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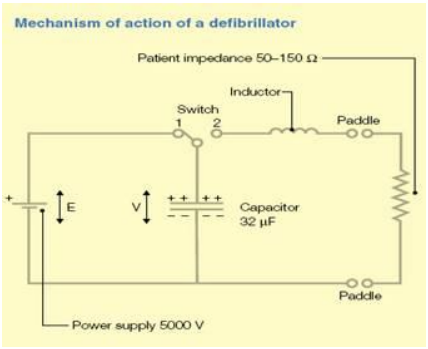
Defibrillator Protection Circuit

The electronics that are used to acquire the ECG during normal use however are not intended to handle the high voltage and current that will be produced by the defibrillator. If this circuitry fails there is the potential for many dangers:

- Permanent damage to medical equipment
- Loss of defibrillator energy into ECG telemetry
- Unwanted conduction paths of defibrillator energy to patient or operator
- Inability to continue monitoring of patient after delivery of shock

In order to create an effective protection circuit, first the defibrillator must be adequately defined. A cardiac defibrillator is classically modelled as a RLC switching circuit shown below:



This setup assumes a mono-phasic pulse of energy is discharged through the patient that has a peak voltage of 5kV at a maximum of 50A and an energy of 400J. In practical circuits the inductor has some resistance so the patient is left to dissipate roughly 360J. Modern day defibrillators deliver a biphasic pulse of lower energy, but these are the standards which are still used for testing. Other known requirements are summarized below:

- Interference with standard ECG circuit operation
- <5% of defibrillator pulse can be shunted away through protection circuitry
- Electrostatic discharge should also be shunted away from ECG telemetry (>5V)
- Must withstand 5kV for at least 3ms

Using these design parameters it was decided the defibrillation protection circuitry should be created as two components

1. High voltage shunting elements to disperse defibrillator pulses
2. Lower voltage shunting elements to disperse ESD interference

For the first requirement two circuit elements that could work are: **gas discharge tubes / NeAr lamps** or **transient voltage suppressor diodes.** The electrical characteristics of each are similar but the foot print of the TVS device is much smaller. This implementation is intended to be created using surface mountable devices and as such a TVS will be used with a reversal voltage ~75V. Preceding this element there will be a 30 $k\Omega$ that is capable of withstanding larger energies. Picking this ensure that in the worst case situation only ~3% of the defibrillators current will be shunted away from the patient. (Assume patient impedance of 150 Ω)

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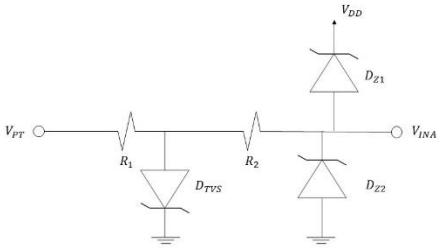
The second circuit will be created using two back-to-back zener diodes with cut off values of ~7V. The instrumentation amplifier can withstand up to 36V at the input which is well above the limit imposed by these elements. An additional current 33 $k\Omega$ current limiting resistor before these diodes ensure that the zener diodes will not be exposed to dangers currents.

Defibrillator Voltage Potection Circuit



Right Leg Drive Circuit

WIP

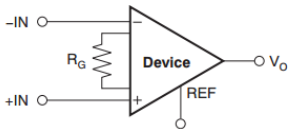
Analog Front End

The Analog front end of this system needs to obtain a differential signal of roughly 1mV, amplify it and filter out as much noise as possible before sending the signal to an ADC. There are two main blocks involved in this process in this implementation:

- A differential instrumentation amplifier (**INA826**)
- A second order low pass filter (Implemented using **OPA4171**)

Head Stage

The head stage was implemented using the INA826 instrumentation amplifier. The primary reason for choosing this device was the fact that the common mode signal was easily accessible and will be needed in the calculation ECG signals later in the design.



Assuming a signal with a maximum value of 1mV, a gain of 100 the needed R_G resistor can be calculated using the supplied equation:

$$G = 1 + \left(\frac{49.4k\Omega}{R_G}\right) \rightarrow R_G = \frac{49.4k\Omega}{100 - 1} = 498.99\Omega \sim \mathbf{500\Omega}$$

At this gain the signal will be linear up to 60 kHz which gives more than adequate room for the signal to pass undistorted.

Low Pass Filter

When designing the anti-aliasing filter the following parameters where chosen:

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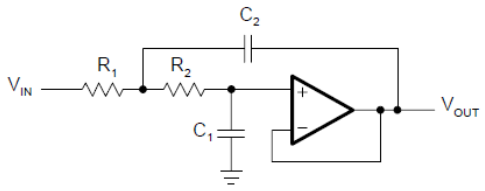
$f_c = 1500Hz$ (Nyquist criteria of ADC requires $f_c \max < 2kHz$)
Butterworth Coefficients ($a_1 = \sqrt{2}$, $b_1 = 1$).
$C_1 = 4.7nF$, $C_2 = 9.4nF$

Using the Sallen and Key topology shown below, other values could be picked as follows using the design process outlined in *Active Filter Design* by Texas Instruments.



The first step in this process is the validation of capacitor vales:

$$C_2 \geq C_1 \frac{4b_1}{a_1^2}$$

$$9.4\,x10^{-9} \geq (4.7\,x10^{-9})\frac{4(1)}{(1.4142)^2} \rightarrow 9.4\,x10^{-9} \geq (4.7\,x10^{-9})\,2 \rightarrow \boldsymbol{9.4\,x10^{-9} \geq 9.4\,x10^{-9}}\surd$$

With the capacitors validated, the resistors for this topology can be found by solving the following equation:

$$R_1, R_2 = \frac{a_1 C_2 \mp \sqrt{a_1^2 C_2^2 - 4b_1 C_1 C_2}}{4\pi f_c C_1 C_2}$$

With all parameters known, the resistor values were found as follows:

$$R_1 = \frac{(1.4142)(9.4x10^{-9}) - \sqrt{(1.4142)^2(9.4x10^{-9})^2 - 4(1)(4.7\,x10^{-9})(9.4\,x10^{-9})}}{4\pi(1500)(4.7\,x10^{-9})(9.4\,x10^{-9})} = 15.963\,k\Omega$$

$$R_2 = \frac{(1.4142)(9.4x10^{-9}) + \sqrt{(1.4142)^2(9.4x10^{-9})^2 - 4(1)(4.7\,x10^{-9})(9.4\,x10^{-9})}}{4\pi(1500)(4.7\,x10^{-9})(9.4\,x10^{-9})} = 15.963\,k\Omega$$

The cut-off frequency needs to be set for anti-aliasing purposes and as such does not need to be exact. The ADCs used in this application have a lower sampling limit of 4 kHz implying a maximal cut off frequency of 2 kHz to satisfy Nyquist criteria.

Choosing $\boldsymbol{R_1 = R_2 = 15k\Omega}$ results in a cut-off frequency of roughly 1.6 kHz which is still acceptable.

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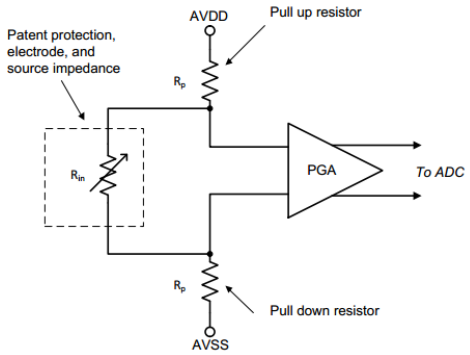
Lead Off Detection

One large issue with monitoring bio potential signals is the electrode-patient interference. High quality ECG signals depend on having low impedance connections between the patients and monitoring hardware. Poor connectivity can be the result of many things including patient movement, drying of contact pads among other things that are expected during patient observation. Due to the fact that this is an expected failure, it becomes necessary to create a telemetry system to monitor the status of electrodes so a technician can rectify the situation. Roughly speaking there are two main strategies used for Lead-Off Detection:

DC Lead off Detection

In order to monitor electrodes using DC excitation pull-up and pull-down reactions are added to the input path of each electrode (shown in the diagram below as R_p). Under proper operating conditions, these two leads are tied together by a low patient impedance (R_{in}). As long as this patient impedance remains low, the difference between these two leads is small. As the electrode contact degrades, this difference becomes larger as each lead is pulled towards their respective power rail. This growing difference can be connected with a comparator, or digitally with a microprocessor as is shown below.



Advantages of this scheme include:

- Continuous monitoring of lead status (DC signal is always on)
- Discrete monitoring does not affect other telemetry in ECG system

Disadvantages of this scheme include:

- Addition of DC bias to ECG signal
- Introduction of additional thermal resistor noise from pull up/down resistors
- Difficulties encountered when using dry electrodes (high base line impedance, little variance with time)

AC Lead off Detection

In order to monitor lead status using AC excitation requires a few more components in comparison to DC monitoring. AC monitoring relies on the injection of a high frequency current (outside the expected range of an ECG signal). As the electrode-patient contact breaks down, the impedance will increase resulting in a larger signal. This AC excitation signal can be applied while recording ECG, or in accordance to some protocol at a given interval. The approach used depends on the processing capabilities of the system.

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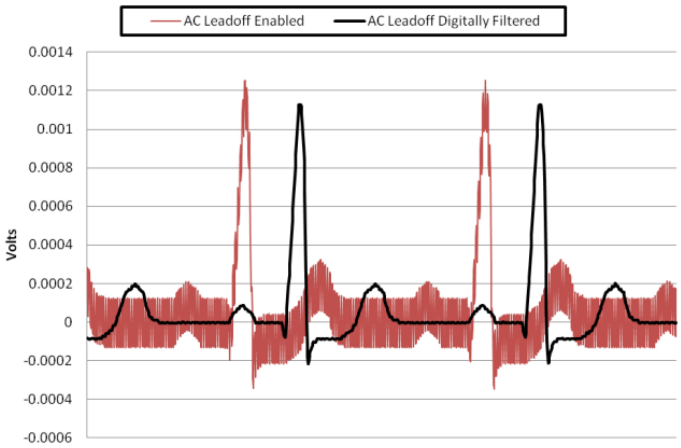


Advantages of this scheme include:

- Scheme works for both dry and wet electrodes
- No additional DC bias imposed on ECG signal

Disadvantages of this scheme include:

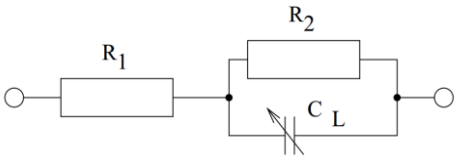
- Requires creation and injection of AC signal
- Requires further signal analysis post-digitization
- Increased ADC sampling frequency to satisfy Nyquist criteria

Plethysmography

The instrumentation created by Project Glia is intended to give physicians as much clinically relevant information about the patient as possible. In first world countries standard vitals are comprised of temperature readings, heart rate, and temperature and oxygen saturation. Oxygen saturation and **temperature** can be acquired from the pulse oximetry telemetry, leaving the need to measure both heart and respiration rate. Details of the **E**lectro**C**ardio**G**ram system that will be used to record heart rate are discussed elsewhere, different schemes to obtain a respiratory trace are explored here. When measuring respiration there are two main approaches that can be used:

1. **Direct Measurement** – Using accelerometers similar to a spirometer volume leaving the lungs can be directly measured
2. **Indirect Measurement** – As the lungs fill and empty with air their electrical properties change. The lung tissue itself is highly conductive due to the high proportion of blood flowing through it, whereas the alveoli are filled with air and can be represented by a perfect insulator. As a patient breathes in the total impedance of their thorax changes which can in turn yield a signal that correlates with respiration.

To minimize additional hardware an indirect method of monitoring respiration will be used that takes advantage of ECG leads that will already be in place. Examining the electrical properties of a person, the most simplistic model is shown below:



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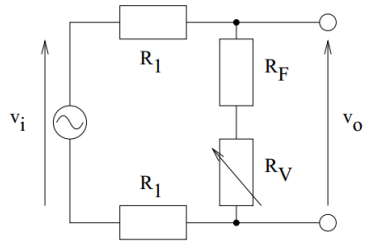
R_1 represents constant resistance of the thorax that is unrelated to the lungs, R_2 represents constant resistance of the lung tissue and C_L is the capacitance portion of lung impedance which changes as air moves in and out of the alveoli. By running a known current through this block a voltage can be measured which will give the impedance and serve as a pseudo measurement of the location of the chest. Due to the fact that the impedance in a capacitor is dependent upon the frequency of excitation an AC signal needs to be used. Using a voltage of a known frequency allows the parallel combination of R_2 and C_L to be combined into R_V as shown in the following diagram:



R_1 serves as a current limiting resistor for safety and also allows for an estimation of $R_F + R_v$ through voltage division. This voltage is the quantity of interest and can be expressed as follows

$$V_o = V_i \left(\frac{R_F + R_v}{2R_1 + R_F + R_v}\right)$$

Typical values for these circuit elements are as follows:

$$R_F \approx 10k\Omega \quad R_v = 5\Omega$$

Comparing these two values it becomes apparent that the signal caused by the variable impendence related to respiratory rate is much smaller than the near DC signal created by constant tissue resistance ($\frac{10\,000}{5} = 2000\times$). Along with this additional DC signal, there is also common mode noise as well as a high frequency tissue capacitor "charging" artifact that will need to be removed in order to acquire the respiration signal. Roughly speaking there are two main strategies that can be used to take advantage of this changing tissue impedance to record a respiration waveform.

Injection of AC Current

The excitation signal V_i can be supplied by an external source that oscillates at twice the maximal expected respiration rate (or greater to satisfy Nyquist criteria). The respiration rate of a healthy human is greatest at birth at roughly 30-60 breather per minute and decreases until adult hood to a rate of 12-20 breaths per minute. Using this background information, impedance pneumography can be incorporated into the Project Glia's ECG as follows:

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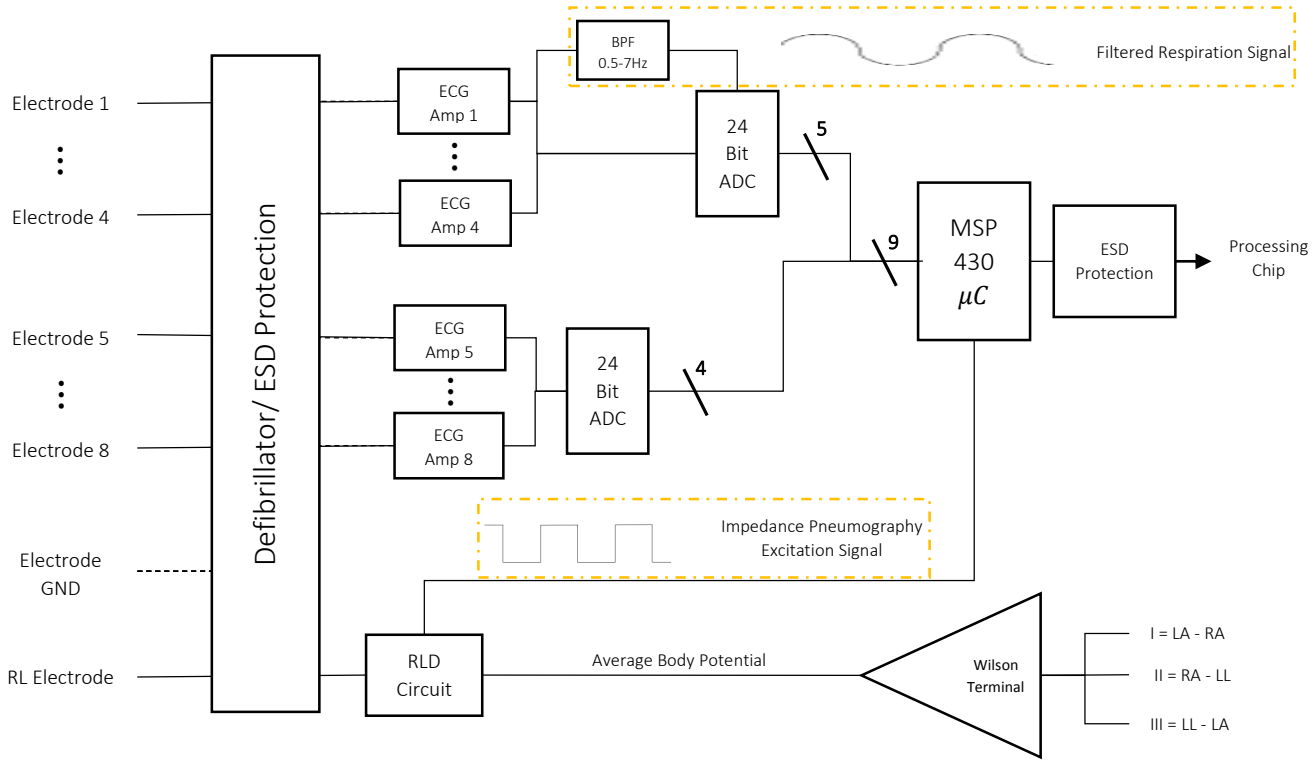

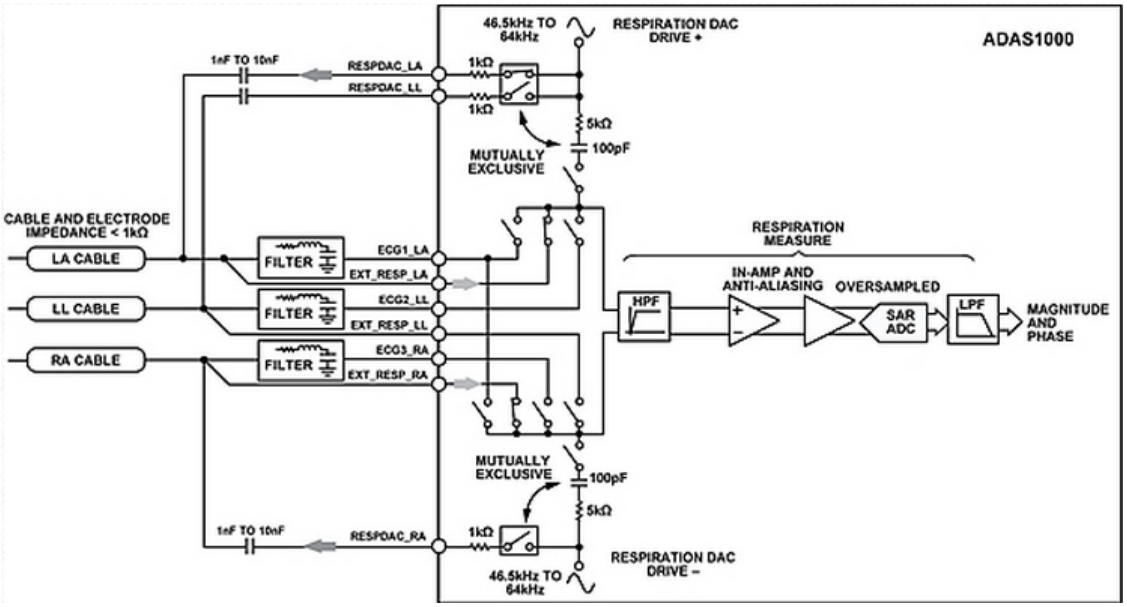

Post ECG Signal Analysis

An alternative method to aquiring a respiration signal requires no additional hardware and instead relies on post signal acquisition processing. The ECG signal can also be analyzed for changes in amplitudes as it will also be modulated by the changing impedance of the chest. With adequate filtering the ECG signal can be removed leaving only breathing waveforms.

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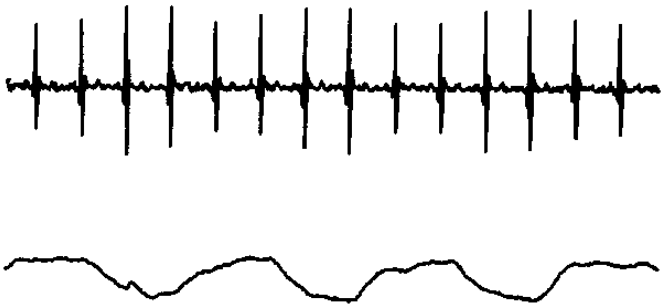

Some considerations to make when using this scheme are listed below:

- The respiration rate is sampled at a much lower rate compared to an external excitation source (1-2Hz vs kHz). That being said a patient's heart rate is typically at least twice that of their respiration rate (Satisfying Nyquist)
- Sampling the impedance using the ECG signal produces a discrete set of information, numerous cycles are necessary before a coherent waveform can be inferred.