrodes via the input impedance of the amplifier at a frequency of 5kHz. The resulting signal present at the amplifier input at this frequency is then extracted and processed before threshold detection of the signal level which is used to indicate whether or not the quality of contact of the patient's fingers with the electrodes is acceptable for the purposes of heart rate measurement.

1 Introduction

The Safer Births Programme [2], sponsored by the Norwegian Research Council as part of a larger World Health Organization initiative [3], is an action plan aimed at eliminating preventable deaths of infants at birth, in developing countries. Many maternity hospitals in these locations are understaffed and are often forced to rely on midwives and nurses alone to contend with the clinical complications that arise around childbirth, without access to the assistance or equipment needed to deal with them. Few places have sufficient equipment for fetal monitoring such that fetuses in distress can be left unattended, adding to the burden of birth asphyxia and stillbirth. Many of the infants born may not be breathing correctly and have pulses which are difficult or impossible to detect manually. On occasions these infants can be misclassified as stillborn, when their hearts are still functioning and they could be resuscitated with suitable equipment. Laerdal Medical AS, based in Norway, designs and markets resuscitation equipment for use in the scenarios described above. One such product is the Moyo unit shown in Figure 1. This unit is a heart rate monitor which uses an ultrasonic transducer placed on the mother's abdomen to detect the infant's heart beat and indicates the fetal heart rate on a LCD. One common problem with ultrasonic heart rate monitors is that when the infant's heart beat is difficult to detect the transducer often picks up the mother's heart rate in error [4]. In order to overcome this problem the Moyo unit incorporates an ECG amplifier which is used to measure the mother's heart rate by way of an ECG signal picked up from her hands. This heart rate value is then indicated

separately on the LCD. The Moyo unit is held by the mother giving birth who presses her fingers onto the stainless steel electrodes of the unit, which then detects her ECG and measures and displays her heart rate. The values of the two heart rates can then be compared by the midwife and used to discern whether or not the ultrasonic transducer is measuring the infant's heart rate reliably or the mother's heart rate in error. As the mother may be in distress during childbirth, the problem arises of ensuring that she maintains an adequate grip on the monitor to allow her ECG to be detected reliably.

It is intended to implement a mechanism in the Moyo unit which will verify that the electrical contact of the mother's fingers with the stainless steel un-gelled electrodes is acceptable. This is done by measuring the contact impedance of the skin-electrode interface at each electrode to assess whether or not the mother has pressed her fingers properly onto the electrodes. Surface contact



Figure 1. The MOYO Unit in Use

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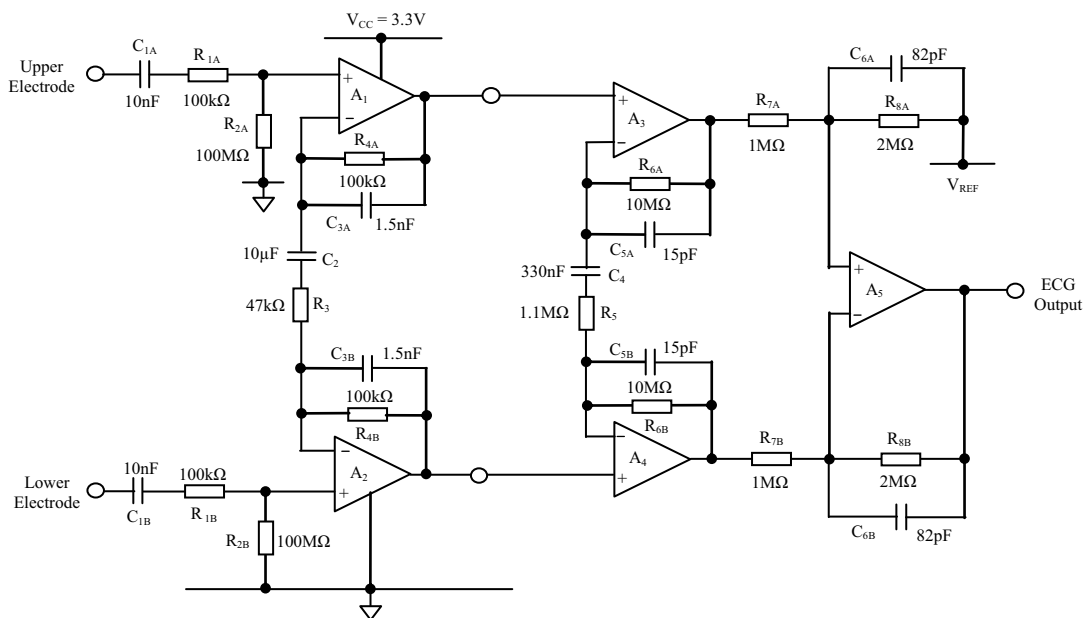


Figure 2. Schematic Diagram of the Original ECG Amplifier

impedance of each electrode is measured independently and a visual alarm is activated when the state of contact of either electrode is unsatisfactory for the purposes of reliable ECG measurement.

2 Background

2.1 Existing ECG Amplifier

The schematic diagram of the ECG amplifier originally used in the Moyo unit is shown in Figure 2. This is a three-stage instrumentation ECG amplifier adapted from an earlier design by Burke & Gleeson [5] and later improved by Assambo & Burke [6]. The 46dB of differential gain is split primarily between the first stage with 14.4dB and the second stage with 25.6dB while the third stage provides 6dB of the gain with differential-to-single-ended conversion. The input stage provides a high differential-mode and common-mode impedance of 100MΩ in order to preserve adequate common-mode-rejection-ratio (CMRR) when interfacing with high-impedance dry electrodes. The amplifier operates from a single 3.3V supply rail and the input stages are biased to a mid-rail voltage of 1.65V using a separate dc-to-dc convertor chip not shown in the schematic. ESD protection and r.f. suppression elements are also omitted for clarity. The operational amplifier used in this design is the AD8648 (Analogue Devices Inc.) which is a quad amplifier package with each op-amp having a gain-bandwidth product of 24Mz and a quiescent current of 2mA. It also offers low offsets and noise and a CMRR of 84dB. The input impedance of each amplifier is unspecified, however.

2.2 Electrodes

The electrodes used in the Moyo unit shown in Figure 1 are made of stainless steel with a matt-finish surface, and

are easily disinfected before use with an alcohol wipe. Electrical contact is made with the electrodes by the user gripping the unit in both hands with their fingers placed on the electrode surfaces. Only the firmness of the user's grip determines the contact pressure. In order to obtain an indication of the contact impedance of these electrodes a method formerly reported by Baba & Burke [7] was used to characterize the electrodes. An equivalent electrical model of the electrode is shown in Figure 3. This can be seen to contain two parallel C-R networks as well as two series resistors. The frequency response of the model is therefore characterized by two time constants. The values of these time constants are very large, which makes their evaluation in the frequency domain very difficult and consequently this was done in the time domain instead.

The current source shown in Figure 4 was used to inject a minute current of 2μA through the electrodes while being held by the user. The current was activated and then deactivated via the relay for a stable period in the region of 30s. A 10kHz sinewave signal was also used to allow the high frequency purely resistive elements to be evaluated. A program in MATLAB (MathWorks Inc.) was then used to fit a two time constant C-R model shown in Figure 3 to the recorded voltage waveforms and to determine the values of the individual components of the model.

This was done for measurements made on 20 subjects of varying age, sex, nationality and skin colour, in order to establish the extent of the variation of electrode impedance to be expected. Only the passive components of the model were of interest in this instance and the dc polarization potentials were not measured. The ranges of values measured for each component of the model are listed in Table.1 below. Plots of the magnitude and phase as functions of frequency of a single stainless steel electrode are shown in Figure 5. Plots are shown for a light grip on the electrode and for a falling current. The impedance varies considerably within the frequency range of the ECG signal. The magnitude falls off abruptly

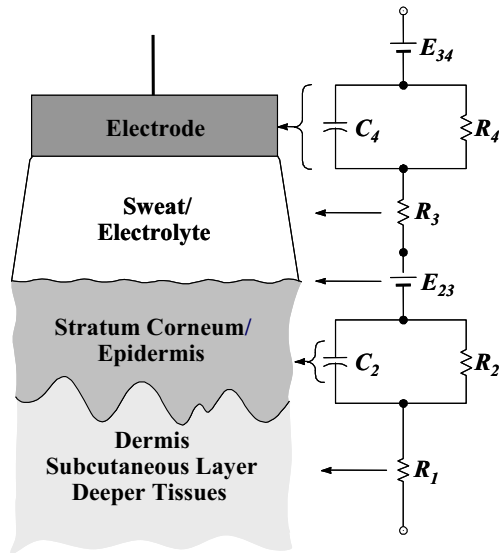


Figure 3. Equivalent Electrical Model of the Electrodes

above 1Hz and the phase falls off above 10Hz. It can be seen that the magnitude of the impedance is less than 30kΩ at frequencies above approximately 10Hz. This is considerably lower than the input common-mode resistance of the amplifier or of the protection resistor, R_1 , shown in the schematic diagram of the amplifier in Figure 2. This indicates that the signal levels involved in measuring the contact impedance of such electrodes are by nature quite low.

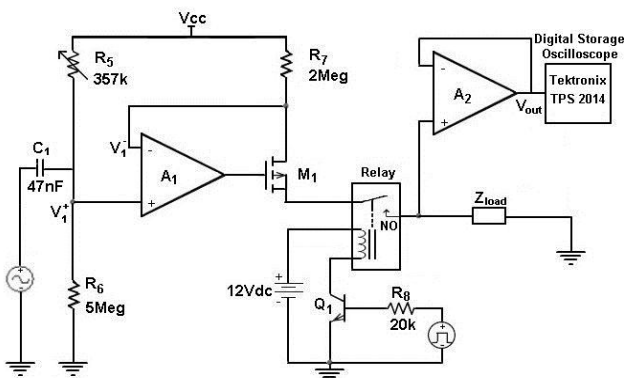


Figure 4. Electrode Impedance Measurement Circuit

Table 1. Range of Values for Electrode Model Elements

Element	Minimum	Maximum	Unit
$R_1 + R_3$	20	45	kΩ
R_2	2.03	331	kΩ
C_2	0.000619	891	μF
R_4	1.96	911	kΩ
C_4	0.00134	976	μF
τ_2	0.0000227	25.4	s
τ_4	0.00039	889	s

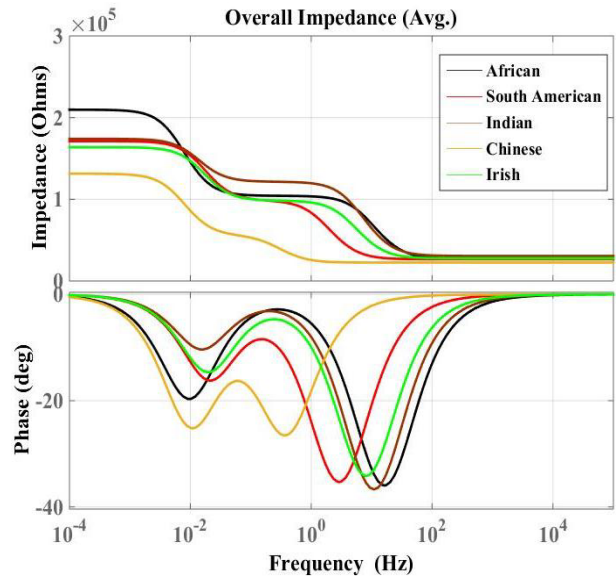


Figure 5. Impedance of Blasted Stainless Steel Electrodes

3 Methodology

A block diagram of the electrode contact monitoring system is shown in Figure 6. An oscillator and bandpass filter are used to provide a sinusoidal source signal for measurement purposes that is outside of the range of the ECG signal spectrum. This signal is then buffered in unity-gain inverting and non-inverting amplifiers to provide antiphase measurement sources. These antiphase signals are then fed through the common-mode input resistors of the ECG amplifier to the electrodes. The first stage of the ECG amplifier is also used to provide initial gain for the contact monitoring signal. Following this, bandpass filters are used to extract the sinusoidal signal from the ECG signal at the outputs of the first stage of the amplifier. The output signal of the upper and lower bandpass filters are then fed into separate half-wave rectifiers in order to provide a steady-state indication of the contact impedance at each electrode. The output voltages of the rectifiers are then compared with reference levels in separate threshold detectors in each channel. When the contact impedance of an electrode exceeds the reference value, the output of the corresponding threshold detector goes high. This output is then used to drive a LED via a suitable buffer circuit. LEDs were used to provide visual alarms in the prototype unit but this will implemented on the LCD display in the final unit to maintain very low power consumption.

4 Measurement Circuit

4.1. ECG amplifier modification

A full schematic diagram of the electrode contact monitoring system and its incorporation into the ECG amplifier is shown in Figure 7. ESD protection diodes, D_3 and D_4 and r.f. suppression capacitors, C_{15} , are also

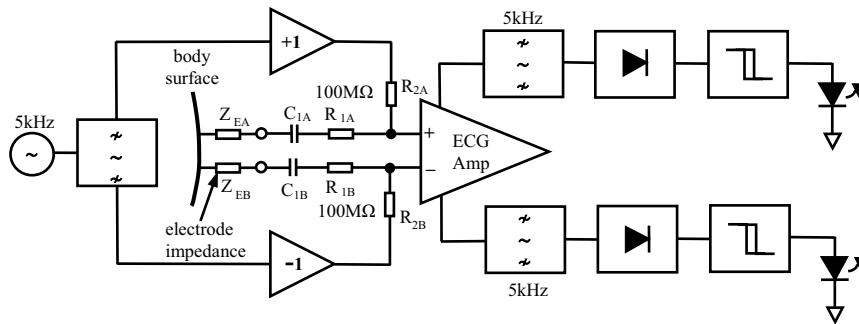


Figure 6. Block Diagram of the Electrode Impedance Measurement System

shown in each channel. It was decided to use the front-end stage of the ECG amplifier to provide the first phase of amplification for the electrode contact monitoring signal. The gains of the three stages of this amplifier were revised accordingly. The first stage gain was changed to 22dB, the second stage to 25dB and the third stage was revised to give unity gain. The revised component values set the 3dB lower cut-off frequency at 0.67Hz as required by the international standards for clinical heart-rate monitoring [8]. It also increased the CMRR by 6dB in the ECG signal bandwidth. The AD8619 (Analog Devices Inc.) op-amp was used in the first stage of the ECG amplifier as it has a gain-bandwidth product of 400kHz with a quiescent current of 38 μ A per channel. Its upper cut-off frequency was self-limited by the op-amp at greater than 10kHz, thus allowing full amplification of the 5kHz contact monitoring signal. The OPA4379 (Texas Instruments Inc.) op-amp was used in the second and third amplifier stages as it provides a gain-bandwidth product of 90kHz with a quiescent current of only 3 μ A per channel. The upper cut-off frequencies of the second and third stages of the amplifier were set at 350Hz and 480Hz respectively. This reduced the power consumption of the redesigned ECG amplifier by a factor of 120.

4.2. Measurement signal generation

An astable multivibrator built around the comparator A_6 is used to generate a square-wave at a frequency of 5kHz. The TLV3201 (Texas Instruments Inc.) is used as the comparator because of its high slew rate and rail-to-rail voltage swing with a low quiescent current of 40 μ A. The bandpass filter centred around op-amps A_7 and A_8 is a multiple feedback configuration having a centre frequency of 5kHz and a Q-factor of 12. This provides a sinusoidal source signal with an amplitude of 1.35V peak. The choice of contact monitoring frequency was a design compromise. A high frequency well above the ECG bandwidth is desirable as it is easy to extract and separate from the wanted ECG signal. However, the parasitic input capacitance of operational amplifiers and r.f. suppression capacitors have a significant shunting effect on the electrode contact impedance at higher frequencies so that the signal developed across this impedance is very small. The design value of 5kHz was chosen to allow a reasonable signal level for measurement, while at the same time maintaining the measurement frequency at least a decade above the ECG band.

4.3. Signal injection

The sinewave is then passed through two buffer amplifiers, a non-inverting unity-gain stage built around op-amp A_{10} and a unity-gain inverting stage built around op-amp A_{11} , which also have mid-rail bias voltages. The in-phase component is fed to the upper electrode point via the upper-channel common-mode input resistor R_{2A} . The inverted component is fed to the lower electrode contact point via the lower-channel common-mode resistor R_{2B} . Both resistors R_{2A} and R_{2B} of 100M Ω are bypassed by parallel capacitors of 1pF in order to lower the series impedance seen at the signal injection frequency of 5kHz to around 24M Ω . This method of feeding the contact monitoring signal to the electrodes preserves the high common-mode impedance at the ECG amplifier input and consequently the CMRR. The use of antiphase signals also allows the signal developed on the mother's body to be kept close to zero.

The levels of 5kHz signal present at the input terminals of the ECG amplifier are those developed across the contact impedances of each electrode namely, Z_{EA} and Z_{EB} . With good contact and low skin-electrode impedance the signal levels are extremely small as the impedance of resistor R_2 in parallel with capacitor C_{14} forms a potential divider with the electrode contact impedance, Z_E , at each terminal of the amplifier. In fact, most of the 5kHz signal present is actually developed across the protection resistor R_1 in each case. This has a value of 100k Ω , which is much greater than the typical contact impedance of the electrode itself.

4.4. Signal extraction

Under normal conditions with good electrode contact the signal at 5kHz present at the amplifier input terminals is of the order of 12mV. On the other hand, when electrode contact is lost altogether, a signal level of just less than 125mV is obtained at the inputs terminals of the amplifier.

If a maximum electrode contact impedance is set at 100k Ω , which is the same value as the resistor, R_1 , then the maximum expected signal with good electrode contact appearing at the input of the ECG amplifier is of the order of 22mV. The gain of 22dB in the first stage of the ECG amplifier provides the initial gain for the contact monitoring signal, raising the signal level at the output of this stage to 268mV.

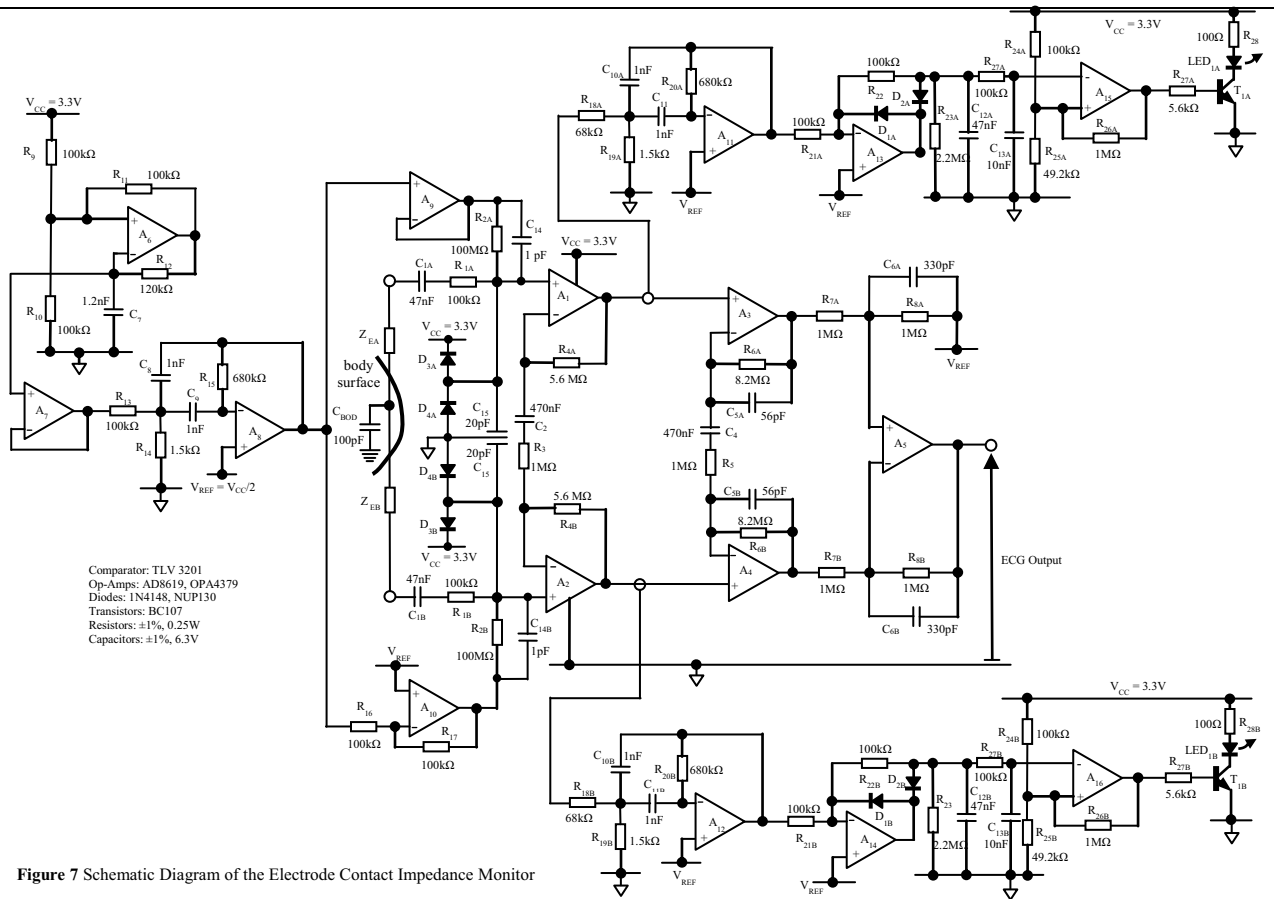


Figure 7 Schematic Diagram of the Electrode Contact Impedance Monitor

The electrode contact monitoring signals are extracted from the outputs of each side of the first stage of the ECG amplifier using band-pass filters built around op-amps A_{11} and A_{12} . These filters are multiple feedback structures having a centre frequency of 5kHz, a gain of 14dB at this frequency and a Q-factor of 12. All of the bandpass filters were implemented using the AD8619 op-amp. This provides an overall gain of 36dB or a factor of 60 to the signal developed across the resistor R_1 and the electrode impedance Z_E in series. The output signal level from the filters with good electrode contact is then of the order of 350mV. At the limit of poor electrode contact impedance of 100k Ω , this rises to approximately 660mV peak.

4.5. Signal level detection

The output sinewaves from the band-pass filters in each channel are then fed into precision half-wave rectifiers built around op-amps A_{13} and A_{14} . The rectification process has unity gain but is inverting and includes envelope detection of the sinewave by means of a C-R smoothing circuit having a time-constant of 1s. This provides a steady-state signal level representing the amplitude of the recovered sinewave at 5kHz. The rectified level is negative-going with respect to the reference voltage of half the supply, $V_{REF} = 1.65V$. Therefore, at the limit of poor electrode contact, the output voltage of op-amp A_{13} or A_{14} falls to 1V.

Finally, the rectified voltages in each channel are fed into threshold detectors built around op-amps A_{15} and

A_{16} . The reference voltage of each threshold detector is set at 1.09V relative to ground by resistors R_{24} and R_{25} with approximately 40mV of hysteresis added by the positive feedback provided by resistor R_{26} . The action of the threshold detectors is also inverting. Consequently, when the rectified 5kHz signal level reaches a threshold of 700mV below V_{REF} or 1.09Vdc, the op-amp acting as a comparator changes state and its output goes from 0V representing good electrode contact, to the supply voltage of 3.3V representing poor electrode contact.

All of the op-amps used in the signal level assessment circuitry are AD8657s in order to minimise power consumption. This HI output voltage of the op-amp is then used to feed a bipolar transistor based light-emitting-diode (LED) driver that provides a current of 10mA to the activated LED. The LED will be replaced in the final design by a segment on the LCD in the Moyo unit.

5 Performance Verification

For test purposes, the electrodes were implemented as matched resistors in each channel, in the first instance. The value of these resistors were varied between 10k Ω and 10M Ω and the signal level at the output of the filters, V_{A11} , V_{A12} , the half-wave rectifiers, V_{R23A} , V_{R23B} and the threshold detectors, V_{A15} , V_{A16} , recorded as shown in Table 2. The threshold detector logic outputs, change state at a value of electrode resistance between 100k Ω and 200k Ω as required. Imbalance in the electrode

resistance in each channel was also investigated and the output logic states of the threshold detectors are shown in Table 3. An electrode impedance above the threshold level in either channel caused the threshold detectors in both channels to register this. This is due to cross-coupling in the differential input stage of the amplifier.

Table 3. Threshold Detector State vs. Electrode Resistance

Upper → Lower ↓	10kΩ	100kΩ	200kΩ	10MΩ
10kΩ	LO/LO	LO/LO	HI/HI	HI/HI
100kΩ	LO/LO	LO/LO	HI/HI	HI/HI
200kΩ	HI/HI	HI/HI	HI/HI	HI/HI
10MΩ	HI/HI	HI/HI	HI/HI	HI/HI

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Table 2. Voltage at Key Circuit Nodes vs. Electrode Resistance

R _E (Ω)	V _{iA} (mV)	V _{iB} (mV)	V _{A1} (mV)	V _{A2} (mV)	V _{A11} (V)	V _{A12} (V)	V _{R27A} (V)	V _{R27B} (V)	V _{A15} (logic)	V _{A16} (logic)
10k	26	22	166	112	0.63	0.58	1.40	1.38	LO	LO
100k	40	25	285	226	0.88	0.81	1.24	1.27	LO	LO
200k	50	38	380	290	1.21	1.04	1.07	1.16	HI	HI
10M	64	54	608	588	2.68	2.12	0.51	0.73	HI	HI