Development and Characterization of an Instrument for Measurement of Generalized Biosignals (November 2001)

Chu, W., Lai, R., Lai S., Park, J., and Pottle, B., AUTHORS, *Members ECE 511*

***Abstract* — This paper presents the development of a generalized instrument for the measurement and amplification of biosignals. The physiological basis of neurological and muscular system signals is discussed, highlighting the relevant design constraints imposed by the nature of the signals. Finally, the system was used to make EMG recordings on the bicep under various conditions. The data from these recordings is filtered using the EMAP 2000 software package and also correlated with the actions performed during the experiment.**

***Index Terms* — Biosignals, ECG, EMG, Instrumentation**

I. Introduction

EMG, also known as *electromyographic activity* or *aggregate motor myopotential activity*, is the electrical response of skeletal muscles during contraction. Usually measured in microvolts, EMG is measured by placing electrodes on the skin at the muscle site. Levels are dependent on the number of motor units firing and the rate of discharge near the electrodes. The electrical activity of the firing motor units is directly proportional to the tensile forces in the muscles. Thus, EMGs can provide good data of muscle contraction or relaxation.

The smallest functional unit of muscle fibers is called a motor unit. When muscle fibers are stimulated by a single axon firing in the brain or spinal cord, they contract at the same time. Motor units then repeatedly emit short bursts of electrical activity called motor unit action potentials. The timing between bursts is random for each motor unit. During a normal EMG recording, several motor units are active. The timing of electrical bursts between different motor units shows no correlation. As a result, a “random interference pattern” of electrical activity is obtained [2].

Manuscript received November 2001. This work was supported in part by the authors and their parents under grants to Cornell University in the form of tuition payments.

W. Chu (whc3@cornell.edu), R. Lai (zl29@cornell.edu), and J. Park (jap33@cornell.edu) are with the department of Electrical and Computer Engineering, Phillips Hall, Ithaca, N.Y.

S. Lai (swl4@cornell.edu) is with the department of Mechanical and Aerospace Engineering, Upson Hall, Ithaca, N.Y.

B. Pottle (btp4@cornell.edu) is with the department of Biological and Environmental Engineering, Riley Robb Hall, Ithaca, N.Y.

In an EMG setup, bipolar electrodes are placed on the subject’s skin surface to measure the interference pattern. The standard deviation of the interference pattern, or the *EMG amplitude*, directly relates to muscle activity. Therefore, as the number of and/or the average firing rate of motor units is increased, both EMG amplitude and total muscle tension increases. This may end up being a dynamic relationship. Depending on the resolution desired, this relationship may be treated as nonlinear.

In our experiment, we measured the electrical activity within the bicep muscle for eight different conditions.

1. Still Mode: Subject’s arm is resting on the table
2. Squeezing Mode: Subject is squeezing a empty toilet paper roll at 1 sec intervals
3. Continuous Squeezing Mode: Subject is squeezing the toilet roll continuously
4. Raising and lowering light object (function generator)
5. Holding function generator continuously
6. Raising and lowering heavy object (chair)
7. Holding chair continuously
8. 4 one hand pushups

Complications

As mentioned above, the surface EMG amplitude directly relates to the force produced by individual muscles. However, there are at least two difficulties in this analysis. The EMG recorded at the surface of the skin may contain "cross-talk" [2]. That is, the EMG from muscles other than which the experimenter intends to record may be included in the signal. Second, relating EMG recordings to individual muscles requires verification by making direct mechanical measurements of those muscle forces. There is currently no reliable experimental method to make these measurements.

One way to solve these limitations is to use the relationship between surface EMG amplitude and *joint torque*. By estimating the torque, some of the cross-talk signals are automatically removed even though they cannot be physically removed from the muscle forces contributing to the signal. Second, net torque about a joint can be verified by direct mechanical measurement. Relating EMG amplitude to joint torque is a continuing research interest.

Background on the Action Potential

A. Resting Potential

A resting potential is the electrical charge across the plasma membrane. In resting conditions, the interior of the cell is negative while the exterior is positive. The magnitude of the resting potential varies. In excitable cells, it approximately goes up to -70 millivolts.

B. Depolarization

External stimuli reduce the charge across the membrane. There are usually three cases: 1. Mechanical stimuli activate mechanically-gated sodium channels. 2. Neurotransmitters open ligand-gated sodium channels. 3. An increase in potential can open voltage gated sodium channels. For all cases, sodium moves into the cell through facilitated diffusion to create an *excitatory postsynaptic potential* (EPSP) [1]. Once the potential is reduced to the threshold voltage, the action potential is generated within the cell.

C. Action Potentials

The decreased voltage opens up the voltage-gated sodium channels in that portion of the plasma membrane. During the millisecond that the channels are open, ~7000 Na^+ ions rush into the cell and suddenly depolarize the membrane. The sudden depolarization opens up more of the voltage-gated sodium channels in the adjacent portions of the membrane. As a result, a "wave" of depolarization is produced along the cell. The action potential is an all or none response. The strength of the stimulus is determined by the frequency of the action potentials generated.

D. Refractory Period

Less than 0.001 second after the first stimulus, if a second one is applied to the muscle fiber (or neuron on the nerve cell), another impulse is not triggered since the membrane is depolarized. The muscle fiber is currently in its refractory period. Once the –70 mV potential is reestablished, the muscle fiber will be ready to fire again. Repolarization is established by the facilitated diffusion of potassium ions out of the cell.

E. Hyperpolarization

Hyperpolarization occurs when specific neurotransmitters inhibit the transmission of nerve impulses. Inhibition occurs when chloride channels and/or potassium channels open up in the membrane. In either case, the membrane potential is increased because negatively charged chloride ions are coming in while positively charged potassium ions are flowing out. As a result, the cell's threshold voltage is unchanged. However, a stronger excitatory stimulus is needed to reach the threshold. Hyperpolarization is also known as the *inhibitory postsynaptic potential (IPSP)* [1].

II. CIRCUIT OVERVIEW

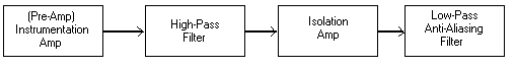
In designing our circuit, we made our design choices based on several criteria. Our most important concern was the safety of the patient and of the equipment. We added a Burr-Brown ISO124P isolation amp to protect the patient and equipment from high current shocks.

The nature of biological signals dictated our design choices. Rather than using a standard operational amplifier, we used an instrumental amplifier because we needed a high Differential Mode Gain (DMG) and a high Common Mode Rejection Ratio (CMRR). Since our signal is only on the order of mV, the 60 Hz noise is significantly larger than our signal. However, the noise is virtually the same on both electrodes, so by rejecting a common signal, we may amplify our desired signal. The input impedance of the entire system is high as well.

We set the overall amplification of the circuit to be approximately 1000 fold. This will amplify the signal sufficiently for detection and analysis. All amplification was done in the instrumentation amplifier. The signal then passed through a 0.2 Hz high pass filter to remove any direct current (DC) drift. The isolation amplifier is operated on the patient side by two 9-V batteries and on the instrument side by a standard power supply. Finally, the signal passes through a low pass filter, which cuts out noise above 1000 Hz.

The output of the system feeds into a National Instruments digital input output (DIO) card, which samples the signal at 5000 Hz. This sampling rate should be more than enough to ensure that no biologically important information is lost.

III. A) CIRCUIT DESIGN : AMPLIFIERS



In designing our circuit, we built the basic system as shown above. An input signal would pass two amplifiers, an instrumentation and isolation amp.

A. Background On Biopotential Amplifiers:

Biopotential signals are very weak signals. Even the strongest EMG signal has a magnitude of less than 10 mV. Furthermore, the skin is a high impedence barrier, resulting in biosignals of low amplitude. Therefore, an EMG amplifier is usually required to have the following properties:

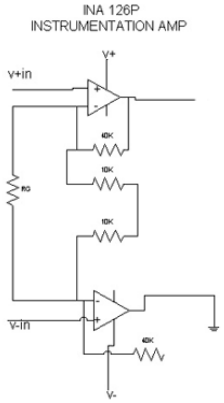
1. Capability to sense low amplitude signals in the range of 0.1 – 10 mV
2. Very high input impedance on the order of at least 10^9
3. Very low input leakage current, 1 μA or below
4. Flat frequency response of 0.1 – 100 Hz
5. A high common mode rejection ratio (CMRR)

B. Instrumentation Amplifier

The instrumentation amplifier used was the Burr-Brown INA126P, designed with two op-amplifiers and an external resistor set to a variable gain. We selected our gain to be approximately 1000, then determined the appropriate magnitude of resistance for the external resistor.

$$\text{Gain} = 1000 = 5 + (80\text{K}) / R_G \qquad (1)$$
$$\rightarrow \quad \mathbf{R_G = 80\ \Omega}$$

However, due to equipment restrictions, 80Ω was unavailable in lab, so we used a 75Ω external resistor. The external resistor itself provides a DC offset to the output voltage in the instrumentation amplifier.

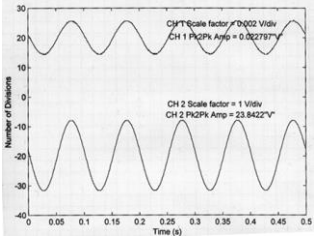


C. Characterizing the Instrumentation Amp:

After the circuit was designed and completed, we tested it by passing input signals at 100 Hz and observing the output signal. At 0.350 V, with a voltage divider, we were able to obtain a clean output signal. Using the oscilloscope, we measured the peak-to-peak amplitude of the output to be 23.85 V, with peak-to-peak amplitude of the input at 0.02322 V. The output to input gain ratio could then be calculated, as below:

Gain = (23.85) / (0.002322) = **1027**
(approximately 1000)

Characterization of Instrumentation Amp
Gain of INA126 at 10 Hz, 0.2 V



A voltage divider was created to prevent clipping of output signal. We used a resistor ratio of approximately 10KΩ : 1KΩ → actual resistor values of 8.1KΩ : 0.99KΩ.

D. Determining Bandwidth:

To determine sufficient bandwidth for measuring signal, we applied signals of increasing frequencies and measured the amplitudes of both input and output signal.

Freq (Hz)	Output Ampl (V)	Input Ampl (V)	Gain
10	23.84	0.022797	1046
20	23.91	0.02374	1007
50	23.8	0.0236	1008
100	25.6	0.02329	1099
200	25.9	0.02268	1142
500	24.1	0.0198	1217
1 K	14.4	0.0205	702

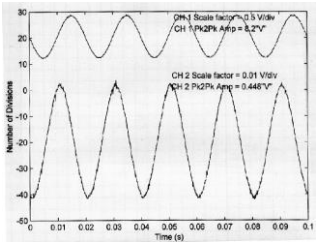
By comparing the output to input amplitude ratio, we monitored the gain, which remained relatively stable around 1000. At the frequency of 500 Hz, gain began to decrease significantly.

By using linear interpolation, we calculated the cutoff frequency to be 850 Hz. This was the point where the gain decreased to 860 (0.707 times the maximum value of 1217).

E. Measuring the Common Mode Rejection Ratio (CMRR)

CMRR is a measurement of a differential circuit's ability to reject a common mode signal V_{CM} at its inputs.

A high CMRR is essential since the capacitive coupling from the external electrical sources such as power lines would create a strong common mode signal in comparison to the differential ECG signal. A high CMRR would mean that the differential



gain is much larger than common gain.

CMRR : 25 dB at 50 Hz, 8.0 V

We measured input at 8.2 V and output at 456 mV, taken at 50 Hz, 8.0 V.

$$\textbf{CMRR} = (V_{in} / V_{out}) = (8.2\text{ V}) / (0.456\text{ V})$$
$$= \mathbf{25\ dB}$$

F. Measuring Input Impedance

In an amplifier, two desirable traits are high input impedance and low output impedance. To measure high input impedance in our circuit, we first took a reasonably large in magnitude resistor feeding into input. We chose a value of 4.13 MΩ. We then passed a signal into the amplifier and measured the voltages with respect to ground at the two points, on either side of the resistor. The voltage difference between the two was determined to be:

$$\mathbf{V_{diff}} = 5.43 - 5.0330 = \mathbf{0.397\ V}$$

By Ohm's Law, we calculated the amount of current passing through the resistor.

$$I = V / R = (0.397\ V) / (4.13 \times 10^{6}\ \Omega) = 9.6 \times 10^{-8}\ A$$

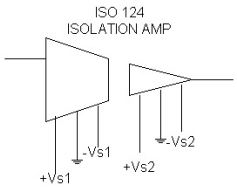
Use this current to find impedance of the loop, from the positive to negative terminals of the amplifier.

$$Z = V / I = (5.43\ V) / (9.6 \times 10^{-8}\ A) = 5.67 \times 10^{7}\ \Omega$$
$$\rightarrow \mathbf{0.567 \times 10^{8}\ \Omega}$$

IV. Final Check of Instrumentation Amplifier

After characterization, and determining CMRR and high input impedance, we ran through a final check to confirm our results. With the voltage divider included, we saw a clean, amplified signal at 10 Hz, 0.150 V. Input peak-to-peak was measured to be 23.2 mV, output peak-to-peak measured to be 18 V.

A. Isolation Amplifier



Isolation amplifiers are used to ensure safety. This is especially important in ECGs since highly amplified signals are dangerous to patients and defibrillator shocks are dangerous to the equipment. We used the Burr-Brown ISO124 to limit current and voltage in our system [3].

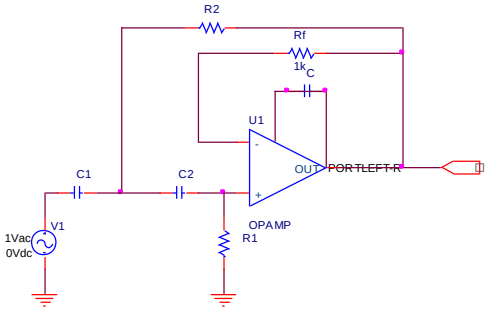
The isolation amplifier limits the current to +/- 5.0 mA on the patient side and +/- 5.5 mA on the instrumentation side.

III. B) CIRCUIT DESIGN: FILTERS

Design of Butterworth Maximally Flat, High-Pass and Low-Pass Filters

Design of High-Pass filter

Two filters are required to assemble the circuit that measures EMG signals. In the first stage, any DC offset in the signal must be eliminated before the next stage in the isolation amplifier. Using a high pass filter, a frequency cutoff was selected as 0.2 Hz to eliminate the DC drift. In this design, an active filter was used, consisting of an operational amplifier biased appropriately to generate the desired transfer characteristics. Below is the standard configuration for an active Butterworth filter. The Butterworth filter is chosen for its gain of 1 within the passband, and hence its adherence to maximally flat filter. This is especially important in reference to EMG signals, whose very low amplitude is easily subjected to distortions in the passband. This amplifier follows a 40dB/decade rolloff below the cutoff frequency ω, which is more than enough for the purpose of eliminating any DC bias entering the circuit.



The circuit selection process follows this set of design steps:

1. A cutoff frequency ω is chosen.
2. $C_1 = C_2$ and the capacitance value is selected according to the availability of parts.
3. $R_1 = 1.414/\omega C$
4. $R_2 = 1/2R_1$
5. $R_f = R_1$ minimizes DC offset

Using the above equations, a cutoff frequency of 0.2 Hz was used as a preliminary starting point. C_1, C_2, and C were selected to be equal to 10 μF each in accordance to the available discrete value capacitors. The following steps were used to achieve the desired results:

$$R_1 = 1.414/(2\pi C)\ R_1 = 1.414(2\pi * 10\mu F) = 22.5k\Omega$$

The closest 5% resistor is chosen at $R = 18k\Omega$

$$R_2 = 0.5R_1 = 9k\Omega$$

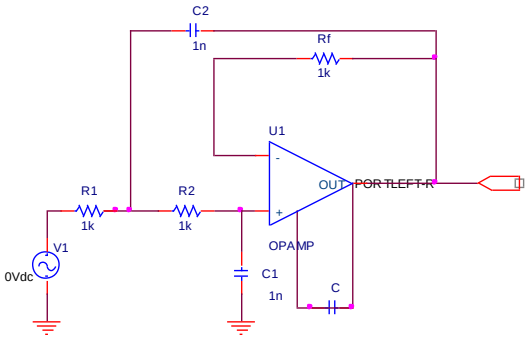
Closest 5% resistor $R_2 = 9.1k\Omega$

$$R_f = R_1 = 18k\Omega$$

Using the above filter template with the chosen component values, the filter was built and set as the second stage in the following cascade: 50 gain amplifier, high-pass filter, isolation amplifier and 1000 gain amplifier.

Design of Low-Pass Filter

In the same manner, a low-pass filter selection design process was followed to build a Butterworth filter with cutoff of 1000 Hz.



This particular frequency was chosen as a sufficient value that allows most EMG signals to be detected (most EMG signals fall between 30-40 Hz) at up to 3rd and 5th harmonics. The following selection process was used:

1. Choose cutoff frequency f_c
2. Let $R_1 = R_2 = R$, and choose available resistor values between $10k\Omega$ and $100k\Omega$
3. Calculate C_1 from

$$C_1 = 0.707/(\omega R)$$

4.

$$C_2 = 2C_1$$

Following the above steps, $R_1 = R_2 = 47k\Omega$

$C_1 = 0.707/(2\pi R) = 2.4\mu F$

5% value of C_1 chosen as $2.2\mu F$

$C_2 = 2C_1 = 4.4\mu F$

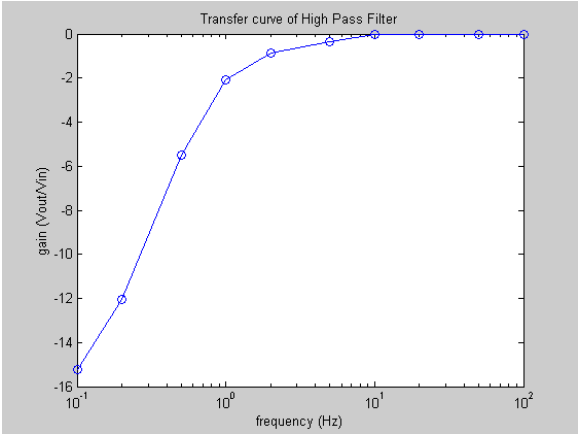
5% value of C_2 chosen as $4.7\mu F$

The filters were implemented with the National Semiconductor LF353 Wide Bandwidth Dual JFET input Operational Amplifier chipsets. Using these values for the filters, the results were evaluated in lab. The transfer characteristics were determined in lab via Matlab and displayed below. The transfer curves match the theoretical design, in that the gain is sufficiently undistorted in the passband, characteristic of Butterworth filters.

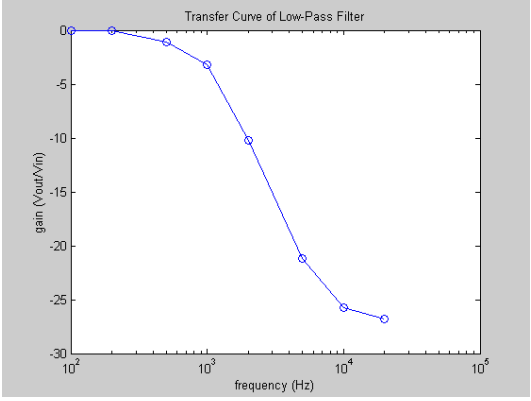
Filter Results

In the lab, the filters were built and tested. The transfer curves below illustrate the properties qualified in lab. The high-pass filter exhibited a – 30.2 dB/decade roll-off in the cutoff region. The low-pass filter exhibits a –30.3 dB/decade roll-off in the cutoff region.

B. Transfer Curve of High-Pass Filter



1) Transfer Curve of Low-Pass Filter



Although a –40dB/decade roll-off was not entirely achieved, the reason for the discrepancy is the fact that the values of

resistors were chosen to be within 5% of the determined values in accordance to the availability of discrete resistor and capacitor values. The corner frequency of the high-pass filter is found at $10^{0.06}$ Hz or 1.15 Hz. For the purpose of eliminating any DC offset, this will clearly suffice. The low-pass filter exhibits its corner frequency at $10^{2.7}$Hz or 501.18 Hz and cuts off completely at around 10^4 Hz. This value will suffice for the purposes of eliminating signals that are out of the frequency spectrum of most EMG signals.

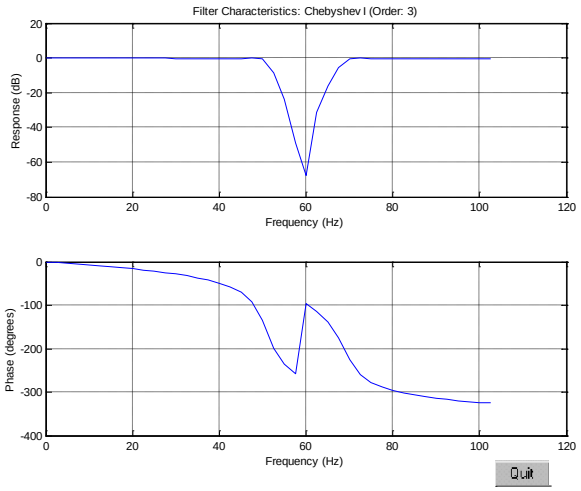
An important feature to note of the resulting transfer curves is the dB gain of 0 in the pass regions for both filters. More clearly demonstrated in the high-pass filter, it appears that the gain does not deviate from 0 dB. This allows the signal transfer without distortion of the low amplitude EMG signals. The same can be seen of the low-pass filter. Both adhere to the maximally flat characteristic of Butterworth filters. The high-pass filter was inserted before the instrumentation amplifier, and the low-pass filter was inserted before the isolation amplifier.

IV. Interpretation of Data

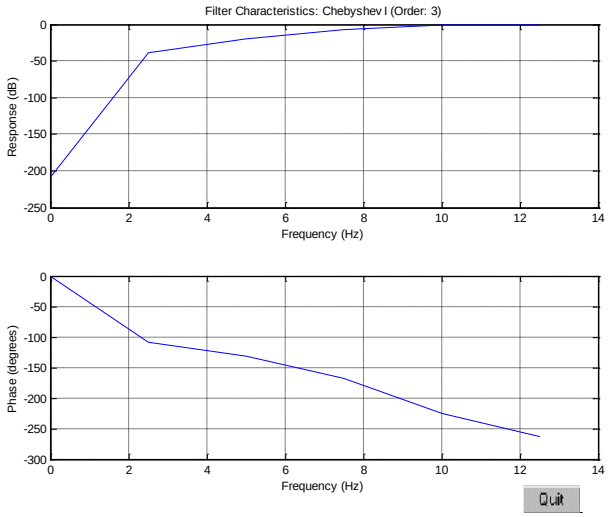
The following figures graphically represent data gathered in the EMG experiment. The graphs were generated and the filters applied in the EMAP 2000 software package. Due to limitations in the implementation of filters in EMAP, filter order was restricted to a value of 3 or less. We then generated .wav files of the data. To hear a signal, click on the name of the particular exercise. We used a consistent algorithm to determine which filters to apply to the signals.

First, we analyzed the frequency spectrum of the signal, looking for two obvious sources of noise.

60 Hz Noise
Many of the signals, especially those that had only light muscle activity had a significant energy peak at 60 Hz. We applied a stopband filter between 50-70 Hz to these signals. The attenuation curve of a 3rd order Chebychev I filter with cutoff frequencies of 50 and 70 Hz is shown below.



DC Drift
Many signals also had a DC drift. We looked for energy below 5 Hz in the frequency spectrum and a low frequency drift in the original signal to find DC drift. These signals were passed through a third order high pass filter. The attenuation curve for a 3rd order Chebychev I filter with cutoff frequency of 10 Hz is shown below.



To avoid redundancy, these steps are not graphically illustrated for every signal. The next step was to see if there was anything special about the particular signal that could be enhanced with another filter.

Specific Circumstances:

1. **Arm is resting on the table**

The original signal is very small, with a magnitude below 0.05 volts (Refer to Fig. 1). The frequency spectrum in Figure 2 shows that the vast majority of the energy of the filtered signal is at 60 Hz. After applying a Butterworth Low-Pass Filter, with cutoff frequency of 60 Hz and filter order of 3, most of the signal was attenuated. The resulting filtered signal has a magnitude less than 0.01 V. This is consistent with

expectations that there should be very little activity in a resting muscle.

2. **Squeezing an object at 1 Hz**

Figure 3 shows the data collected by squeezing a toilet paper roll at 1 Hz. The spectrum (Refer to Fig. 5) shows energy at 60 Hz and a large amount of DC drift. After removal of those two sources of noise, the final signal (Fig. 4) shows bursts with magnitudes between 0.1 V and 0.2 V occurring about every second. This result is consistent with the expected result of periodic muscle contractions due to the subject's mechanical movements.

3. **Continuous Squeezing**

The signal recorded for continuous squeezing of a toilet paper roll is shown in Figure 6. The frequency spectrum (Refer to Fig. 7) shows that there is significant noise at 60 Hz and at its 3rd harmonic of 180 Hz. We applied stopband filters to remove both sources of noise within a 20 Hz window.

The filtered signal is relatively constant and of fairly low magnitude. The reason the magnitude is low that the squeezing an object in the hand does not primarily use the biceps muscle. Forearm muscles, such as the *brachioradialis* and *extensor digitorum* [4] are much more involved.

4. **Raising and lowering light object (function generator)**

The data contains large DC drift because of motion of the electrodes. Figure 9 shows the original and filtered data, and Figure 10 shows the spectrum after removal of noise. There are slower 'bursts' with a magnitude of about 1 V. The reason that the peaks do not have a high amplitude relative to the rest of the signal is that the muscle is working throughout the entire process, thus contributing to a high level of background activity.

5. **Holding function generator continuously**

Figure 11 shows the final signal after removal of DC drift and 60 Hz noise. The signal is relatively constant with an amplitude around 0.3 V. There is a short burst of activity in the beginning of the recording. This was likely due to the muscle originally flexing to position and stabilize the object in the hand.

6. **Raising and lowering heavy object (chair)**

Figure 12 depicts the data gathered while raising and lowering a heavy object. The heavy object resulted in a strong signal. The signal to noise ratio was high enough that the 60 Hz noise was not significant and only the DC drift was removed by a high pass filter. Figure 13 shows the frequency spectrum. The filtered image has minimal energy in the lower frequencies. The final trace shows bursts of activity obtained during contraction (raising object) followed by periods of relaxation (lowering object). The amplitude of the signal is about 2 V, which is twice as high as the amplitude obtained for raising and lowering a light object.

7. **Holding chair continuously**

The data obtained for holding a heavy object continuously is shown in Figure 14. There is one continuous burst of energy with a magnitude around 1 V. This is similar in shape to holding a light object continuously, but this signal is larger in amplitude.

The 60 Hz noise and DC drift were significant. Figure 16 shows removal of DC drift, and Figure 15 shows the final filtered signal.

8. **One handed pushups**

An exceptionally high signal to noise ratio was obtained when the subject did one handed pushups. The DC drift was very significant here, as the subject was moving. After removal, a clean signal with an amplitude up to 4 V was observed (Refer to Figs. 17 and 18). The signal is relatively strong because most of the subject's weight is supported by his bicep.

V. Figures

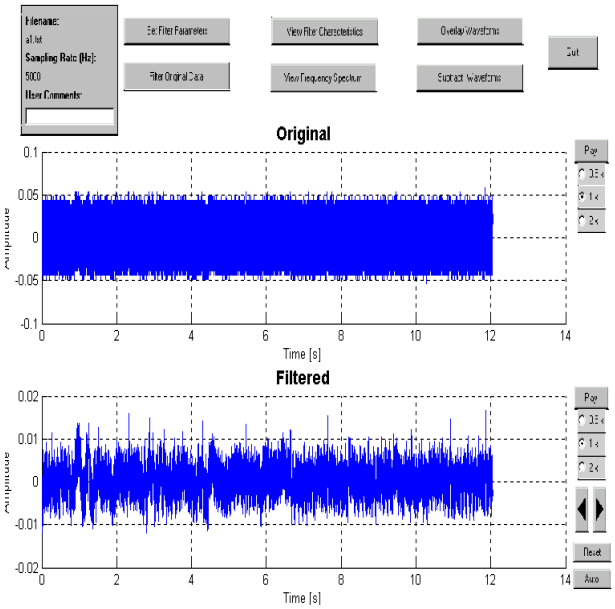


Fig. 1. Arm resting on the table. Filtered with Butterworth Low-Pass Filter, with cutoff frequency of 60 Hz and filter order of 3.

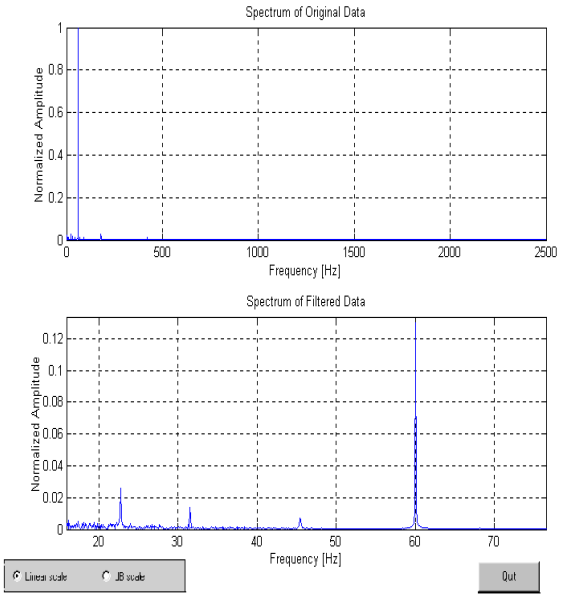


Fig. 2. Filter spectrum of subject's arm resting on the table.

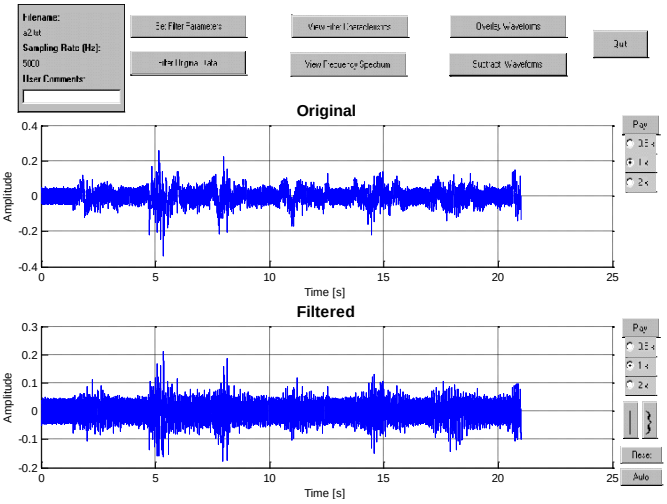


Fig. 3. Squeezing of a toilet roll, with one second squeeze intervals. Filtered to remove DC drift with a Chebychev I High-Pass Filter, with cutoff frequency of 5 Hz and filter order of 3.

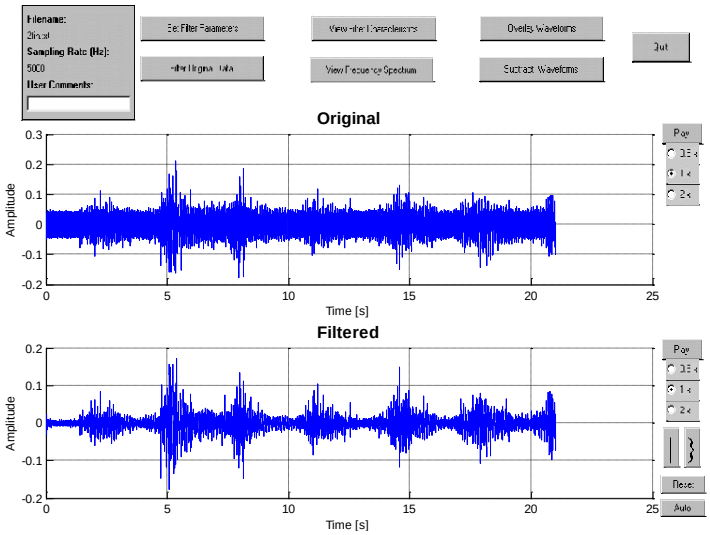


Fig. 4. Repeated filtering of squeezing a toilet roll with one-second intervals to remove more noise with Chebychev I notch Filter, with cutoff frequencies from 50 Hz-70 Hz and filter order of 3.

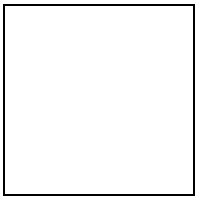


Fig. 5. Filter spectrum of squeezing a toilet roll with one-second intervals. Note the removal of DC drift

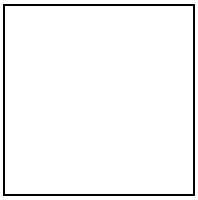


Fig. 6. Squeezing of a toilet paper roll continuously. Filtered with Butterworth Stop-Band Filter, with cutoff frequencies from 50 Hz-70 Hz and filter order of 3.

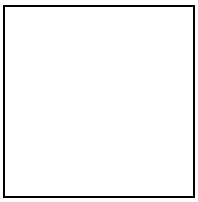


Fig. 7. Repeated filter of squeezing of a toilet roll continuously. Filtered with Butterworth High-Pass Filter, with cutoff frequency of 30 Hz and filter order of 3.

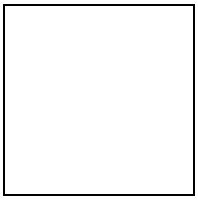


Fig. 8. Filter spectrum of squeezing a toilet roll continuously. Note removal of noise at 60 and 180 Hz.

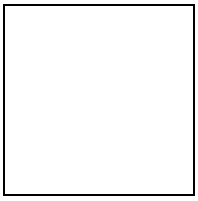


Fig. 9. Raising and lowering of a lightweight object. Filtered with a Chebychev I High-Pass Filter with cutoff frequency of 10 Hz and filter order of 3.

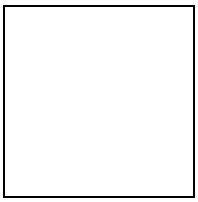


Fig. 10. Filter spectrum of raising and lowering a lightweight object. Note removal of DC drift and no significant noise in original signal at 60 Hz.

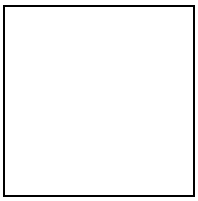


Fig. 11. Holding a lightweight object continuously. Top portion depicts filtered signal by Chebyshev I Stop-Band filter, with cutoff frequencies from 50 Hz-70 Hz and filter order of 3. Final filter (bottom) is done by Chebyshev I High-Pass Filter, with cutoff frequency of 10 Hz and filter order of 3.

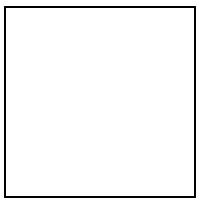


Fig. 12. Raising and lowering of a heavyweight object. Filtered with Chebyshev I High-Pass Filter, with cutoff frequency of 10 Hz and filter order of 3.

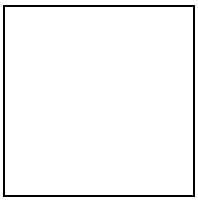


Fig. 13. Filter spectrum of raising and lowering a heavyweight object.

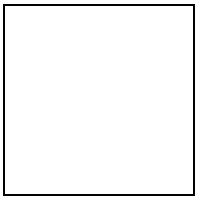


Fig. 14. Holding a heavyweight object continuously. Filtered with Chebyshev I Stop-Band Filter, with cutoff frequencies from 50 Hz-70 Hz and filter order of 3.

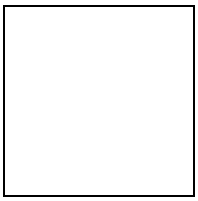


Fig. 15. Repeated filtering of holding a heavyweight object continuously. Filtered with Chebyshev I High-Pass Filter, with cutoff frequency of 10 Hz and filter order of 3.

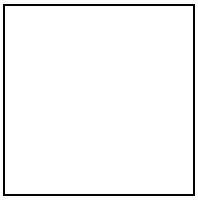


Fig. 16. Filter spectrum of holding a heavyweight object continuously. Note removal of 60 Hz and DC drift noise.

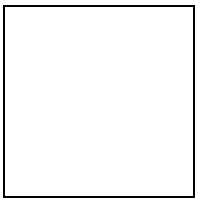


Fig. 17. Filter of the activity of one-handed pushups. Filtered with Chebyshev I High-Pass Filter, with cutoff frequency of 10 Hz and filter order of 3.

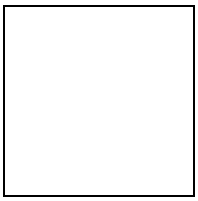


Fig. 18. Repeated filtering of one-handed pushups. Filtered with Chebyshev I Stop-Band Filter, with cutoff frequencies from 50 Hz-70 Hz and filter order of 3.

VI. Conclusion

The system described successfully measured EMG signals,

and after digitally filtering the signals, waveforms consistent with expected muscular activity were generated. The system performed significantly better when the signal to noise ratio was high (under heavy exertions of the biceps muscle).

ACKNOWLEDGMENT

ECE 511 Group V thanks Professor John Belina and Teaching Assistant Jeffrey S. Lee for the time spent in Phillips second floor lab, building the entire circuit and characterizing the amplifiers and filters.

REFERENCES

[1] http://www.ultranet.com/~jkimball/BiologyPages/E/ExcitableCells.html.
[2] http://www.ece.wpi.edu/~ted/research.
[3] http://www/s.ti.com/sc/psheets/sbos074/sbos074.pdf
[4] http://www.eaton.hands.com