Project Number: ***EXC – 11B2***

Design of a High–Resolution Surface Electromyogram (EMG) Conditioning Circuit

A Major Qualifying Project Report

submitted to the Faculty of

WORCESTER POLYTECHNIC INSTITUTE

in partial fulfillment of the requirements for the

Degree of Bachelor of Science

by

Marzhan Mozhanova

Date: January 4, 2012

Approved:

Professor Edward A. Clancy

Advisor

Acknowledgements

The author would like to thank:

Professor Ted Clancy for his guidance and continued support throughout the development and completion of this project.

Thomas Angelotti for his advice regarding components used in the project's design.

Abstract

The electromyogram — the electrical activity of skeletal muscle — finds useful applications in many fields, such as biomechanics, rehabilitation medicine, neurology, gait analysis, exercise physiology, pain management, orthotics, incontinence control, prosthetic device control, even unvoiced speech recognition and man–machine interfaces. An electromyogram signal detection and conditioning circuit has been designed by Prof. E. Clancy in 2003. Although the circuit's performance characteristics are satisfactory, the circuit is large in size and power consumption in comparison to technology that may be used today to perform the same functionality. The goal of this project was to investigate technological advances that may be used to improve Clancy's design size, power consumption and, potentially, performance. This research can then be used to design a new electromyogram signal conditioning circuit that would have at least the same performance characteristics, will be reduced in size and power and be possible to manufacture and maintain in an academic setting.

Table of Contents

Table of Figures

Table of Tables

Glossary of Terms

Biomechanics – the study of the application of mechanical laws and the action of forces to living structures

CMRR (Common Mode Rejection Ratio) – in differential amplifiers, value that measures the tendency of the device to reject input signals common to both input leads

Electromyography – recording of the changes in electric potential of muscle by means of surface or needle electrodes

Endomysium – delicate bands of connective tissue interspersed among muscular fibers

Fibrillation potential – spontaneously contracting muscle fibers, which are a sign of interruption of the nerve connection to an organ

Gait analysis – study of a particular manner of walking

Neurology – a medical specialty concerned with the study of the structures, functions, and diseases of the nervous system

Motor unit – a grouping of muscle cells and the motor neuron that innervates them

Motor neuron – neurons which activate muscle cells

Muscle crosstalk – a phenomenon in which signal recorded over one muscle was in fact generated by a neighboring muscle and conducted to the recording electrodes

Muscle fascicles – a bundle of muscle cells that are associated functionally

Muscle fiber – a muscle cell

Orthotics – also called orthetics, the science that deals with the use of specialized mechanical devices to support or to supplement weakened or abnormal joints or limbs

Perimysium – the connective tissue demarcating a fascicle of skeletal muscle fibers

1. Introduction

Surface electromyography (sEMG) is the non–invasive recording of electrical muscle activity that is used to diagnose neuromuscular disorders, among other applications. Muscle fibers are activated by motor neurons and the resulting electrical signals produced by the muscle fibers can be detected by electrodes placed on the surface of the skin. EMG finds many useful applications in biomechanics, rehabilitation medicine, neurology, gait analysis, exercise physiology, pain management, orthotics, incontinence control, prosthetic device control, even unvoiced speech recognition and man–machine interfaces (Merletti *et al.*, 2004, Rechtien *et al.*, 1999).

Most sEMG signals have a frequency content ranging from 0 to 500 Hz, with dominant energy between 50 to 150 Hz. However, content at up to 2000 Hz may be useful. The amplitude of the signal may vary from less than 50 μV up to 30 mV (Clancy *et al.*, 2002). The nature of sEMG signals and various EMG equipment properties determine the quality of a recording. Besides motor unit (muscle cell group) and electrode characteristics, there are other factors that distort and add undesirable noise to the signal. A DC offset due to half–cell potentials within the tissues can be as high as 300 mV. Muscle crosstalk (signals from motor units of neighboring muscles) may result in misleading information about the investigated muscle. Ambient 60 Hz noise from power supplies (50 Hz for European power supplies) may result in power line noise three fold the magnitude of the sEMG itself. Inherent noise generated by electronic equipment can range from zero to thousands of Hz. Motion artifacts due to electrode or cable movement add noise to the EMG signal in the frequency range of 1 to 5 Hz (Merletti *et al.*, 2004).

Typically, sEMG is acquired in two stages. The first (electrode–amplifier) stage includes signal transduction/detection and pre–amplification. This stage is placed as close to the skin as possible, generally with the electrodes and the electronic amplification arranged within a single package. The second (signal conditioner) stage provides further signal conditioning. An electrode–amplifier circuit that detects the EMG signal, amplifies and removes some of the unwanted noise was designed by E.A. Clancy of Worcester Polytechnic Institute in 2003. The electrode–amplifier has a high input impedance to limit current drawn from the subject and therefore minimize signal distortion and attenuation. Low output impedance drives the following electronic stage without change in output voltage. Modern operational amplifiers have an input impedance of over 100 MΩ and output impedance under 100 Ω. A gain of 20 is implemented to improve electronic performance within the electrode–amplifier (increase common mode rejection ratio, decrease noise) and decrease inherent noise in subsequent stages. The amplifier is situated very close the electrodes, insuring the most precise information about the sEMG prior to amplification and diminishing noise from cable movement. This configuration is preferred by many researchers because it attenuates noise and motion artifact by buffering the acquired signal near the source and amplifying the signal early in the process; and minimizes physical dimensions of a device by only including a pre–amplification stage near the subject. To prevent hazardous current from entering the subject, the power supply of all components on the subject's side of the circuitry is isolated from earth ground. The design by Clancy also includes an eight channel signal conditioning circuit. Each channel begins with an eighth–order high pass filter with cutoff frequency of 15 Hz and gain of 10, to attenuate motion artifact and any DC offset. The channel continues with selectable gain of 1, 2, 4, 8, 16, 32 or 64 to optimize the signal range for subsequent digitization and allow biasing of the following stage. The signal is then electrically isolated via opto–coupling. Power and signal on the subject's side of the isolator are referenced to an isolated ground; the other side of the isolator is earth–referenced. The final stage is fourth–order low pass filtering to reject noise from frequencies above the EMG signal frequency range of about 1800 Hz and for anti–aliasing purposes (Clancy 2003).

Although the existing device has a satisfactory performance, it was designed in 2003. The circuit requires a relatively large power supply and has relatively large physical dimensions. It is also comprised of a large number of parts. The goal of this Major Qualifying Project was to investigate advances in EMG instrumentation and find alternative ways to design a sEMG recording device. The electrode–amplifier circuit is sufficiently small and meets necessary signal detection and transmission parameters. Therefore, it is the signal conditioning circuit that needs alteration to minimize size and power consumption. Depending on the results of the investigation, parts of or the entire existing EMG signal conditioning circuit will be modified. Possible modifications include finding alternative power and signal isolation options, analog–to–digital conversion options, decreasing size and power consumption of the front–end or even finding implementations that would permit elimination of some of the signal processing stages. Improvements can be achieved by various methods which may include hardware modification, using integrated circuits, new components of smaller size and power supply needs, new components that permit the combination of multiple functions, and a PC–based system. The design should meet or exceed performance characteristics of the existing device. Performance specifications of the current device by Clancy include a common mode rejection ratio of minimum 100 dB at 60 Hz, high input impedance of each of the front–end channels, ability to accept signals with amplitude in the range of 0 to 30 mV with 16–bit resolution and frequency range of up to 1800 Hz, accommodate DC offsets up to 300 mV, eliminate DC offsets and attenuate motion artifacts, and have a noise power less than 1,5 μV RMS referred to the input over the frequency range of 20 to 500 Hz. It is highly desirable that the new design is manageable to build and maintain in an academic setting.

Upon the completion of the research, two major electromyograph circuit designs were created. Both design's approaches to the problem were essentially very similar, as their initial and final stages included the same type of filtering and were storing collected data on a medically isolated personal computer. The main differences between designs included approaches to signal amplification and digitization.

While this chapter gives a brief introduction to the project, Chapter 2, Background, is dedicated to EMG signal physiology and a more detailed analysis of the existing circuit. The Problem Statement contains main goals the design needs to satisfy, the Methodology is a detailed description of the approaches for the two new designs and Chapter 5, Discussion, summarizes advantages and disadvantages of the designs.

2. Background

Designing a functional system for EMG signal analysis is impossible without understanding the physiology of an EMG signal. The Background contains information about the structure of skeletal muscle, EMG signal generation and physical parameters, factors that affect a signal and various types of noise the signal acquires while propagating through an electromyograph.

2.1 EMG Physiology

Muscle contraction and relaxation is controlled by the central nervous system. The nervous system sends a signal through a motor neuron to a grouping of muscle cells, called fibers. That grouping of muscle cells, and the motor neuron that innervates them, is a motor unit — a basic building block of the neuromuscular system, the smallest functional part of muscle tissue. The signal of the motor neuron causes a chemical reaction that changes the membrane potential of muscle fibers. If the threshold potential is reached a motor unit action potential (MUAP) occurs, causing the electrical activation to spread along the entire surface of the muscle fiber at a rate of approximately 3–5 m/s. A muscle contraction twitch results (Martini, 2005). Figure 1 is an image of a motor neuron connected to a group of muscle fibers.

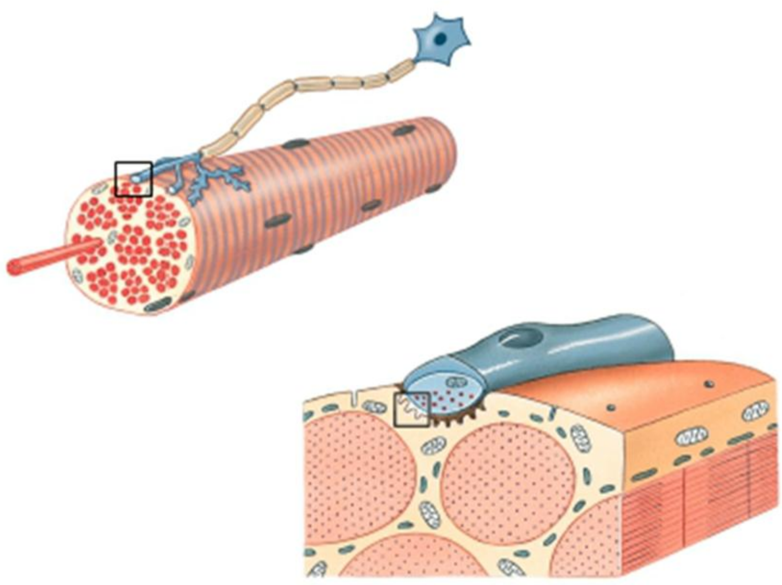

Figure 1 – Motor Neuron Attached to a Motor Unit

(Martini, Frederic H. "Chapter 10. Muscle Tissue." *Anatomy & Physiology*. San Francisco: Benjamin Cummings, 2005.)

Figure 2 depicts the composition of a skeletal muscle. A muscle consists of bundles of muscle fibers that are wrapped by endomysium and form muscle fascicles. Muscle fascicles are wrapped by a connective tissue called perimysium (Mann).

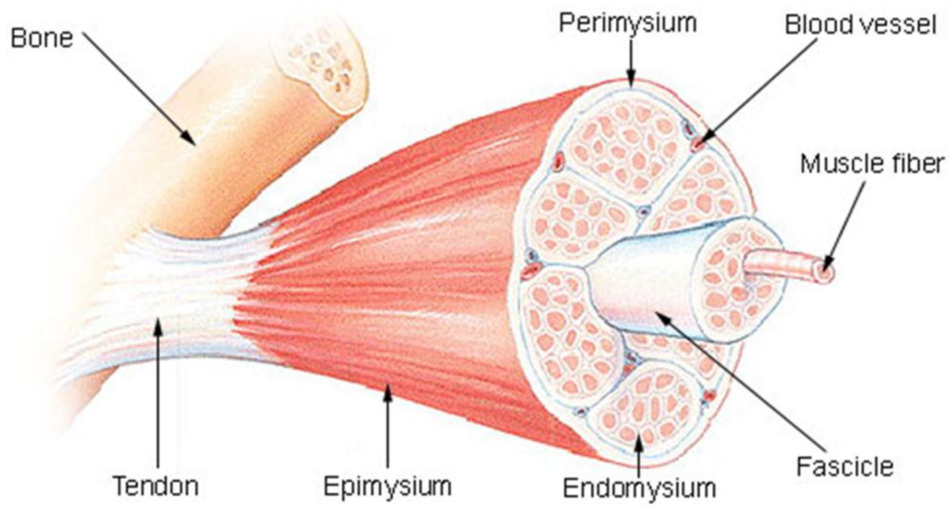

Figure 2 – Skeletal Muscle Structure

(Martini, Frederic H. "Chapter 10. Muscle Tissue." *Anatomy & Physiology*. San Francisco: Benjamin Cummings, 2005.)

Electromyography (EMG) is a technique that is used to detect and record MUAPs non–invasively by placing a conductive element (electrode) on the surface of the skin overlying the muscle of interest (this method is referred to as surface EMG) or directly inside the muscle (this invasive EMG detection method utilizes needles or fine wires). Inserted electrodes record from a very small region of muscle, making it possible to view individual MUAPs. On the contrary, surface electrodes tend to record from much larger regions of the muscle. Thus, individual MUAPs are not clearly visible, as many motor units tend to be contracting concurrently. Only the superposition (interference) pattern is recorded. Needle EMG gives access to deep musculature, has higher sensitivity and higher frequency and voltage ranges. Needle electrodes are suitable for detecting changes in motor unit size and internal structure, as well as revealing their abnormal function. Unlike surface EMG, needle EMG can detect action potentials of spontaneously contracting muscle fibers — fibrillation potentials, which are a sign of interruption of the nerve connection to an organ. Surface EMG is better for gathering data on: various aspects of behavior; temporal patterns of activity, or fatigue of the muscle as a whole or of muscle groups; location of the innervation zone; length and orientation of fibers; and highly accurate measurement of muscle–fiber conduction velocity. Surface electrodes are indispensable when needle measurement is not appropriate or measurements need to be made repeatedly (Stoykov, 2005; Merletti *et al.*, 2004). Surface electrodes do not require medical supervision, cause minimal discomfort and are easier to apply than wire electrodes, although they are mainly used on superficial muscles, have limitations with recording dynamic muscle activity and have concerns with crosstalk of adjacent muscles. A needle and surface electrode comparison by Giroux and Lamontagne (1990) revealed no significant differences in either isometric or dynamic muscle contraction detection. However, needle EMG acquisition has several disadvantages that can make use of surface EMG more desirable: detection may not represent an entire muscle while repositioning is nearly impossible; insertion of the needle into the muscle requires sterilization of the instrumentation and the environment; needles are difficult to use on muscles located in sensitive areas, such as muscles in the tongue, lips, and face; there is minor muscle tissue damage, that in turn affects shapes of action potentials; although low, there is a risk of infection; needles might not be suitable for children or persons who poorly tolerate procedures involving use of needles (Groh). The two methods are better suited for a different span of applications and have their own advantages and disadvantages and are therefore both currently used for EMG signal detection. The instrumentation required for this project is dedicated entirely to noninvasive (surface) EMG methods (Mademli, 2010).

The amplitude of a sEMG signal may vary from less than 50 μV up to 30 mV (DeLuca, 1979). It is commonly accepted that most sEMG signals have frequency content between 0–500 Hz with dominant energy between 50–150 Hz. However, there exists content at up to 2000 Hz that may be useful (Clancy *et al.*, 2002). There are numerous physiological and non–physiological factors that influence sEMG signal interpretation, such as electrode shape, size, inter–electrode distance, skin contact and location over the muscle; motor unit physiology; subject's muscles' properties; etc. As a sEMG signal travels through different media it acquires noise. There are several types of electrical noise which affect a sEMG signal.

Ambient noise. Ambient noise lies in a wide range of frequencies but its dominant component is 50 Hz or 60 Hz which is the most common source of electrical noise in the EMG signal and corresponds to power line noise. Such noise results in a signal whose voltage can be larger than the EMG signal itself. Proper skin preparation, shielding electrode leads and using a differential amplifier with a CMMR of at least 100 dB at 50/60 Hz can help attenuate this type of noise. A right leg drive circuit incorporated into front–ends of biometrical equipment also helps to reduce common–mode interference. Since the dominant energy of the EMG signal is located in the 50–150 Hz range some experts do not recommend the use of a 50 or 60 Hz notch filter as it partially removes frequency components adjacent to the unwanted ones. However, when skin preparation and quality equipment is not sufficient to attenuate ambient noise and if only a rough EMG signal amplitude is sought, a sharp notch filter may be acceptable. An important issue to keep in mind is that the power frequency harmonics can contain more power than the fundamental frequency, therefore repeated notch filtering (at the harmonic frequencies) may be necessary (Clancy *et al.*, 2002).

Motion artifact. This type of noise is caused by the movement at the interface between the detection surface of the electrode and the skin. It generally lies in the 1–10 Hz frequency range and has a voltage comparable to the magnitude of the EMG. This noise can be reduced by high–pass filtering frequencies in the 1–10 Hz range. As has been mentioned previously, skin preparation can reduce the electrode–skin impedance and also helps to minimize motion artifacts. A better alternative for motion artifact attenuation is an electrode connected to an operational amplifier with high input and low output impedance. This design will have a high impedance at the electrode and low impedance at the cable, serving as an impedance transformer. The displacement current flows through low impedance to ground and minimizes the noise. In addition, because the electrode–amplifier acts as an impedance transformer, less skin preparation is required.

DC offset potential. Oil secretions and dead skin cells increase impedance on the outermost layer of the skin, which cause DC voltage potential generation during skin and electrode contact of up to 200–300 mV. This DC potential is common to all electrodes and can be minimized with proper skin preparation. The quality of contact is typically reduced by at least a factor of 10 with proper preparation (Merletti and Migliorini, 1998). There are no set procedures; application and desired quality of the collected data determine the degree of skin preparations. Usually skin cleaning involves use of special abrasive and conductive pastes to remove dead skin or fine sandpaper and alcohol swabs to clean the skin, all to lower skin impedance. If not addressed, high electrode–skin impedance may result in signal amplitude reduction, distortion of the waveform and power line interference (Quach, 2007).

Muscle crosstalk. Muscle crosstalk is a phenomenon in which signal recorded over one muscle was in fact generated by a neighboring muscle and conducted to the recording electrodes. Crosstalk can be minimized by choosing electrode size and inter–electrode distances carefully, as well as via the placement of the electrodes. Smaller inter–electrode distances tend to lead to less crosstalk. But, overly small distances can lead to electrode shorting (e.g., due to sweat). The recommended inter–electrode distance is typically 1–2cm or the radius of the electrode. Electrode alignment with the direction of muscle fibers increases the probability of detecting the same signal (Clancy *et al.*, 2001; Sellers, 2011).

Inherent noise. Inherent noise of the electronics instrumentation. Any electronic equipment will generate noise up to thousands of Hertz. Although this type of noise cannot be eliminated, modern electronics tend to have noise less than 1.5 mV RMS (referred to the input) over the band from 20–500 Hz (Clancy *et al.*, 2000; Thought Technology, 2008).

.

2.2 Existing Design

This section provides a description of the existing electrode amplifier and conditioning circuit designed by Clancy. It depicts the composition of each functional block in the circuit. The electrode–amplifier and its signal conditioning circuit are comprised of five main blocks: electrode–amplifier array, high pass filter, selectable gain amplifier, signal path isolator and low pass filter; and accommodates eight channels for inputs from eight electrodes. The subject's side of the circuit, comprised of the electrode–amplifier, high–pass filter, selectable gain amplifier and the input side of the signal isolator, is powered with a medically isolated power supply. The output side of the isolator and the low–pass filter are powered with earth–referenced power.

The electrode–amplifier has a gain of twenty and will not be modified for any of the design alternatives. It utilizes an "active electrode" (an instrumentation amplifier placed very close to the electrodes' detection surfaces) to detect and amplify an EMG signal at the surface of the skin. The impedance at the junction of dry skin and electrode surface may range from kiloohms to megaohms. To minimize current drawn from the subject and therefore to prevent attenuation or distortion of the signal, the input impedance should be as large as possible. An instrumentation amplifier is very suitable for this purpose since it rejects much of the common signal; the common mode interference is greatly attenuated. Figure 3 illustrates the connection of a differential amplifier to the skin. Two inputs of the amplifier are connected to the electrodes. Two signals are detected and the difference is amplified. Any "common" signals (such as power line interference) will be removed (or greatly attenuated) and the physiologic signal that is different at the two sites will be amplified. Any signal that originates far away from the detection surfaces will appear as a common signal, while signals in their vicinity will be different and will be amplified.

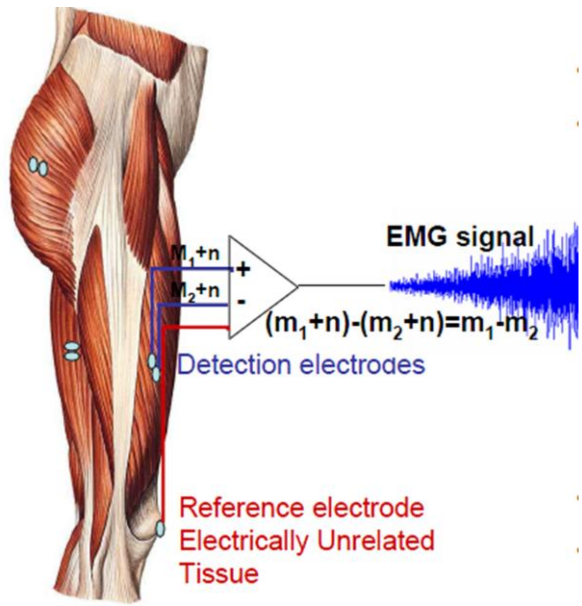

Figure 3 – Schematic of the Differential Amplifier Configuration

(Mademli, Lida." The physiological background of EMG". *Centre of Research & Technology – Hellas Informatics & Telematics Institute. 2010)*

The need for high input impedance introduces capacitive coupling at the input of the amplifier. A small capacitance between the input of the amplifier and the power line will introduce power line interference. Hence, very close placement of the amplifier to the electrode is desired. Additionally, the output impedance of an amplifier is low, which drives the following electronic stage without change in output voltage. Prior to any further conditioning, the detected signal should be amplified to counteract noise it will acquire propagating through further stages. Higher gain (above DC) generally provides better noise performance and CMMR within the instrumentation amplifier. These improvements are non–linear. For example (using the Analog Devices AD620 Datasheet), the noise density at 60 Hz is 75 $nV/Hz^{1/2}$ for a gain of one, 10.3 $nV/Hz^{1/2}$ for a gain of 10, and 9 $nV/Hz^{1/2}$ for a gain of 100; the CMRR at 60 Hz is 95 dB for a gain of one, 112 dB for a gain of 10, and 120 dB for a gain of 100. Figure 4 is a graphical representation of the voltage noise dependence on the frequency and the gain.

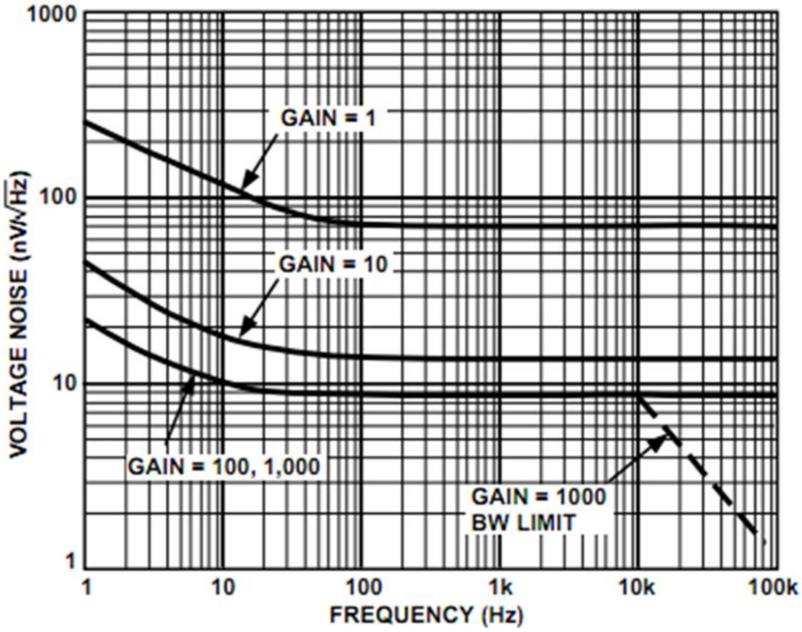

Figure 4 – AD 620 Instrumentation Amplifier Voltage Noise Spectral Density vs. Frequency (G= 1–1000)

(Analog Devices. "AD620: Low Drift, Low Power Instrumentation Amp with Set Gains of 1 to 10000 Datasheet." Web. 20 Dec.2011. <http://analog.com>)

However, because of a large DC offset potential of a sEMG signal (200–300 mV), the instrumentation amplifier of the electrode–amplifier cannot have a large DC gain (e.g., much higher than about 10–20) or the amplifier will saturate. This issue can be dealt when using the AD620 by selecting the impedance Z_G to be purely resistive and small, so that the transfer function is a constant, low value (e.g., 10–20). In addition, the noise contribution of latter stages is diminished by a larger first–stage gain in the instrumentation amplifier. One technique to manage the offset problem while achieving substantial gain in the passband is to use a complex impedance for Z_G. In particular, if impedance Z_G is formed as the series combination of a resistor R_G and capacitor C_G, then

$$Z_G = R_G + \frac{1}{j\omega C_G} \quad \text{and}$$

$$\frac{v_{Out}}{e_2 - e_1} = 1 + \frac{2 \cdot R_1}{R_G + \frac{1}{j\omega C_G}},$$

where v_{Out} is the AD620 output voltage and e_1 and e_2 are the differential inputs (Clancy, 2003). Figure 5 is schematic representation of the AD620 instrumentation amplifier. The amplifier is programmed for a gain of 20 (R_G = 2050 Ω) with the capacitor value of C_G = 22 μF.

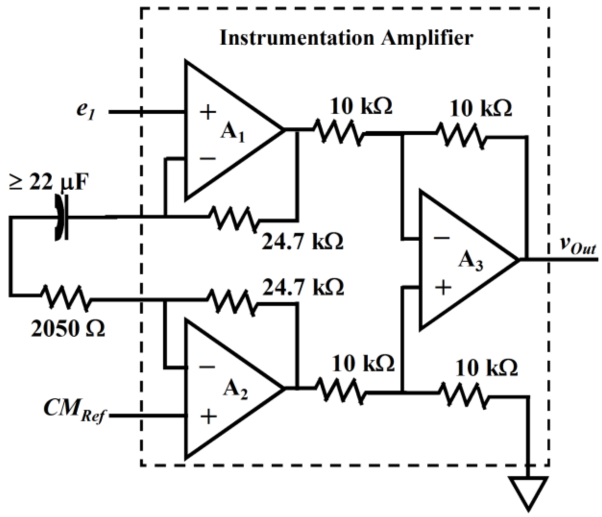

Figure 5 – Single Channel Electrode–Amplifier Design with a High Pass Characteristic

(Clancy, E. A. "Design of a High–Resolution, Monopolar, Surface Electromyogram (EMG) Array Electrode–Amplifier and Its Associated Signal Conditioning Circuit." TS. Worcester Polytechnic Institute, Worcester, MA.)

Design by Clancy implemented a gain of 20 due to the fact that to achieve a larger gain a rather large C_G capacitor value is required. An instrumentation amplifier with gain of 100 required a capacitor C_G as large as 100 µF, else the passband characteristic would attenuate relevant EMG frequencies. Capacitors with such large value had large dimensions in 2003. But now there are 100 µF 10 V ceramic capacitors as small as 3.20mm × 1.60mm. Using a 100 µF capacitor of such dimensions lets an amplifier gain increase from 20 to 100 and a gain of one at 0 Hz. This way, although the offset potential may be large, the actual sEMG signal having been amplified by a factor of 100 will be significantly larger than the offset. A larger instrumentation amplifier gain has better noise characteristics – a 25 percent improvement of 12 $nV/Hz^{1/2}$ at the gain of 20 down to 9 $nV/Hz^{1/2}$ at the gain of 100. Implementing a higher gain electrode amplifier can be considered in future circuit improvements.

After the electrode–amplifier, the signal enters the signal processing circuit, one channel of which is shown in Fig. 6. The circuit is comprised initially of an eighth–order Sallen–Key high pass filter with a cutoff frequency of 15 Hz and embedded gain of 10. The filter diminishes signal components with the frequency of 0–15 Hz. That portion of the signal contains motion artifact caused by movement at the interface between the detection surface of the electrode and the subject's skin. The DC offset potential (0 Hz) is also rejected by the high pass filter. The signal is further amplified by a factor of 10. The high pass filter and electrode–amplifier produce a total gain of 200. Increasing the signal voltage is an important step in sEMG signal conditioning as EMG signals have very small amplitudes. Amplification optimizes signal resolution for digitizing equipment. This choice of gain is adjusted to maximize the signal without saturation for digitization in latter stages of signal conditioning. Next, the filtered amplified signal enters a selectable gain amplifier with gains of 1, 2, 4, 8, 16, 32 or 64, with the gain of one for signals exhibiting the maximum amplitude and gain of 64 for signals exhibiting the minimum amplitude. Selectable gain requires an operational amplifier and seven gain selection buttons for each of the channels. This design is high in cost and occupies a significant amount of space. Overall maximum gain of the existing circuit ranges from 200–12,800, which is the product of the signal amplification gain of 20 by the electrode–amplifier, gain of 10 by the high–pass filter and gains 1–64 of the selectable gain amplifier. The signal path is then isolated with an Agilent HCNR201 photocoupler, transferring the desired signal to the earth reference side of the circuitry using light waves. This measure prevents hazardous current from entering the subject. After the signal passes though necessary conditioning stages, its components outside of the EMG signal physiologic frequency range are attenuated with a low pass filter with a cutoff frequency of 1800 Hz. The existing design utilizes a fourth–order, Sallen–Key, low–pass filter with cutoff frequency of 1800 Hz as shown.

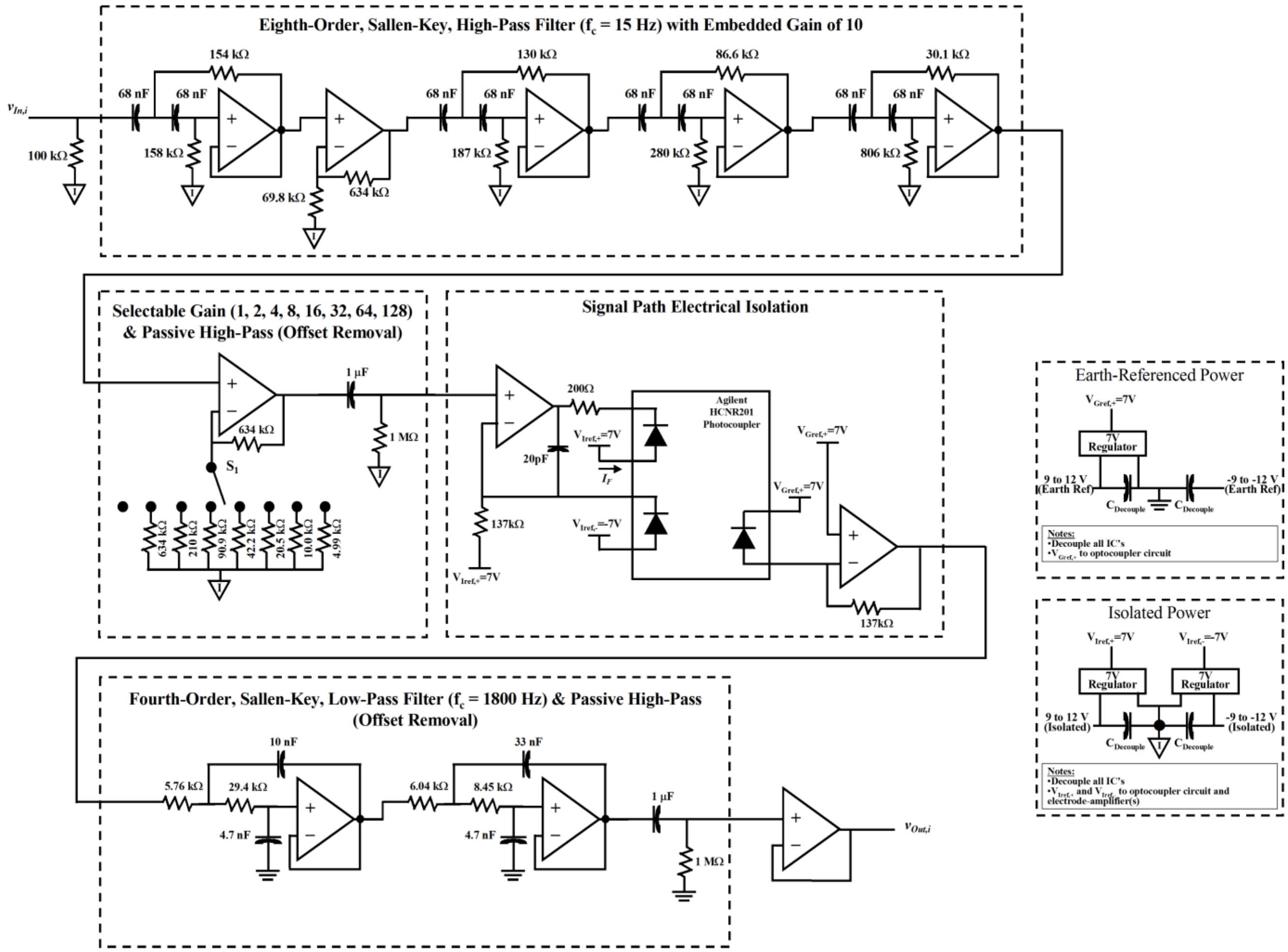

Figure 6 – Schematic of the sEMG Conditioning Circuit by Clancy

(Clancy. "Design of a High–Resolution, Monopolar, Surface Electromyogram (EMG) Array Electrode–Amplifier and its Associated Signal Conditioning Circuit". 2003)

3. Problem Statement

Given the vast range of EMG applications, it is hard to overestimate the importance of accurate sEMG signal recording. While performance and safety are the most crucial parameters in sEMG recording device design, operation and maintenance simplicity, physical dimensions and power consumption characteristics are also important. There are potentially numerous ways to implement a sEMG signal conditioning circuit. The goal of this project was to find a solution with best performance specifications that is manageable to build and maintain in an academic setting. This design is an open–ended project, therefore while there are no specific guidelines to the approaches for the circuit design, the main objective is to improve the existing design by making it simpler, smaller, more energy–efficient and easy to operate, maintain and modify in the future. As has been mentioned in Chapter 1, Introduction, the newly designed circuit must meet or exceed performance specifications of the existing signal conditioning circuit. **Error! Reference source not found.** outlines performance specifications that are satisfied by the electrode–amplifier and that need to be satisfied by the sEMG conditioning circuit.

Table 1 – Design Specifications

	Characteristic	Value
Electrode–Amplifier	Input signal voltage range	0–30 mV
	Input signal frequency	≤1800 Hz
	Amplification	20
	CMRR at 60 Hz	100 dB
	Noise power (25–500 Hz)	≤1.5 mV RMS
	Input impedance	≥100 MΩ
Conditioning Circuit	Input signal voltage range	±9 V
	Input signal frequency	15–1800 Hz
	Amplification	1, 2, 4, 8, 16, 31, 64
	Noise attenuation	Motion artifact, DC offset, anti–aliasing
	Maximum DC offset	±300 mV (referred to the input)
	ADC resolution	≥16 bits
	Isolation	Signal, power on subject's side
	Filtering	Eighth–order Butterworth high–pass filter Fourth–order Butterworth low–pass filter

4. Methodology

The methodology is comprised of three main sections: analog–to–digital conversion, design #1 description, and design #2 description. The analog–to–digital conversion section is meant to give an understanding of how circuit hardware modifications can be compensated by the resolution of the signal's analog–to–digital converter.

4.1 Analog–to–Digital Conversion

This section elaborates on trading off the use of a lower–order high–pass filter for motion artifact attenuation with an additional bit of resolution and using a 24–bit ADC to implement the gain selection stage digitally.

4.1.1 ADC Resolution and Use of Lower–Order High–Pass Filter

To minimize the amount of analog hardware, the eighth–order high–pass Sallen–Key Butterworth filter with cutoff frequency of 15 Hz and gain of 10 can be partially implemented in hardware prior to signal digitization, with the remaining filtering implemented in software.

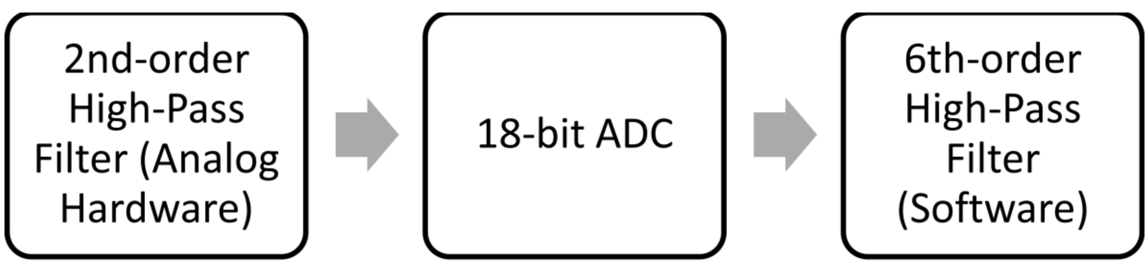

Figure 7 – Diagram of Software and Hardware Sections of the 8th–order High Pass Filtering Using an ADC

This filter implementation lets a portion of the interference at low frequencies that would have been attenuated by a higher order filter, enter the ADC with the rest of the signal. To prevent the signal from saturation the gain should be reduced accordingly. The effect of the lower gain on the resolution can be counteracted by a higher resolution ADC. Once the signal is digitized, the remaining filtering can be applied digitally. Separated filter stages overall will then have the characteristics of an eighth–order high–pass filter and the resulting signal will fit back into 16–bit resolution.

The main objective when implementing this design is to determine the reduction in signal attenuation due to the use of a lower–order filter and the necessary increase in ADC resolution to compensate that reduction. A function was written in MATLAB to design the complete eighth–order high–pass filter. The function also extracts the frequency response corresponding to the first two, four and six orders of the complete filter. These responses are equivalent to building only the first stage (two orders), first two stages (four orders), or first three stages (six orders) of the eighth–order analog filter. Fig. 8 – Fig. 10 contain the magnitude responses of these filters and the ratio of the eighth–order filter to the remaining filter stages.

For example, Figure 8 shows the ratio plot of the second–order analog “extracted” filter, i.e., the first two orders (first stage) of the complete analog filter. The left graph shows the signal cutoff frequency at 22.7 Hz. The right graph shows the ratio of the magnitude response of the eighth–order filter to remaining six orders of the complete filter, with 18 Hz being the "worst-case" frequency at which that ratio is at its maximum (approximately 1.7 for the second–order filter implementation). Note the graph on the right is also the magnitude response of the remaining six orders of the original high−pass filter.

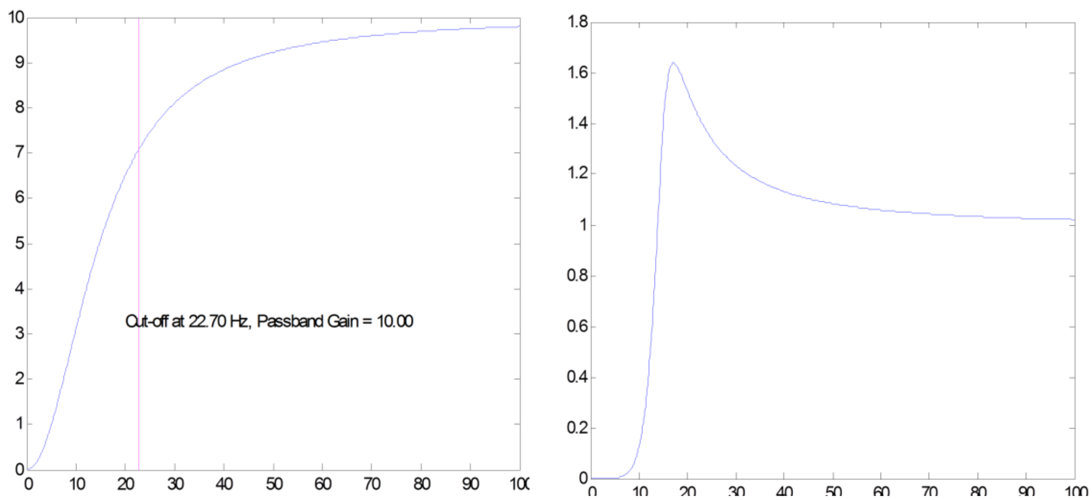

Figure 8 – Magnitude Response of the First Two Orders of the Complete Eighth–Order Analog High–Pass Filter (left) and the Magnitude Response Ratio of the Complete Eighth–Order Response Divided by Remaining Six Orders (right).

These results show that when using a second–order extracted filter, the largest signal will still only be amplified by a factor of 1.7 in the worst case scenario (in relation to the complete eighth–order filter), in contrast to a ratio of approximately 2.5 (worst case) for the fourth–order extracted filter (Figure 9) and 2.7 (worst case) for the sixth–order filter (Figure 10). Hence, use of only the first two filter orders (first analog filter stage) is clearly the best option for this scheme. The maximum additional signal gain, compared to the complete eighth–order high–pass filter is less than 1.7. For discussion hereafter, we will round this value to an addition gain of less than two.

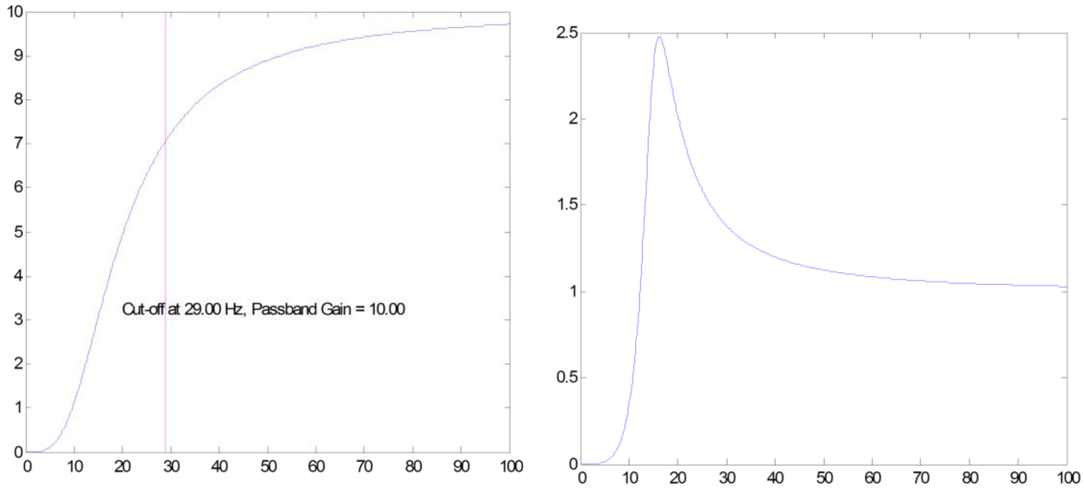

Figure 9 – Magnitude Response of the First Four Orders of the Complete Eighth–Order Analog High–Pass Filter (left) and the Magnitude Response Ratio of the Complete Eighth–Order Response Divided by Remaining Four Orders (right).

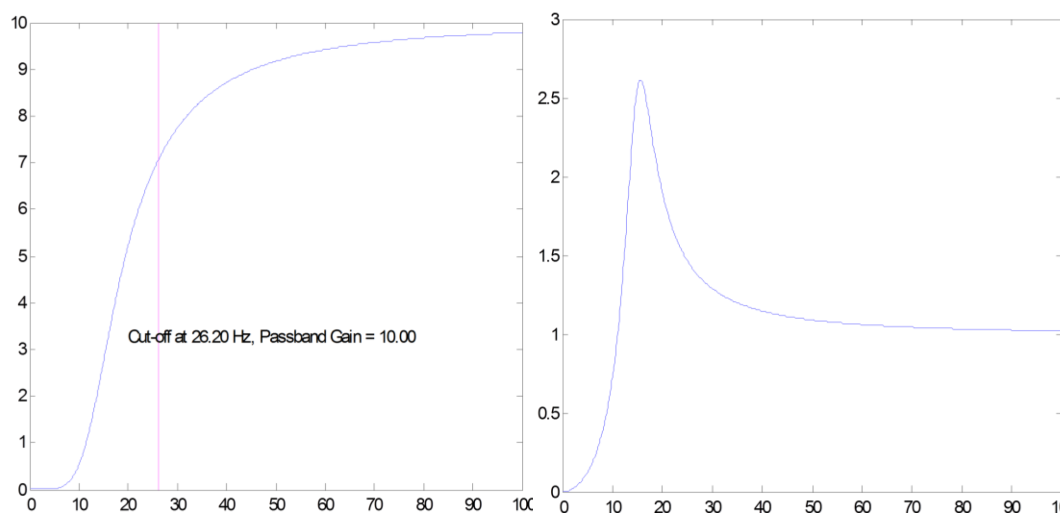

Figure 10 – Magnitude Response of the First Six Orders of the Complete Eighth–Order Analog High–Pass Filter (left) and the Magnitude Response Ratio of the Complete Eighth–Order Response Divided by Remaining One Order (right).

Thus, a decrease in the gain of 10 in the high–pass stage by a factor of two is necessary to prevent saturation at the ADC. Accordingly, the ADC resolution should be doubled, resulting in 2^{17}=131,072 digital values instead of 65,536 values for 16–bit conversion. Doubling the resolution requires one more bit in addition to the minimum 16 bits specified by the performance requirements. The aforementioned modifications will require 17–bit resolution sampling, scaling the signal by a factor of 1/2 to compensate for a second–order extracted filter amplification and still retaining 16–bit sampling resolution for the complete system.

For example, consider a signal propagating though the *original* eighth–order filter. Let the signal be at the frequency of 18 Hz, which is the worst–case frequency when using a second–

order extracted analog high–pass filter. Since an 18 Hz signal is near the cutoff frequency of the complete high–pass filter, it will be amplified by a factor of approximately 7.07 (10 is the passband gain of the filter and 0.707 is the cutoff frequency gain of a normalized high–pass filter). With no further gain and ADC range of ±5 V the maximum input to the high–pass filter is then ±5 V/7.07 = ±707 mV. Then, consider applying an *extracted* second–order analog high–pass filter to the ±707 mV, 18 Hz signal. This filter passband gain of five (half of the original gain) is followed by an ADC with an additional bit of resolution. As has been shown, approximately twice the original gain of the extracted second–order high–pass filter applied to a 18 Hz is compensated by the passband gain being reduced by a factor of two (meaning the signals in the passband will only receive half of the gain compared to the complete eighth–order analog filter). The extra bit of resolution of the ADC makes up for the gain difference. Therefore, the signal output from the extracted second–order filter will still fit in the ±5 V range of the ADC.

The remaining six orders of the eighth–order Sallen–Key high–pass filter will be applied in software to the signal after the AD conversion. The exact new composition of the six filter orders will need to have a passband gain of one and a cutoff frequency of approximately 15 Hz. The precise shape of this digital filter must match that of the remaining six orders of the original analog filter. This filter can be designed off-line once when the instrumentation is being developed, and then used thereafter. Various techniques exist to produce the design, given the known frequency response of the remaining six analog filter orders.

4.1.2 ADC Resolution and Selectable Gain Implementation

Another way to minimize signal conditioning circuitry in hardware is to implement selectable gain in software using an ADC with a higher resolution than defined by the problem statement. Consider the case of a subject who produces maximum contraction of a monitored muscle such that the contraction gives a sEMG signal of maximum amplitude. Let this maximum amplitude be ±25 mV. A signal with a maximum amplitude would be acquired by the existing analog hardware using the minimum selectable gain of 200 (gain of 20 in the electrode amplifier, gain of 10 in the high–pass filter, minimum selectable gain of one). So, a signal with range of ±25 mV x 200 = ±5 V would be acquired at the 24–bit ADC. After sampling the signal for a short amount of time, for example five seconds, each sample of the digitized signal of interest is stored in the 16 higher–order bits (bit positions 9–24 highlighted with color blue on the image below), keeping in mind the desired signal resolution is 16 bits. Had a 16–bit ADC been used, then all 16 bits would provide the recorded signal.

LSB	2^2	2^3	2^4	2^5	2^6	2^7	2^8	2^9	2^{10}	2^{11}	2^{12}	2^{13}	2^{14}	2^{15}	2^{16}	2^{17}	2^{18}	2^{19}	2^{20}	2^{21}	2^{22}	2^{23}	MSB

Now consider a subject who produces a smaller maximum input signal, for example, a signal with half of the input voltage range of ±12.5 mV. With the existing hardware, the signal will be amplified by a factor double the gain for a maximum amplitude signal. Hence, a gain of

400 (gain of 20 in the electrode amplifier, gain of 10 in the high–pass filter, selectable gain of two) would be selected. A signal with range of ±12.5 mV x 400 = ±5 V would be acquired at the 16–bit ADC. Instead, let the revised hardware implement a fixed signal path gain of 200 (i.e., the gain needed to satisfy the largest possible physiologic signal), but with the 24–bit ADC. With a signal path gain of 200, the signal range of the ±12.5 mV EMG at the (24–bit) ADC would be ±2.5 V. Hence, the MSB of the ADC would always be equal to zero. Therefore, 16 usable bits of resolution can be achieved by ignoring the MSB and keeping the prior 16 bits (i.e., keeping bits 8–23) of each sample.

LSB							2^8								2^{16}							2^{23}	MSB

In this way, we have traded one bit of ADC resolution for a gain of two in the analog hardware. In the same way, this argument can be repeated to show achieving an analog gain of four by ignoring the two highest bits of the ADC; a gain of eight can be achieved by ignoring the three highest bits of the ADC; … an analog gain of 64 can be achieved by ignoring the six highest bits. If more significant bits are ignored, much greater analog gain can be implemented; therefore even smaller signals can be recognized by the ADC. If the lowest 16 bits are considered, an analog gain of 256 can be realized, permitting a minimum amplitude of the signal at the input of the electrode–amplifier of 0.1 mV.

If the EMG data being acquired do not require simultaneous display, the aforementioned technique can be used to implement selectable gain in an adaptive manner. After a complete data epoch is acquired at 24–bit resolution, the epoch can be analyzed to determine the voltage range of the signal. Once the range is known, appropriate amplification gain can be selected by ignoring the higher bits according to the desired gain value. A header file would be written to identify which gain to apply to the sEMG signal stored in the data file. Consider if a sEMG signal with maximum of ±12.5 mV is acquired. It is stored in a data file and the desired gain of two is calculated. The gain of two implies that bit positions 8–23 are used to express the signal. The header file will then have the information about which of the 24 bits are used for digitization of this particular signal. Alternatively, only the necessary sixteen bits need be stored in the final data file, along with the appropriate gain in an associated header file.

4.2 Alternative Design #1

The following two sections of the methodology chapter describe the two alternative sEMG signal conditioning circuit designs. The first design consists of four main blocks: first two orders of the complete analog eighth–order Sallen–Kay high–pass filter, fourth–order low–pass filter, an 18–bit multichannel ADC and a laptop computer. The laptop computer is powered by a medically isolated power supply (or simply powered from its battery), with no hardwired connections to other external devices or networks. The laptop computer may be replaced by a desktop personal computer connected to an AC outlet through a medically isolated transformer.

The system needs to accommodate eight input channels. Consider signal propagating through each of the system blocks separately.

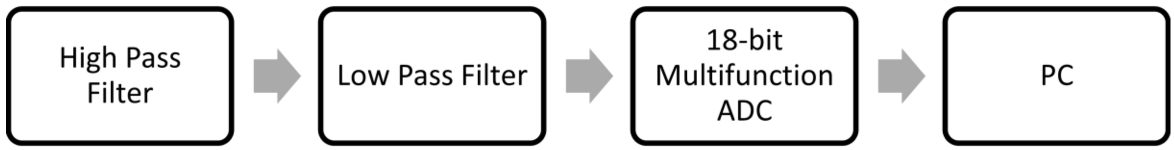

Figure 11 – Alternative Design #1 Block Diagram

High–pass filter. For each input channel, implement the first two orders of the complete eighth–order Sallen–Key high–pass filter with an embedded gain of five using an amplifier, two capacitors of 68 nF, two resistors of 58.5 kΩ and 416 kΩ, and embedded gain with two resistors with value ratio 4/1, for example 40 kΩ and 10 kΩ. This design requires eight operational amplifiers or two Texas Instruments TL074 quad JFET low noise operational amplifier ICs. Each IC will require a supply voltage of ±18 V, having power dissipation of 680 mW. The high–pass filters will amplify signal from the electrode–amplifier by a factor of five and attenuate frequency components below 22.7 Hz, taking care of the motion artifact. One TL074 op amp has a length of 6.2mm and width of 0.68mm. The signal will then propagate through the low–pass filter to the ADC.

Low–pass filter. For anti–aliasing purposes, frequencies outside of the sEMG physiologic range need to be filtered out. The existing fourth–order low–pass filter is sufficient and trivial, so it will be used prior to signal digitization.

Analog–to–digital conversion. In this design National Instruments USB 6289 High–Accuracy 18–Bit Multifunction DAQ for USB serves the purposes of selectable gain and analog–to–digital conversion. The operation of the device can be controlled in various software applications, such as LabVIEW, Visual Studio .NET, ANSI C/C++, C#, or Visual Basic using Measurement Studio. This data acquisition device has up to 32 analog inputs with a multichannel sampling rate of 500 kS/s (500 kHz), which is well over the required total throughput of 8×4 kHz = 32 kHz (to accommodate eight channels with signals, satisfying Nyquist rate of twice the maximum frequency of 2 kHz). The device also accommodates a selectable range of inputs of ±10 V, ±5 V, ± 2 V, ±1 V, ±0.5 V, ± 0.2 V, ± 0.1 V. These gains do not precisely match those of the existing system, but are comparable. The range of these gains is 100, while the range of gains in the existing circuit is 64 (each method has seven gain selections). Once a subject produces a maximum amplitude muscle contraction, the voltage range of the input signal is known, so the DAQ can be adjusted accordingly to the input range in software. Once the input range is selected, a signal is digitized using 16 bits and an additional bit to compensate for use of a second–order extracted high–pass filter (see the detailed discussion on using 17–bit ADCs in Section 4.1.1 ADC Resolution and Use of Lower–Order High–Pass Filter). For this device, 18 bits are available. The resulting signal is transmitted through a USB 2.0 to a laptop. USB can be operated at Hi–Speed (up to 480Mbit/s) or full speed (12Mbit/s). Each ADC sample is 32 bits, which means, accounting for the need of throughput of 32 kHz, USB speed is sufficient. As using full speed lowers the device performance, Hi–speed will be implemented, giving us the throughout well over the desired 8 × 4 kHz = 32 kHz. USB 6289 requires a 20 W, 11 to 30 V power supply, locking or non–locking power jack with 0.080 in (2.032m) diameter center pin, 5/16–32 thread for locking collars and a 2 A, 250 V power supply fuse. The power supply will need to be electrically isolated. Power for the analog electronics will also need to be electrically isolated, although much less power is needed compared to the existing system.

Signal and power isolation. Remaining signal conditioning and storage. The digitized signal enters an isolated laptop or a desktop PC, where the remaining high–pass filtering is applied in software for motion artifact attenuation. The last six orders of the complete eighth–order filter attenuate signals below 15 Hz and have a passband gain of one. Once this last signal conditioning is applied to the digitized signal it can be stored or displayed using a laptop or a desktop PC. Both types of computers require a medical grade power isolation transformer, such as Tripp Lite 120 V 1000 W 4 Outlet hospital medical grade isolation transformer (the laptop can operate from its battery). A transformer will need several outlets to accommodate any peripherals a desktop computer might have, such as a display, printer, etc. The computer cannot be hardwire connected to anything but the EMG conditioning circuit and medically isolated power supply while connected to a subject.

4.3 Alternative Design #2

The second design consists of three main blocks: first two orders of the complete analog eighth–order Sallen–Key high–pass filter identical to the one in design #1, multi–channel front end with a 24–bit ΣΔ ADC IC and a laptop computer. The laptop computer is powered by a medically isolated power supply (or via its battery). The laptop computer may be replaced by a desktop personal computer connected to an AC outlet through a medical isolator transformer. The system needs to accommodate eight input channels. Consider a signal propagating through each of the system blocks.

Figure 12 – Alternative Design #2 Block Diagram

High–pass filter. The implementation of this block has been stated in Section 4.2. Alternative Design #1.

Analog–to–digital conversion. In this design, Texas Instruments low–power, eight–channel, Texas Instruments ADS 1298 24–bit analog front–end IC with a ΣΔ ADC and an onboard low–pass filter will be used for signal digitization. This IC accepts eight analog inputs with a maximum data rate of 32 kS/s. The device has an on–board low–pass filter, which rejects high–frequency noise and prevents aliasing. In this way, the need for a hardware fourth–order Sallen–Key low–pass filter is eliminated. The filter needs to be programmed and the cutoff frequency needs to be selected with the rest of the programming of the ADC. The filtering has a start–up transient, so the ADC needs to be run for some time before data start being stored. The 24–bit ADC can be used to replace selectable gain in hardware as described in Section 4.1.2 ADC resolution and Selectable Gain, the low–pass filter and permits software implementation of the remaining six orders of high–pass filter.

Signal and power isolation. Remaining signal conditioning and storage. The process is identical to the process descried in Section 4.2 Alternative Design #1.

4.3 Noise Calculations

In this section voltage noise referred to the input will be discussed. The noise will be estimated over the range of 20–500 Hz (commonly accepted range of useful frequencies of a sEMG signal) and 20–1800 Hz (the range of frequencies that are considered useful in this project). Resistor, capacitor and cable noise will be neglected in this argument.

The first design consists of the electrode amplifier, the first two orders of the complete eighth–order high–pass filter, a fourth–order low–pass filter, National Instruments 18–bit ADC and a PC. As has been previously mentioned, when using a gain of 20 the AD620 instrumentation amplifier has a voltage noise (referred to input) of 12 $nV/Hz^{1/2}$. Two orders of an eighth order high pass filter consist of two capacitors, four resistors and one operational amplifier. Analyzing the graph of noise to frequency ratio of a Texas Instruments TL074 quad op–amp, at the gain of one it has a noise of approximately 25 $nV/Hz^{1/2}$. The fourth–order low–pass filter requires two operational amplifiers. The same operational amplifiers as for the high–pass filter will be used and therefore the same amount of noise (25 $nV/Hz^{1/2}$) is specified. The second design only has the high–pass filter stage, constructed from same TL074 op–amps, and the ADC chip, that has most of noise within 1 $nV/Hz^{1/2}$ range.

To calculate the total noise of the system, first the electrode-amplifier stage noise is calculated, while noise values of subsequent circuit stages are divided by the total signal amplification gain starting from the input up to the component that is being analyzed. So, consider the total circuit noise of design #1 and design #2 at frequencies 20–500 Hz (total of 480 Hz) and 20–1800 Hz (total of 1780 Hz). Calculations shown in Table 2 demonstrate that the noise performance of the two designs differ only by tens of nV RMS (referred to the input). The Total Voltage Noise is found as the RMS sum of the noises from each stage. Note that the Total Voltage Noise is almost entirely specified by the noise attributed to the electrode-amplifier stage.

Table 2 – System Noise of Designs #1 and #2

		Electrode–Amplifier	**High–Pass Filter and HPF Gain=5**	**Low–Pass Filter**	**Total Voltage Noise**
Design #1	20–500 Hz	$12nV/\sqrt{Hz}\times\sqrt{480} = 0.263\ \mu V_{RMS}$	$25nV/\sqrt{Hz}\times\sqrt{480} = 0.548\ \mu V_{RMS}$ $0.548/20 = 0.027\ \mu V_{RMS}$	$25nV/\sqrt{Hz}\times\sqrt{480}/ = 0.548\ \mu V_{RMS}$ $0.548/100 = 0.005\ \mu V_{RMS}$	**0.2645 μV_{RMS}**
	20–1800 Hz	$12nV/\sqrt{Hz}\times\sqrt{1780} = 0.506\ \mu V_{RMS}$	$25nV/\sqrt{Hz}\times\sqrt{1780} = 1.05\ \mu V_{RMS}$ $1.05/20 = 0.053\ \mu V_{RMS}$	$25nV/\sqrt{Hz}\times\sqrt{1780} = 1.05\ \mu V_{RMS}$ $1.05/100 = 0.011\ \mu V_{RMS}$	**0.5089 μV_{RMS}**
		Electrode–Amplifier	**High–Pass Filter and HPF Gain=5**		
Design #2	20–500 Hz	$12nV/\sqrt{Hz}\times\sqrt{480} = 0.263\ \mu V_{RMS}$	$25nV/\sqrt{Hz}\times\sqrt{480} = 0.548\ \mu V_{RMS}$ $0.548/20 = 0.027\ \mu V_{RMS}$		**0.2645 μV_{RMS}**
	20–1800 Hz	$12nV/\sqrt{Hz}\times\sqrt{1780} = 0.506\ \mu V_{RMS}$	$25nV/\sqrt{Hz}\times\sqrt{1780} = 1.05\ \mu V_{RMS}$ $1.05/20 = 0.053\ \mu V_{RMS}$		**0.5089 μV_{RMS}**

5. Discussion

Both designs require less hardware implementation than the original design. Removal of six stages of the high–pass filter (total of six quad operational amplifiers), programmable gain with switches and the optical isolator reduces the size of the design. Elimination of the optical isolation stage also decreases power consumption significantly, due to removal of high–current LEDs. Individual designs have advantages over each other.

Design #1 requires slightly more hardware for low–pass filter implementation and has a higher implementation cost due to the high cost of a USB6289 DAQ device. However, the design is simple to implement as the DAQ utilizes software which is fairly simple to use and is available for students and faculty. In addition, if the high–pass and low–pass filters are replicated, the device can accommodate up to 32 inputs.

Design #2 only requires a high–pass filter stage to be implemented in hardware and has a much lower implementation cost. However, programming the ADS1298 ADC is fairly complicated and will require a lot more non–recurring engineering. In addition, the ADS1298 IC will need to be incorporated into the hardware, as well as it various control lines and control logic.

Another aspect to consider is the total cost of the design. A new design accounts for being constructed in an academic setting, which means much lower labor price. Both designs use a very small amount of capacitors and resistors, so consider that the total price of implementing the high–pass and low–pass filters is very cheap. So, the main expenses of design #1 are the ADC, PC and the isolation transformer, resulting in approximately $1200 for the ADC, $430 for the isolator transformer and a varying cost of a personal computer depending on the consumer. The main expenses of the design #2 will total up to approximately $50 for the ADC, $430 for the isolator and a varying cost of a personal computer depending on the consumer. (Price estimates for year 2012.)

Analysis of the voltage noise due to the amplifiers showed that nearly all of the noise is attributed to the electrode-amplifier circuit. The subsequent electronics (high–pass filter, selectable gain filter, low–pass filter) contributed a negligible amount of noise to the signal chain. Additional reduction in noise might be achieved by increasing the gain of the electrode-amplifier circuit or by choosing a lower noise instrumentation amplifier. Additional noise reduction within the signal processing circuitry has little influence on the total noise.

Both of the designs satisfy the desired specifications. The choice of the design is dependent on whether the implementation cost is a priority and lower labor cost is available, and whether it is more desirable to use off–shelf components for the implementation.

Bibliography

Analog Devices. "ADuM4160 Full/Low Speed 5 kV USB Digital Isolator Datasheet." Web. 20 Dec.2011. <http://analog.com> .

Analog Devices. "AD620: Low Drift, Low Power Instrumentation Amp with Set Gains of 1 to 10000 Datasheet." Web. 20 Dec.2011. <http://analog.com>.

Cantrell, Mark. "Digital Isolator Simplifies USB Isolation in Medical and Industrial Applications". *Analog Dialogue 43–06.* June 2009.

Clancy, E. A. "Design of a High–Resolution, Monopolar, Surface Electromyogram (EMG) Array Electrode–Amplifier and Its Associated Signal Conditioning Circuit." TS. Worcester Polytechnic Institute, Worcester, MA.

Clancy, E. A., E. L. Morin, and R. Merletti. "Sampling, Noise–reduction and Amplitude Estimation Issues in Surface ElectromyographySampling, Noise–reduction and Amplitude Estimation Issues in Surface Electromyography." *Journal of Electromyography and Kinesiology* 12 (2002): 1–16.

De Luca, Carlo J., Alexander Adam, Robert Wotiz, Donald Gilmore, and S. Hamid Nawab. "Decomposition of Surface EMG Signals." *Journal of Neurophysiology* 96.3 (2006): 1646–657.

De Luca, Carlo J. "Physiology and Mathematics of Myoelectric Signals." *IEEE Transactions on Biomedical Engineering* BME–26.6 (1979): 313–25.

De Luca, Carlo J. "SURFACE ELECTROMYOGRAPHY: DETECTION AND RECORDING". *Delsys Inc.* 2002. < http://www.delsys.com/Attachments_pdf/WP_SEMGintro.pdf> 24 Nov. 2011.

Farina, D., R. M. Enoka, and R. Merletti. "The Extraction of Neural Strategies from the Surface EMG." *Journal of Applied Physiology* 96.4 (2004): 1486–495.

Gilmore K., Meyers J. "Using Surface Electromyography in Physiotherapy Research." The Australian Journal of Physiotheray 29.1. February 1983.

Groh, David. "Electromyography (EMG) Instrumentation." *SlideServe*. Web. 24 Nov. 2011. <http://www.slideserve.com/Samuel/electromyography–emg–instrumentation>.

Mademli, Lida." The physiological background of EMG". *Centre of Research & Technology – Hellas Informatics & Telematics Institute.* 2010.

Mann, Michael D. "Chapter 14. Muscle Contraction." *The Nervous System In Action*.

Martini, Frederic H. "Chapter 10. Muscle Tissue." *Anatomy & Physiology*. San Francisco: Benjamin Cummings, 2005.

Medical Dictionary. <http://medical-dictionary.thefreedictionary.com/>. Web. 6 Jan. 2012.

Merletti, Roberto, and Philip Parker. *Electromyography: Physiology, Engineering, and Noninvasive Applications*. Hoboken, NJ. IEEE/John Wiley & Sons, 2004. Print.

Merletti, Roberto. "Standards for Reporting EMG Data." *Journal of Electromyography and Kinesiology* 6.2 (1996): 3–4. *ISEK*.. 31 Nov. 2011.

National Instruments. "High-Accuracy M Series Multifunction DAQ for USB - 18-Bit, up to 625 kS/s, up to 32 Analog Inputs Datasheet." Web. 20 Dec.2011. <http://ni.com>

Pullman, S. L., D. S. Goodin,, A. I. Marquinez, S. Tabbal, and M. Rubin. "Clinical Utility of Surface EMG. Report of the Therapeutics and Technology Assessment Subcommittee of the American Academy of Neurology." *Neurology;55:171–177*. 2000. Web. 28 Nov. 2011.

Quach, Jee Hong. "Surface Electromyography: Use, Design & Technological Overview." 10 Dec. 2007. Web. 30 Nov. 2011. <http://www.bfe.org/articles/Surface%20Electromyography%20–%20Use,%20Design%20&%20Technological%20Overview.pdf>.

Reaz, M. B. I., M. S. Hussain, and F. Mohd–Yasin. "Techniques of EMG Signal Analysis: Detection, Processing, Classification and Applications (Correction)." *Biological Procedures Online* 8.1 (2006): 163. *PubMed*. Web. 26 Nov. 2011. <http://www.ncbi.nlm.nih.gov/pmc/articles/PMC1455479/>.

James J. Rechtien, DO, PhD; Jeffrey B. Gelblum, MD; Andrew J. Haig, MD; and Andrew J. Gitter, MD. " TECHNOLOGY REVIEW: DYNAMIC ELECTROMYOGRAPHY IN GAIT AND MOTION ANALYSIS." Muscle Nerve 22: Supplement 8: S233-S238, 1999

Rubio, Graciela. "Improvement of the electrode–amplifier circuit for an electromyogram recording device." Thesis. Worcester Polytechnic Institute, 2009. *Gordon Library Database*. Web. 26 Nov. 2011.

Sellers, Bill. "Introduction to EMG." Web. 27 Nov. 2011. <http://mac–huwis.lut.ac.uk/~wis/lectures/>.

Stan Jou S., Schultz T. "Automatic Speech Recognition based on Electromyographic Biosignals." Carnegie Mellon University, Pittsburgh, PA, USA; Karlsruhe University, Karlsruhe, Germany. 2008

Soderberg, Gary L., and Thomas M. Cook. "Electromyography in Biomechanics." *Journal of Americal Physical Therapy Association* 64 (1984): 1813–820.

Stoykov, Nikolay S., Madeleine M. Lowery, Charles J. Heckman, Allen Taflove, and Todd A. Kuiken. "Recording Intramuscular EMG Signals Using Surface Electrodes." *Proceedings of the 2005 IEEE 9th International Conference on Rehabilitation Robotics* (2005). Web. 31 Nov. 2011.

Thought Technology Ltd. " Surface Electromyography Applied to Psychophysiology." 2008 Web. 24 Nov 2011.

Texas Instruments. "ADS1294 Low–Power, 8–Channel, 24–Bit Analog Front–End for Biopotential Measurements Datasheet." Web. 27 Nov. 2011.

Texas Instruments. " TL071, TL071A, TL071B, TL072, TL072A, TL072B, TL074, TL074A, TL074B Low-Noise JFET-Input Operational Amplifiers Datasheet." Web. 23 Dec. 2011 <http://ti.com>

Appendices

Appendix A

MATLAB Files (Written by E.A.Clancy)

butter_hi_design.m

```
function H = butter_hi_design(f, Fc, A, Stage1, Stage2, Stage3, Stage4,
Stage5)
%
% H = butter_hi_des(f, Fc, A, Stage1[, Stage2[, Stage3[, Stage4[, Stage5]]]])
%
%   Helps design an electronic circuit to build even-order, high-pass
% Butterworth filters of orders 2, 4, 6, 8, 10.  Plots the magnitude of
% the resulting frequency response.
%   See [Thomas Kugelstadt, "Chapter 16: Active
% Filter Design Techniques," Literature Number SLOA088, Texas Instruments
% Incorporated, Post Office Box 655303, Dallas, Texas  75265, 2001.
% Excerpted from "Op Amps for Everyone," Literature Number SLOD006A,
% Texas Instruments.  Available on the Internet at: http://www.ti.com].
%
% f:      Frequency axis (Hertz) for all calculations and plotting (vector).
% A:      Overall circuit gain (>=1).  Applied in 1st stage.
% Fc:     Desired cutoff frequency in Hertz (scalor).
% StageX: Up to 5 StageX arguments (one per stage) are permitted.  For each
stage,
%         StageX is a vector of 1-4 elements, corresponding to C1, C2,
%         R1 and R2, respectively.  C1 must be supplied (in Farads) and
%         is the first vector element.  If a second argument is supplied,
%         then it is C2 (in Farads).  If C2 is not supplied, it is set
%         equal to C1.  If
%         a third argument is supplied, it is R1 (Ohms).  If not supplied,
%         it is set as required to form a Butterworth filter.  If a fourth
%         argument is supplied, it is R2 (Ohms).  If not supplied, it is
%         set as required to form a Butterworth filter.
%
% H: Resulting (complex-valued) frequency response corresponding to the
%    frequency axis f.
%
% USAGE RECOMENDATIONS:  To build a filter, initially call script with only
%   C1 and C2 specified for each stage.  Use the recommended
%   R1 and R2 values to find R1's and R2's that are manufactured.  Call the
%   script a second time with all values to see the resultant nominal
frequency
%   response.

% Table of ai values.  For even-order Butterworth, all bi values equal 1.
aiTable = [1.4142    NaN    NaN    NaN    NaN;  % For 1-stage filter.
           1.8478 0.7654    NaN    NaN    NaN;  % For 2-stage filter.
           1.9319 1.4142 0.5176    NaN    NaN;  % For 3-stage filter.
           1.9616 1.6629 1.1111 0.3902    NaN;  % For 4-stage filter.
           1.9754 1.7820 1.4142 0.9080 0.3129]; % For 5-stage filter.

% Determine the number of stages.
```

```
Nstage = nargin - 3;
if Nstage<1 | Nstage>5, error('Must have 1-5 stages.'); end
if A<1, error('"A" must be >= 1.'); end

% Extract/develop parameters for each stage and build the frequency response.

H = ones( 1, length(f) );  % Initialize frequency response to unity.
w = 2*pi*f;                % Convert to radian frequency.
MagBig = A;                % Cascade passband gain.

for S = 1:Nstage

  if S>1, A = 1; end  % Set gain to one after first stage.

  % Prepare to parse parameter vector.
  eval(['Param = Stage' int2str(S) ';']);  % Copy parameters to a scratch
vector.
  if length(Param)<1, error(['Stage ' int2str(S) ' parameter list < 1.']);
end;
  if length(Param)>4, error(['Stage ' int2str(S) ' parameter list > 4.']);
end;
  ai = aiTable(Nstage,S);

  % Parse parameter vector.
  % C1.
  C1 = Param(1);
  % C2.
  if length(Param)>1,  C2 = Param(2);  else,  C2 = C1; end
  % R1.
  if length(Param)>2,  R1 = Param(3);
  elseif A==1
    R1 = ( C1+C2 ) / ( 2*pi*Fc*ai*C1*C2 );
  else
    a = ((2*pi*Fc)^2) * C1*C2*C2*(1-A);
    b = -ai * 2*pi*Fc * C1*C2;
    c = C1 + C2;
    R1 = ( -b - sqrt(b*b - 4*a*c) ) / (2*a);
  end
  % R2.
  if length(Param)>3,  R2 = Param(4);
  else,  R2 = 1 / ( ((2*pi*Fc)^2)*R1*C1*C2 ); end

  % Print component values.
  fprintf('Stage %d: C1=%e F, C2=%e F, R1=%e Ohms, R2=%e Ohms\n', S, C1, C2,
R1, R2);

  % Update frequency response.
  a = ( R2*(C1+C2) + R1*C2*(1-A) ) ./ (R1*R2*C1*C2);
  b = 1 ./ (R1*R2*C1*C2);
  Hstage = A ./ (  1 + ( a./(j*w) ) + ( b./(-w.*w) )  );
  H = H .* Hstage;
end

% Now, plot the frequency response magnitude.
plot( f, abs(H) )
xlabel('Frequency in ')
ylabel('Frequency Response Magnitude')
```

```
L1 = find( abs(H) >= (MagBig*sqrt(2)/2) );
hold on, plot([f(L1(1)) f(L1(1))], [0 MagBig], 'm'), hold off
Thing = sprintf('Cutoff  at %0.2f Hz, Passband Gain = %0.2f', f(L1(1)),
MagBig);
text( max(f)/5, max(abs(H))/3, Thing);
title('Butterworth High-Pass Filter');

figure(gcf);

return
```

butter_lo_design.m

```
function H = butter_lo_design(f, Fc, A, Stage1, Stage2, Stage3, Stage4,
Stage5)
%
% H = b_lo_des(f, Fc, A, Stage1[, Stage2[, Stage3[, Stage4[, Stage5]]]])
%
%   Helps design an electronic circuit to build even-order, low-pass
% Butterworth filters of orders 2, 4, 6, 8, 10.  Plots the magnitude of
% the resulting frequency response.
%   See [Thomas Kugelstadt, "Chapter 16: Active
% Filter Design Techniques," Literature Number SLOA088, Texas Instruments
% Incorporated, Post Office Box 655303, Dallas, Texas  75265, 2001.
% Excerpted from "Op Amps for Everyone," Literature Number SLOD006A,
% Texas Instruments.  Available on the Internet at: http://www.ti.com].
%
% f:      Frequency axis (Hertz) for all calculations and plotting (vector).
% A:      Overall circuit gain (>=1).  Applied in 1st stage.
% Fc:     Desired cutoff frequency in Hertz (scalor).
% StageX: Up to 5 StageX arguments (one per stage) are permitted.  For each
stage,
%         StageX is a vector of 1-4 elements, corresponding to C1, C2,
%         R1 and R2, respectively.  C1 must be supplied (in Farads) and
%         is the first vector element.  If a second argument is supplied,
%         then it is C2 (in Farads).  If C2 is not supplied, it is set
%         equal to C1.  If
%         a third argument is supplied, it is R1 (Ohms).  If not supplied,
%         it is set as required to form a Butterworth filter.  If a fourth
%         argument is supplied, it is R2 (Ohms).  If not supplied, it is
%         set as required to form a Butterworth filter.
%
% H: Resulting (complex-valued) frequency response corresponding to the
%    frequency axis f.
%
% USAGE RECOMENDATIONS:  To build a filter, initially call script with only
%   C1 and C2 specified for each stage.  Use the recommended
%   R1 and R2 values to find R1's and R2's that are manufactured.  Call the
%   script a second time with all values to see the resultant nominal
frequency
%   response.

% Table of ai values.  For even-order Butterworth, all bi values equal 1.
aiTable = [1.4142    NaN    NaN    NaN    NaN;  % For 1-stage filter.
           1.8478 0.7654    NaN    NaN    NaN;  % For 2-stage filter.
```

```
           1.9319 1.4142 0.5176    NaN    NaN;  % For 3-stage filter.
           1.9616 1.6629 1.1111 0.3902    NaN;  % For 4-stage filter.
           1.9754 1.7820 1.4142 0.9080 0.3129]; % For 5-stage filter.

% Determine the number of stages.
Nstage = nargin - 3;
if Nstage<1 | Nstage>5, error('Must have 1-5 stages.'); end
if A<1, error('"A" must be >= 1.'); end

% Extract/develop parameters for each stage and build the frequency response.

H = ones( 1, length(f) );  % Initialize frequency response to unity.
w = 2*pi*f;                % Convert to radian frequency.
MagBig = A;                % Cascade passband gain.

for S = 1:Nstage

  if S>1, A = 1; end  % Set gain to one after first stage.

  % Prepare to parse parameter vector.
  eval(['Param = Stage' int2str(S) ';']);  % Copy parameters to a scratch
vector.
  if length(Param)<1, error(['Stage ' int2str(S) ' parameter list < 1.']);
end;
  if length(Param)>4, error(['Stage ' int2str(S) ' parameter list > 4.']);
end;
  ai = aiTable(Nstage,S);

  % Parse parameter vector.
  % C1.
  C1 = Param(1);
  % C2.
  if length(Param)>1,  C2 = Param(2);  else,  C2 = C1; end
  % R1.
  if length(Param)>2,  R1 = Param(3);
  else
    a = ((2*pi*Fc)^3) * C2 * ( C1 + (1-A)*C2 );
    b = -ai * C2 * ((2*pi*Fc)^2);
    c = 2*pi*Fc;
    R1 = ( -b - sqrt(b*b - 4*a*c) ) / (2*a);
  end
  % R2.
  if length(Param)>3,  R2 = Param(4);
  else,  R2 = 1 / ( ((2*pi*Fc)^2)*R1*C1*C2 ); end

  % Print component values.
  fprintf('Stage %d: C1=%e F, C2=%e F, R1=%e Ohms, R2=%e Ohms\n', S, C1, C2,
R1, R2);

  % Update frequency response.
  a = C1*(R1+R2) + (1-A)*R1*C2;
  b = R1 * R2 * C1 * C2;
  Hstage = A ./ (  1 + ( a*j*w ) - ( b*w.*w )  );
  H = H .* Hstage;
end
```

```
% Now, plot the frequency response magnitude.
plot( f, abs(H) )
xlabel('Frequency in Hertz')
ylabel('Frequency Response Magnitude')

L1 = find( abs(H) <= (MagBig*sqrt(2)/2) );
hold on, plot([f(L1(1)) f(L1(1))], [0 MagBig], 'm'), hold off
Thing = sprintf('Cutoff  at %0.2f Hz, Passband Gain = %0.2f', f(L1(1)),
MagBig);
text( max(f)/5, max(abs(H))/3, Thing);
title('Butterworth Low-Pass Filter');

figure(gcf);

return
```

Butter_hi_design2.m

```
function [H, CR] = butter_hi_design2(f, Fc, A, Stage1, Stage2, Stage3,
Stage4, Stage5)
%
% H = b_hi_des(f, Fc, A, Stage1[, Stage2[, Stage3[, Stage4[, Stage5]]]])
%
%   Helps design an electronic circuit to build even-order, high-pass
% Butterworth filters of orders 2, 4, 6, 8, 10.  Plots the magnitude of
% the resulting frequency response.
%   See [Thomas Kugelstadt, "Chapter 16: Active
% Filter Design Techniques," Literature Number SLOA088, Texas Instruments
% Incorporated, Post Office Box 655303, Dallas, Texas  75265, 2001.
% Excerpted from "Op Amps for Everyone," Literature Number SLOD006A,
% Texas Instruments.  Available on the Internet at: http://www.ti.com].
%
% f:      Frequency axis (Hertz) for all calculations and plotting (vector).
% A:      Overall circuit gain (>=1).  Applied in 1st stage.
% Fc:     Desired cutoff frequency in Hertz (scalor).
% StageX: Up to 5 StageX arguments (one per stage) are permitted.  For each
stage,
%         StageX is a vector of 1-4 elements, corresponding to C1, C2,
%         R1 and R2, respectively.  C1 must be supplied (in Farads) and
%         is the first vector element.  If a second argument is supplied,
%         then it is C2 (in Farads).  If C2 is not supplied, it is set
%         equal to C1.  If
%         a third argument is supplied, it is R1 (Ohms).  If not supplied,
%         it is set as required to form a Butterworth filter.  If a fourth
%         argument is supplied, it is R2 (Ohms).  If not supplied, it is
%         set as required to form a Butterworth filter.
%
% H: Resulting (complex-valued) frequency response corresponding to the
%    frequency axis f.
% CR: Nstage by 4 matrix. CR(1,:) holds 4 elements, the first stage values
for
%    C1, C2, R1 and R2.  Etc.
%
% USAGE RECOMENDATIONS:  To build a filter, initially call script with only
%   C1 and C2 specified for each stage.  Use the recommended
```

```
%   R1 and R2 values to find R1's and R2's that are manufactured.  Call the
%   script a second time with all values to see the resultant nominal
frequency
%   response.

% Table of ai values.  For even-order Butterworth, all bi values equal 1.
aiTable = [1.4142    NaN    NaN    NaN    NaN;  % For 1-stage filter.
           1.8478 0.7654    NaN    NaN    NaN;  % For 2-stage filter.
           1.9319 1.4142 0.5176    NaN    NaN;  % For 3-stage filter.
           1.9616 1.6629 1.1111 0.3902    NaN;  % For 4-stage filter.
           1.9754 1.7820 1.4142 0.9080 0.3129]; % For 5-stage filter.

% Determine the number of stages.
Nstage = nargin - 3;
if Nstage<1 || Nstage>5, error('Must have 1-5 stages.'); end
if A<1, error('"A" must be >= 1.'); end

% Extract/develop parameters for each stage and build the frequency response.

H = ones( 1, length(f) );  % Initialize frequency response to unity.
w = 2*pi*f;                % Convert to radian frequency.
MagBig = A;                % Cascade passband gain.
CR = zeros(Nstage,4);            % Pre-allocate.

for S = 1:Nstage

  if S>1, A = 1; end  % Set gain to one after first stage.

  % Prepare to parse parameter vector.
  eval(['Param = Stage' int2str(S) ';']);  % Copy parameters to a scratch
vector.
  if length(Param)<1, error(['Stage ' int2str(S) ' parameter list < 1.']);
end;
  if length(Param)>4, error(['Stage ' int2str(S) ' parameter list > 4.']);
end;
  ai = aiTable(Nstage,S);

  % Parse parameter vector.
  % C1.
  C1 = Param(1);
  % C2.
  if length(Param)>1,  C2 = Param(2);  else  C2 = C1; end
  % R1.
  if length(Param)>2,  R1 = Param(3);
  elseif A==1
    R1 = ( C1+C2 ) / ( 2*pi*Fc*ai*C1*C2 );
  else
    a = ((2*pi*Fc)^2) * C1*C2*C2*(1-A);
    b = -ai * 2*pi*Fc * C1*C2;
    c = C1 + C2;
    R1 = ( -b - sqrt(b*b - 4*a*c) ) / (2*a);
  end
  % R2.
  if length(Param)>3,  R2 = Param(4);
  else  R2 = 1 / ( ((2*pi*Fc)^2)*R1*C1*C2 ); end
```

```
  % Print component values.
  fprintf('Stage %d: C1=%e F, C2=%e F, R1=%e Ohms, R2=%e Ohms\n', S, C1, C2,
R1, R2);

  % Update frequency response.
  a = ( R2*(C1+C2) + R1*C2*(1-A) ) ./ (R1*R2*C1*C2);
  b = 1 ./ (R1*R2*C1*C2);
  Hstage = A ./ (  1 + ( a./(1j*w) ) + ( b./(-w.*w) )  );
  H = H .* Hstage;

  % Update output vector with C1, C2, R1, R2 values.
  CR(S,:) = [C1 C2 R1 R2];
end

% Now, plot the frequency response magnitude.
plot( f, abs(H) )
xlabel('Frequency in Hertz')
ylabel('Frequency Response Magnitude')

L1 = find( abs(H) >= (MagBig*sqrt(2)/2) );
hold on, plot([f(L1(1)) f(L1(1))], [0 MagBig], 'm'), hold off
Thing = sprintf('Cutoff  at %0.2f Hz, Passband Gain = %0.2f', f(L1(1)),
MagBig);
text( max(f)/5, max(abs(H))/3, Thing);
title('Butterworth High-Pass Filter');

figure(gcf);

return
```

marzhan01.m

```
function marzhan01

Fc = 15;  A = 5;
f = 0:0.1:100;
figure(1), clf
[H1, CR1] = butter_hi_design2(f, Fc, A, 68e-9, 68e-9, 68e-9, 68e-9);

figure(2), clf
%[H2, CR2] = butter_hi_design2(f, NaN, 10, CR1(1,:), CR1(2,:), CR1(3,:),
CR1(4,:));
[H2, CR2] = butter_hi_design2(f, NaN, 10, CR1(1,:));

figure(3), clf
plot(f, abs(H1)./abs(H2))

return
```

Appendix B

List of Components

Design #1

Name	Model/Value	Amount
Capacitor	68 nF 4.7 nF 10 nF 1 μF	2 2 1 1
Resistor	58.5 kΩ 416 kΩ 90 kΩ 10 kΩ 5.76 kΩ 29.4 kΩ 6.04 kΩ 8.45 kΩ 1 M kΩ	1 1 1 1 1 1 1 1 1
High–accuracy 18–bit multifunction DAQ for USB	National Instruments USB 6289	1
Hospital medical grade isolation transformer	Tripp Lite IS1000 120 V 1000 W 4 Outlet	1

Design #1

Name	Model/Value	Amount
Capacitor	68 nF 4.7 nF 10 nF 1 μF	2 2 1 1
Resistor	58.5 kΩ 416 kΩ 90 kΩ 10 kΩ	1 1 1 1
24–bit analog front–end IC with a ΣΔ ADC	Texas Instruments ADS1298	1
Hospital medical grade isolation transformer	Tripp Lite IS1000 120 V 1000 W 4 Outlet	1