Measurement of Impedance with a Heart Catheter

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Introduction

In this lab we aimed to design a heart volume monitor system, which should monitor the blood volume in a heart as it beats. The system will perform this by measuring the impedance of the blood and muscle around the catheter. The theory by Karthek is that blood is almost purely resistive, while the heart muscle contains both resistance and capacitance to the system. By measuring the impedance, we can determine the blood volume of the heart.

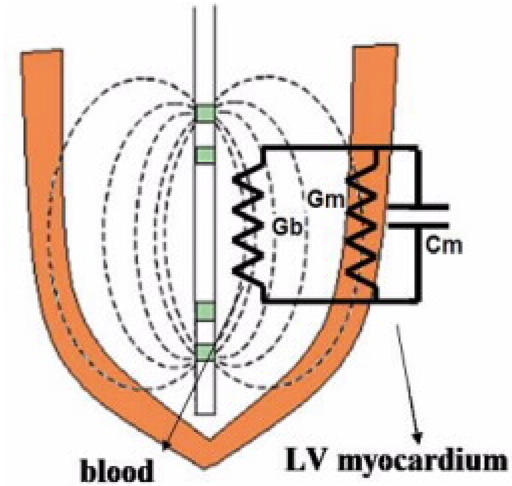

Figure 1: Heart electric model (from Karthik)

Design Goal

The goal of this design is to first generate a clean sinusoid current signal onto the two outer probes on the catheter. The two inner probes will have a sinusoid voltage that is the result of a sinusoid current on an unknown impedance. The phase can be measured by using a comparator to convert both the input sinusoid and output sinusoid into a square wave and measure the lag in the edges. The amplitude of the output sine-wave can be measured by using a full-wave rectifier.

Restriction

Following the safety requirement for a heart catheter, our input current signal into the catheter is restricted to < 15 uA. In addition, we are not allowed to inject any charges into the catheter, which means capacitor decoupling from the monitor system to the catheter.

Sinusoidal Signal Generation Hardware Designs

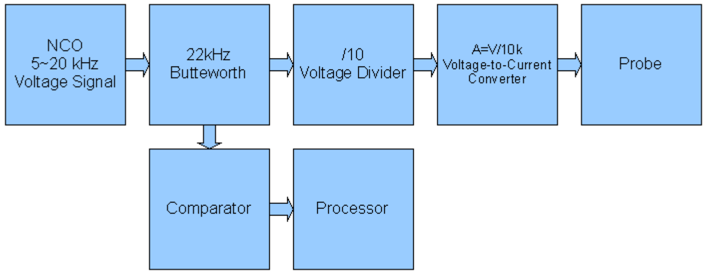

Illustration 1: Input Sinusoid Generation Block Diagram

Numerically Controlled Oscillator

In order to measure the impedance without requiring significant amount of processing, we needed the ability to generate a clean sine-wave. In addition, we want the ability to generate sine-waves with different frequency to allow the potential of measuring impedance at different frequency. Early calculations regarding the speed of the processor and the digital-to-analog chip available to us ruled out a software based solution. We settled on an AD9833BRMZ, a numerically controlled oscillator. The oscillator outputs a 0~650 mV signals and can generate the 5~20 kHz sine-wave signal we need. The chip can be programmed to output a sine-wave with just a sequence of serial-peripheral interface command. The chip requires a clock, which is supplied using DP512's pulse width modulation function. However, the sine-wave output came from an internal sine-wave table through an internal DAC, therefore we expect some quantization noise on the output.

Low-pass Filter

To compensate for the quantization noise, we included a 2-pole Butterworth low-pass filter with a 22.5 kHz cutoff. This allows us to filter out the quantization noise, which is dependent on the pulse-width modulator input into the numerically-controlled oscillator.

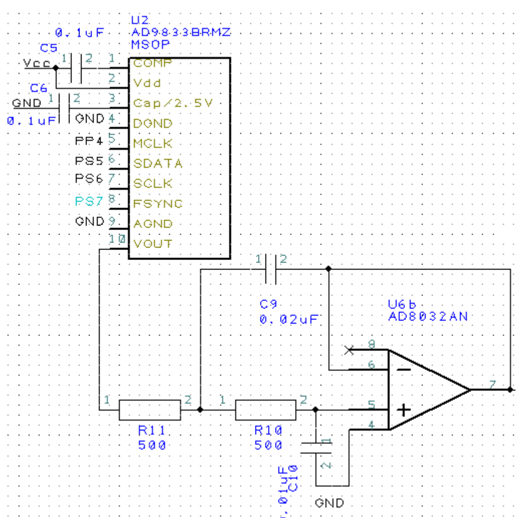

Illustration 2: Numerically Controlled Oscillator + Low Pass Filter

Phase Detector

Impedance measurement requires that we know the phase difference between the input and output sine-wave (in addition to the magnitude). For this we designed a voltage comparator that detects the peak points of the sine-wave. The input passed through a high pass filter in order to allow the signal to swing around 2.5V. The output of the comparator generates a rising edge at the lowest point in the sine-wave while generating a falling edge at the peak of the sine-wave. This signal is fed into the PT5 of DP512 for input capture.

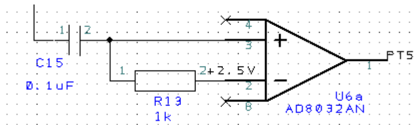

Illustration 3: Phase Detector

Voltage Divider

To limit the current through the catheter, we reduced the voltage by a factor of 10. The signal is followed by a voltage follower to help isolate the effect of the voltage-to-current converter on the filters.

Voltage-to-Current Converter

We used the standard voltage-to-current converter using a 10k current sensing resistor. Combined with the previous voltage divider, the circuit will output a 6 uA sinusoidal current output. The operational amplifier will try to output a voltage that will generate a current to maintain the voltage across sensing resistor. Therefore, by controlling the voltage on the sensing resistor, we can control the current going through catheter. The capacitors on the two outputs are sufficiently large to allow at least a 5 kHz signal through while preventing charge injection. In addition, a 1 MOhm resistor across the output prevents the opamp from operating in open feedback state.

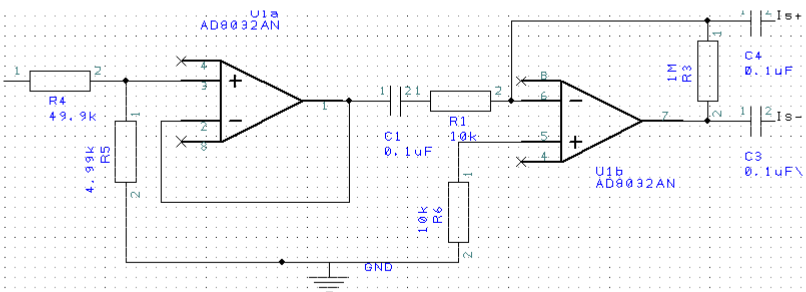

Illustration 4: Voltage-to-Current Converter

Impedance Sensing Hardware Design

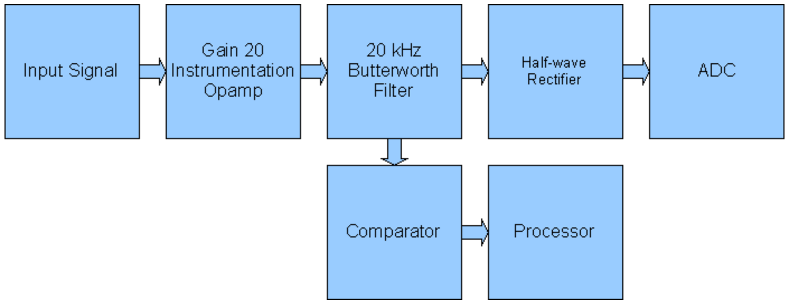

Illustration 5: Output filter block diagram

Instrumentation Amplifier

The output of the two probes on the catheter is in the form of a differential voltage with low amplitude. Since the input current is only 5uA while the general resistivity of the blood is less then 1 kOhms (according to John Porterfield), the differential output on the probe is around 1 mV. We used a gain of 1000, the limit of the instrumentation amp, in order to get a 1V signal.

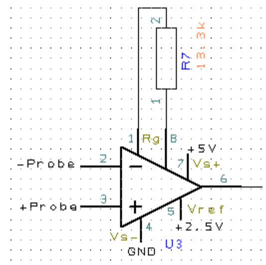

Illustration 6: Instrumentation Amplifier

Low Pass Filter

The same 2-poles Butterworth filter with 22.5 kHz in order to remove any high frequency noise introduced on the catheter. The filter primarily removes the high frequency noise (around 40 kHz) caused by the florescent light in the lab.

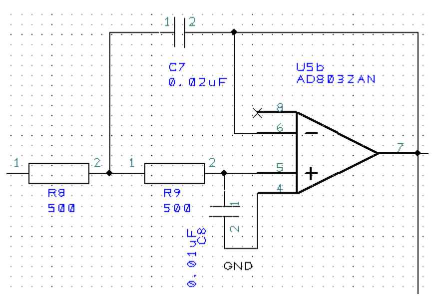

Illustration 7: Low Pass Filter

Phase Detector

The same comparator as the one in the sinusoidal signal generator is used on the filtered signal in order to acquire the phase of the signal. This signal is fed into PT4 on the DP512 for input capture.

Half-wave Rectifier

A half-wave rectifier is used to acquire the magnitude of the signal due to simplicity. And with the input capture system capable of triggering on the peaks of the sine-wave, the half-wave rectifier's purpose will be to hold the signal long enough for the input capture interrupt to execute and sample it.

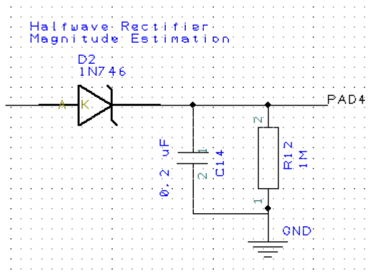

Illustration 8: Half-wave Rectifier

Rectifier Note

During experimentation, we discovered that a potential rectifier design allow us to convert a sine-wave with arbitrary DC into a half wave with nearly double the amplitude. The design is essentially a single stage Cockcroft–Walton generator. However, due to lack of time, we did not determine the configuration that will allow for safe voltage output (preliminary experiment indicates a small signal can result in sufficient high voltage that will damage the chip).

Software Design

Controlling Driving Sinusoidal Frequency

The AD9833BRMZ is controlled through the SPI0 interface and a pulse width modulator. During initialization, the pulse width modulator is setup to output a 1.2 MHz square wave signal with 50% duty cycle, and SPI0 sends out commands to set the two frequency control registers (FREQ0 and FREQ1) and two phase registers (PHASE0 and PHASE1). The write to the frequency register changes the frequency divider, which results in an output frequency equals to 1.2 MHz divided by FREQ0.

Phase Sampling

The phase differences between the signals is sampled using two input captures running with a 42ns period TCNT clock, the fastest clock available for input capture. The phase is computed by dividing the difference between the IC4 and IC5, the delay of the two signals, by the period of IC5 and multiplied by 1024. This can be converted to actual phase with the following formula.

$$Phase_{Actual} = 2\pi * \frac{Phase_{IC}}{1024}$$

Because phase measurement can only be done at the edges generated, the sampling rate is the frequency of the given tot he numerically-controlled oscillator.

Magnitude Sampling

The magnitude of the signal is sampled inside the IC4 interrupt. Since IC4 captures the falling edge, and a falling edge indicated that the sine-wave has reached the peak and is now falling, sampling here should get the nearest value to the maximum magnitude. The magnitude measured is passed into the foreground through a FIFO.

Digital Filtering

Two filters are used on both signals. Due to the rate at which data arrives, a length 64 accumulated averaging filter is used. This reduces the effective sampling rate from 20 kHz to 312.5 Hz. The resulting data is filtered once more by a length 64 running averaging filter. The filter length is chosen after experimentation with the system after it's built, and noting the highly noisy input signal.

Conversion to Real and Imaginary

Once given the magnitude and phase angle, the real and imaginary part of the impedance can be computed as followed.

$R = mag * \cos(phase)$

$I = mag * \sin(phase)$

The cos and sin function are implemented as two 1024 entry lookup table with 3 decimal fixed point results.

User Interface

We designed two switches into the system for user interface. One switch will switches between 3 preset frequencies, 20 kHz, 10 kHz, and 5 kHz. The second switch cycles through 3 different displays: Printing out impedance being measured, plotting magnitude over time, and plotting phase over time.

Calibration

We compute the phase offset of the two signals by measuring the time delay between the two comparators as shown in the Hardware Design section. For calibration we connected RC circuits with known phase offsets directly into the input ports on our measurement system, shown below.

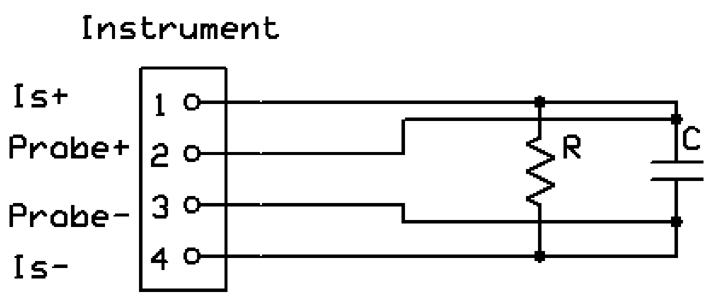

Illustration 9: Phase Calibration Circuit

Where $phase = \arctan(-2\pi fRC)$. We calibrated the system at f = 20kHz using 4 different values of RC.

Data point	Width between the pulses (counts of 42ns)	Phase offset of RC circuit (Degree)
1	265	80.9
2	287	85.4
3	290	87.7
4	301	89.1

Table 1, Phase Calibration

The calibration graph is

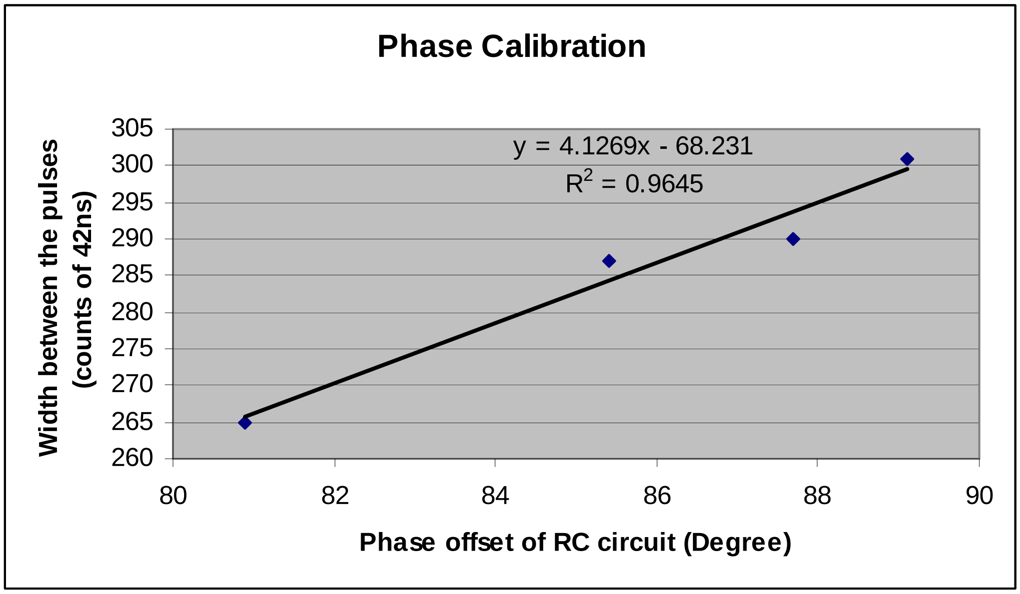

Figure 2, Phase Calibration

Magnitude is calibrated using 4 high tolerance resisters of the same value connected as below:

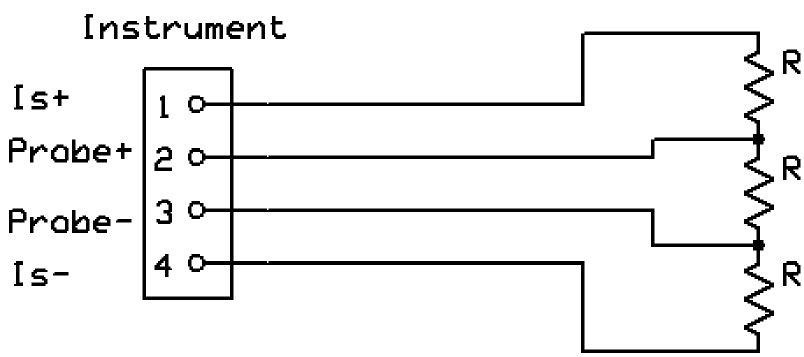

Illustration 10: Mag. Calibration Circuit

Three tests give the following calibration table and graph.

Test	ADC Read Value (5V/1024)	R (Ohm)
1	121	500
2	240	1000
3	482	2000

Table 2, Mag. Calibration

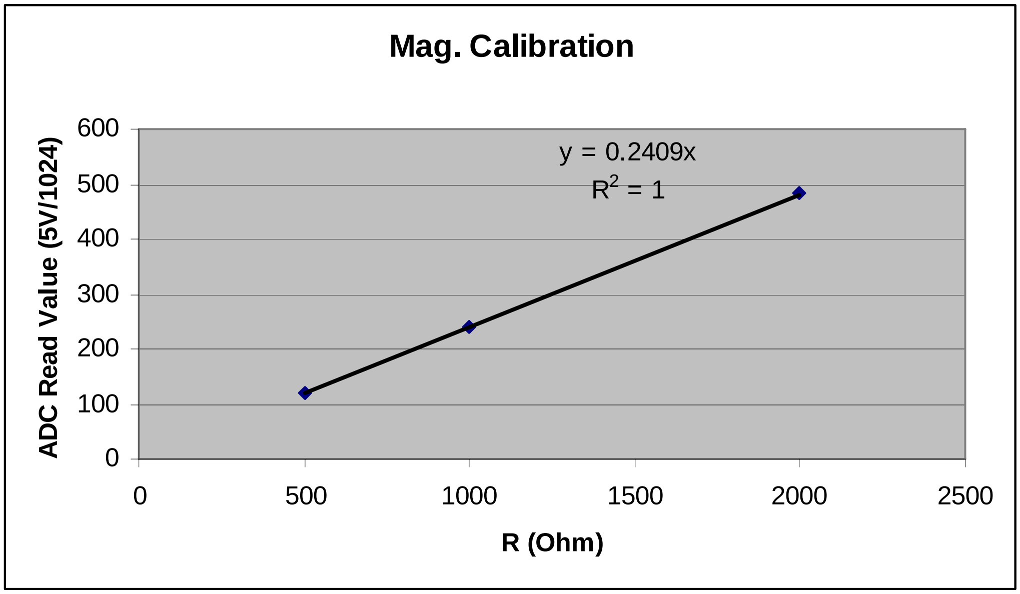

Figure 3, Mag. Calibration

We then connected the catheter to the system, and submerge the catheter in saline solution of known conductivity. Below is the saline calibration table.

Cond (uS/cm)	Mag (Ohm)	Phase(degree)	ImY(uS)	RealY(uS)
6400	1153	53.5	697	540
3200	1173	83.2	847	101
1600	1213	147.2	449	69

Table 3, Saline Calibration

Where Cond is the conductance of the saline as measured using a commercial conductance meter; Mag is the measured resistance in Ohms, and Phase is the measured phase angle in degrees. ImY is the imaginary part of the admittance, and RealY is the real part of the admittance calculated using the Mag and Phase measurements.

This data can be used to eliminate the effect of the catheter from the results by estimating the field form factor F, and probe constant K from RealY, as shown below:

Cond (uS/cm)	ImY(uS)	RealY(uS)	F(cm)	K(1/cm)
6400	697	540	0.0843	11.83
3200	847	101	0.0316	31.68
1600	449	69	0.0431	23.18

Table 4, Field factor and Probe constant

However, since our instrument does not display F or K, this data is not included as part of our calibration.

Sensitivity

The static sensitivity of the instrument can be represented by the slope m, of the straight line through the calibration graph that gives the least squared error. The value m, as shown in the calibration graphs in figure ### and ### give a static sensitivity of

$$0.2409\frac{(5V/1024)}{\Omega}\times\frac{5000(mV)}{1024(5V/1024)}=1.176\frac{mV}{\Omega} \quad \text{for magnitude measurement}$$

$$4.1269\frac{(42ns)}{Degree}\times\frac{1}{42}=0.098(ns/Degree) \quad \text{for phase measurement}$$

Time constant

Since our phase measurements are very noisy, we used a number of averaging and median filters to get a stable reading. Thus the time constant of our system can be calculated based on the number of elements in these filters. The sampling rate of the system is the frequency of the generated sinusoid.

For the magnitude measurements the only filter applied is a length 64 averaging filter, therefore the time constant τ is (1/f) *64.

For the phase measurement, the filters used are a 64 length averaging filter and a 15 length median filter, the time constant τ is (1/f)*64*15.

For the admittance measurements, the filters used are a 64 length averaging filter, a 15 length median filter, and a 32 length averaging filter, the time constant τ is (1/f)*64*15*32.

The time constant for each measurement are listed below.

Frequency (kHz)	τ -Magnitude (ms)	τ - Phase (ms)	τ -Admittance (ms)
20	3.2	48	1536
15	4.2	64	2048
10	6.4	96	3072

Table 5, Time Constant

Resolution

The instrument resolution is the smallest input signal difference that can be detected by the entire system.

We measured the resolution of our system by starting with a cuvette filled with 100 mL of saline with conductivity of 6.4mS/cm and gradually add purified water until we can observe a difference on the instrument readout. We observed a difference in the readout when we added 5mL of purified water. Since conductivity varies linearly with concentration, the conductivity of the diluted saline is 5.82 mS/cm.

In addition, we modified the code to record 1000 data points before and diluting the sample. By performing a student's T-test on the collected phase and magnitude data, we are 99% confident that the resolution of our system is at least 6.4 - 5.82 = 0.58 mS/cm.

Accuracy

The accuracy of the instrument can be measured by measuring a known input with the instrument. We evaluated the accuracy of the system by measuring saline of a known conductance of 6.4 mS/cm. Below is a table of 10 samples taken when the catheter is submerged in this saline solution.

Measurement #	1	2	3	4	5	6	7	8	9	10
Mag (Ohm)	1176	1176	1176	1176	1176	1176	1176	1176	1176	1176
Phase (Deg)	50.4	51.3	56.5	49.1	55.4	53.2	51.0	55.7	53.2	53.9

Table 6, Accuracy

Saline solution with 6.4mS/cm should give us a magnitude reading of 1176 ohms and a phase reading of 53.5 degrees. Assuming these values are the true values, the average accuracy of reading is:

Magnitude (Ohm) = 0.0%
Phase (Degree) = 9.9%

Repeatability

We measured the repeatability of the system by submerging the probe into the saline solution and resting the catheter on the edge of the beaker to avoid motion artifacts in our measurements.

We then took 1000 data over 30 seconds. The repeatability is the full range of measurements calculated as the maximum minus the minimum of the results.

Repeatability for Magnitude = 1 Ohm
Repeatability for Phase = 9.2 Degree

Reproducibility

We evaluated the reproducibility of the system by keeping the same beaker of saline solution and measure the properties of the solution over 3 days. To keep the temperature of the saline constant, we always conducted the measurements in an air-conditioned room with the temperature held relativity constant.

Assuming the conductivity of the saline does not change over time. The results of the measurements and the calculated reproducibility over 3 days are shown below:

	Day 1	Day 2	Day 3	Reproducibility
Mag. (Ohm)	53.2	54.5	51.7	2.8
Phase (Degree)	1176	1177	1176	1

Table 7, Reproducibility

Where reproducibility is calculated by the maximum minus the minimum of the results.

Power consumption
We measured the power consumption of the system by attaching a ammeter to the power supply of the system, with the result of a constant current draw of 100 mA. Since the 9V battery used to power the system has a capacity of 650 mA-hour. Our system would be able to operate for 6 and a half hours continuously before the battery needs to be replaced.

Conclusion and Discussion

We encountered a number of issues during both the design and construction phase of the project.

The majority of the problems can be attributed to our inexperience with the measurement setup and inadequate testing during the design phase. We did not realize how small the magnitude would be after the catheter is submerged into the saline solution. As such we did not add enough gain after the instrumentation amplifier. This isn't as big of a problem for the magnitude measurement; however this makes trigging the comparator very unreliable, thus making phase measurement unstable and prone to noise. A scope trade of the signal after the instrumentation amplifier is shown below, together with the output of the comparator.

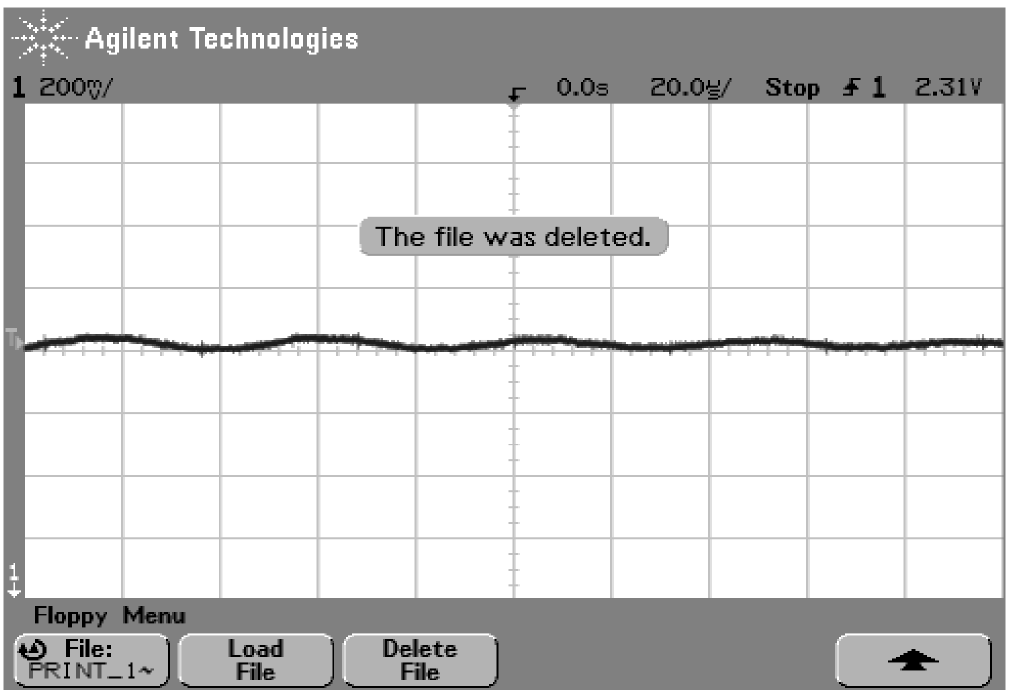

Figure 4, Output from the instrumentation amplifier

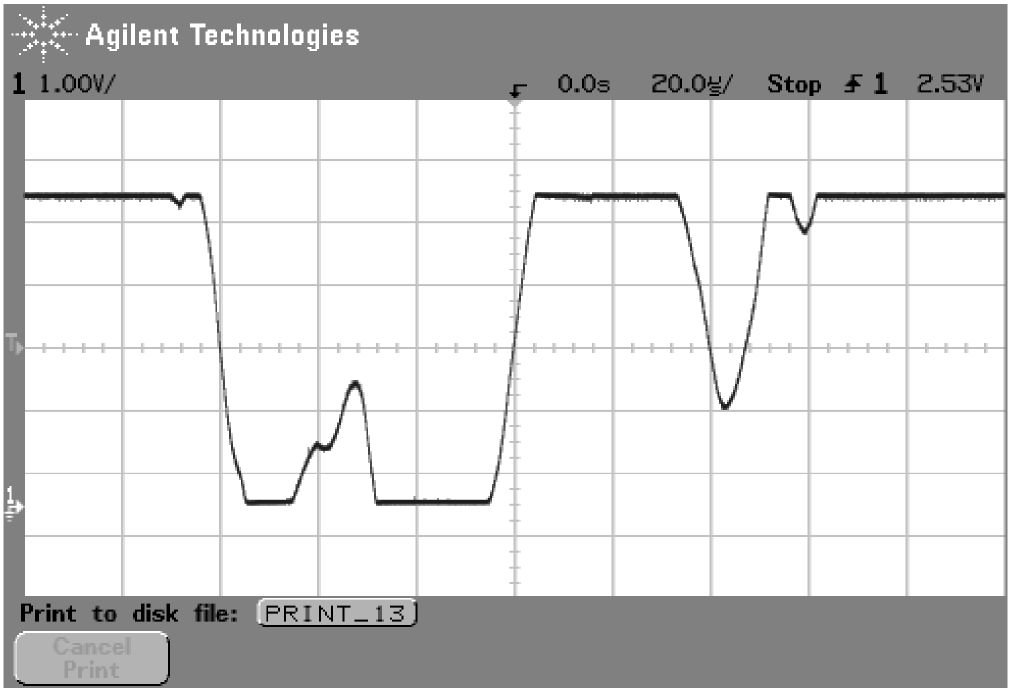

Figure5 , Comparator Output with catheter submerged

Compare with the comparator output when the catheter is in the air, and the signal from the INA is large.

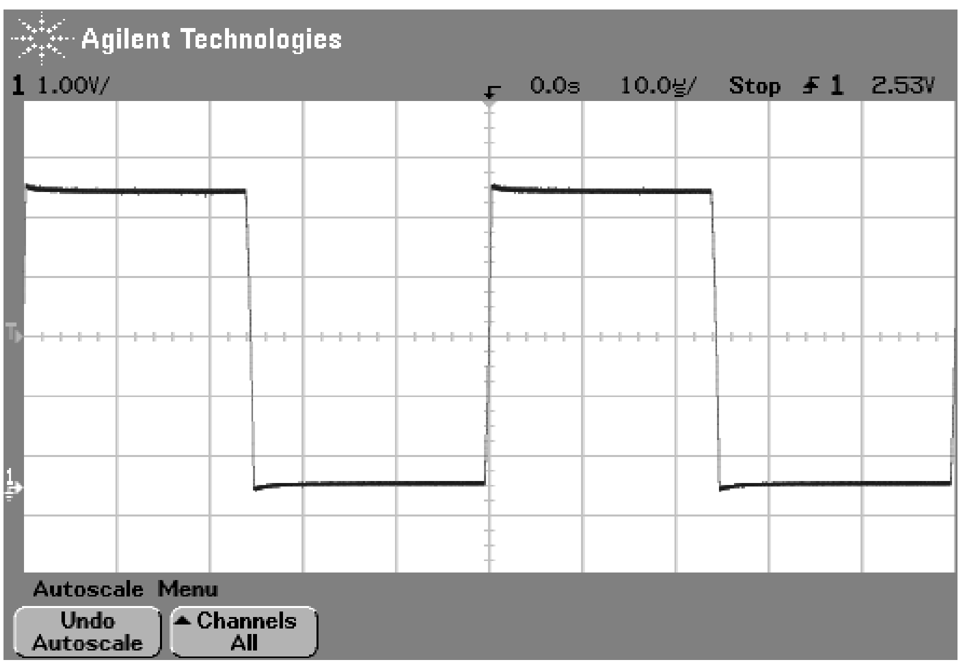

Figure 6, Comparator output with catheter in the air

As a result of noise problem with the comparator, we had to use both a 15 length median filter together with a 64 length running average filter on the phase measurements before calculating and displaying the admittance. Without these filters the measurement would be so noisy as to be unreadable. As a result, the response time of the admittance calculation is very slow, often times lagging behind the phase and magnitude changes by 1 or 2 seconds. This is a definite flaw in a measurement system designed to measure heart stroke volume, since the frequency response of the admittance calculations is much lower than the heart.

If we are to build the system again, we would implement higher pole filters, and multiple gain stages. In addition, we would most liking choose a micro-processor that has a lower current requirements as well as a smaller form factor. For power we will use a rechargeable lithium-ion battery together with a charger circuit to extend the operational lifetime of the system.