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MMG Sensor for Muscle Activity Detection – Low Cost Design, Implementation and Experimentation

By

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Abstract

There is always a demand for cheaper, simpler and more effective human-machine interfaces. Currently, the most reliable and common muscle activity sensor is the EMG. The EMG is an expensive and complex piece of equipment which is far from ideal all round solution.

The purpose of this research is to explore various methods of muscle activity detection and using the information gathered design and implement a sensor capable of detecting muscle activation. The major focus is on mechanomyography (MMG), the measurement of mechanical response of muscle during muscle activity. It is well documented that muscles produce low frequency vibrations (5 – 100Hz) during muscle activity. A microphone is able to capture these vibrations when they reach the surface of the skin.

The prototype sensor consists of a microphone, microphone preamplifier, low pass Butterworth hardware filter, data acquisition hardware and accompanying data acquisition software.

During the experimentation phase, we explored various documented muscle events and phenomena, both general muscle events and ones unique exclusively to MMG. This includes things such as muscle fatigue, exponential relationship between force and vibration, and any other undocumented events. This experimentation also helped determine the effectiveness of the sensor.

Testing showed that periods of muscle activity were clearly visible and there was a definitely relationship between the force and vibrations, however there were shortcomings in terms of the sensor design. This included finding a better method of attaching the sensor to the surface of the skin, the dimension deformations of the muscle caused unwanted artefacts in the results and muscle fatigue was not observed. Despite its imperfections, it can be concluded that the design and implementation was a limited success.

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Martin Ma.

I, Martin Ma, hereby declare that this thesis, unless otherwise stated, is of my own work. External sources of work have been referenced and acknowledged accordingly.

Signature:

Date: / /

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List of Abbreviations

MMG – Mechanomyography

AMG – Acoustic Myography

EMG – Electromyography

PEC or PE – Piezoelectric

ACC – Accelerometer

HMI – Human-Machine Interface

DAQ – Data Acquisition

Chapter 1 Introduction

Often, when faced with a difficult problem to solve, it is not unusual to draw inspiration from something familiar or something you have a good understanding of, this type of thinking is evident in many of the robots and machines we make and use; robots that walk or lift, robots made to grasp items, neural networks are based on how the human brain works. For example, in the case of robotics, many engineers approach all-terrain robots by mimicking human walking and movement, even though the human form is far from effective, it is something familiar to them and thus easier to understand. A muscle sensor provides a means of interfacing between man and machine and because of the close relationship between humans and machines; an effective, low cost muscle sensor would be an invaluable tool to engineers and researchers.

It is often far too easy for engineers to be so focused on the on the larger picture that they lose sight of the small details, so focused on developing the next Mars rover or satellite that they forget that these things were only made possible with invention of a little thing called the transistor. It is important that we do not lose sight of the details which allow us to achieve the big picture. The motivation for prompting the development of this sensor was brought about by the inadequacies of current human-machine interfacing devices. Machine interfacing devices all seem to have compromises between effectiveness, robust, ease of use and cost. Highly sensitive sensors seem to restrict its user's movement or are very expensive while more robust and easy to use devices are far less effective. In addition to dealing with current inadequacies in this particular field, it is necessary for development to continue even if there is now immediate need to do so, it is important to maintain the some kind of forward momentum and not be stagnant. The purpose and motivation of this thesis project is to further advance the development of human-machine interfaces and muscle sensor technology.

Even though so much of what humans design is based on this close relationship between man and machine, there's seems to be a major lack of interfacing options. Currently, muscle activity detection is separated into two distinct groups: electromyography (EMG) and mechanomyography (MMG). EMGs detect the

electrical activity involved with muscle activity; it detects the changes in electrical potential of muscle nerves during contraction. EMGs are currently the most favoured and common method of muscle activity detection. MMGs on the other hand works by detecting the mechanical changes associated with muscle contractions, whether it is muscle deformation or muscle vibration. Within each group there are a variety of different methods in which to achieve activity detection; one of which is the detection muscle vibration to determine activity. Muscle vibration is a phenomenon which occurs when muscle contract, due to the nature of a muscle contraction, low infrasonic pressure waves are generated and can be picked up via a vibration transducer.

Purpose of Study

In the 1980s with the development of higher quality electronic components and technology researchers were able to begin to investigate the potentials of MMG as method muscle activity detection. Even though the core concepts of MMGs have been around since the early 1800s, it is estimated that, currently, MMG development is perhaps as much as 20 – 25 years behind that of EMGs. MMGs have a lot of potential and with more attention to its development could be a more effective alternative to EMGs. The purpose of this study is to further explore the potential of MMGs sensors.

The objectives of this thesis are:

- Explore the background and common methods of myography done in previous research.
- Design and implement a working prototype a muscle detection sensor which is both simple to implement in an application and one of practical cost.
- Design a basic user interface for simplistic visual sensor output.
- Through experimentation determine the effectiveness of the sensor and also attempt to observe events and occurrences discussed in previous research.

This thesis consists of six chapters: Chapter 1, introduction; Chapter 2, literary review, explores previous research on muscle behaviour and muscle activity detection; Chapter 3, conceptual framework, will discuss in more detail the methods

of muscle activity detection and examine the benefits and draw backs of each method; Chapter 4, methodology, discusses design and development of both the hardware and software of the sensor system; Chapter 5, testing, discusses testing methods and, analyzes and discusses the results; Chapter 6, conclusion, discusses the researches significance , the impact and explores possible improvements or future research possibilities.

Chapter 2 Literary Review

2.1 Muscle Behaviour

Muscles emit ultra low frequency vibrations while undergoing sustained contraction (Oster and Jaffe, 1980; Barry, Geiringer & Ball, 1985). Though not always audible to most people, these vibrations can usually be heard by pressing ones ear up against a muscle. This phenomenon was first noticed by William Hyde Wollaston in 1810, when Wollaston plugged his fingers in his ears he observed a low rumble, even though there was clearly no external sounds to be heard.

Wollaston observed that he was able approximate the frequency of the rumbling to the rumbling of a carriage which being driven on the cobble streets which produced a similar rumbling. Using the known size of the cobblestones and diameter of the carriage wheels he was able to calculate that the vibration's frequency was approximately 25Hz (Oster and Jaffe, 1980). Further research in later studies revealed that vibrations from muscle activity can range from a lower limit of 2Hz to 100Hz (Courteville et al., 1998).

The source of these muscle vibrations can be attributed to minute pressure waves that are generated by the muscle during contraction (Barry & Cole, 1988). Even though to a muscle contraction may seem like a smooth process, a contraction is actually achieved through a series of smaller muscle fibre convulsion, which when strung together produces what seems to be a smooth motion; it is these micro convulsions which causes the pressures waves. Barry and Cole were able to find a quantitative description of muscle pressure wave production as well as directly link the pressure amplitude to the acceleration of the muscle.

In 1985, Barry, Geringer and Ball, researched how muscle fatigue affected the muscle vibrations. The experiment was conducted on the basis that as a muscle fatigues the muscle force will decrease and due to the correlation between vibration amplitude and force (Oster and Jaffe, 1980); the vibration should also decrease, whether or not it's caused by the voluntary release or involuntary fatigue. The reason for the interest in determining whether or not fatigue can effect an AMG result is because there will be some situations where compensation for fatigue will be necessary so that you will not have a decrease in performance of your task.

Additionally, fatigue would cause degradation of the signal amplitude will also drop, compensations would need to be made so that the user can continue to use the equipment without having to over exert himself should he become fatigued. To test this 5 subjects were chosen, each person would hold a series of differently weighted bags for a varying series of time intervals. An AMG and a needle EMG were placed on the subject and results collected simultaneously. Readings were taken continuously with the subject holding the weight at 90° extensions for 20secs. The results showed that AMG amplitude had a relatively linear relationship with the force, whereas the EMG had no relationship to the force. This experiment also reinforced Oster and Jaffe's 1980 conclusion that vibration amplitude and force output of a muscle had a linear relationship.

2.2 Myography

Myography is the study, in particular the measuring of muscle activity. It is used extensively but not exclusively in medicine as a diagnostic tool to map muscle activity. Aside from medicine, myography has also been used in areas such as sports science and in a lesser degree engineering.

2.2.1 Electromyography

Electromyography is currently the standard method of muscle activity detection; it is both reliable and precise. Even though EMGs are traditionally used as a diagnostic tool and has undergone little development in diagnostics there has been increasing focus on man-machine interfacing by utilizing EMG sensors. EMGs are ideally interfaced with computers.

It is well known that the position and the location of the placement of an EMG electrode can drastically alter the characteristics of the EMG (Mercer, Bezodis, DeLion, Zachry, & Rubley, 2006). Mercer et al. (2006) investigated whether or not it electrode positioning influenced the ability to detect the difference in contraction intensity. Three pairs of electrodes were arranged in the 3 standard muscle axis' central, lateral and medially on the Biceps Brachii. The subject was asked to perform 2 different sets of contractions, isometric and isotonic, each set consisted of 3 contractions at varying efforts. Mercer et al. (2006) stated several times throughout

the article that the objective of the research was not to compare or analyse EMG intensities and locations but merely to determine subjectively whether or not the ability to detect the difference between muscle contraction intensity was influenced by the location of the sensors. Unfortunately, due to the vague nature of the objective, the conclusion was equally vague. Mercer et al., (2006) concluded that depending on the statistic of interest (EMG_{RMS} , EMG_{AVG} and mean power frequency (MPF)) you would see varying levels of influence associated with the electrode position. The only decisive conclusion was that researchers must take extreme care when selecting the location for EMG sensors and the locations must be adequately described in order that studies can be replicated (Mercer et al., 2006, p.203).

In 2009, Claudio Castellini and Patrick van der Smagt, developed an advanced and highly dexterous 4 fingered prosthetic human hand able to precisely mimic an extraordinary set of human hand actions. 2 different methods of control interfacing were considered: invasive and non-invasive. The invasive interface used sensors directly implanted to the user's nervous system to allow information to be gathered directly from the brain or nerve stems. The second method was use of surface EMG electrodes which allowed the muscle activation information to be collected without surgical implantation. Castellini and Smagt decided on the EMG method for its simplicity and relatively low cost. The prosthesis used an array of 10 pairs of EMG electrodes strapped to the user's right forearm. Electrode placement was mainly centred on muscle involved in finger control; extensor and flexor Digitorum, extensor and flexor Carpi Ulnaris, etc. Their research noted 4 major variables which required special consideration:

- Inter-subject Variability – Each user's personal physiology will alter the results of the EMG.
- Arm Posture – Each muscle may have multiple roles finger, EMG measurements will vary depending on the current position of the arm.
- Electrode Displacement – As discussed by Mercer et al. (2006), electrode position drastically influences the electrode measurements. Particularly in the case of surface EMGs, exact electrode positioning may not always be repeatable.

- Muscle Fatigue – Continual muscle activation will cause muscle fatigue which will require compensation to maintain signal quality/amplitude.

The EMG sensors coupled with neural networks allowed extraordinarily dextrous control of the prosthesis. Able to hold hammers without loss of grip and eggs without crushing or cracking them. Similar to Catellini and Smagt's application, the ultimate goal of this proposed sensor is to be used in a similar application, i.e. assistive device or control device.

2.2.2 Mechanomyography

Measurement of the transverse displacement of the skin over a contracting muscle is known as mechanomyography (Shinohara & Sogaard, 2006). Additionally, in recent times there have been new experimental methods of myography which rely on the detection of muscle deformation which fall under the scope of MMG.

Accelerometer

Currently, the majority of MMG use accelerometers, they are small and compact enough to use on the human body and are relatively cheap and easy to use. Furthermore, they are considered to be the most accurate method of vibration detection. Accelerometers are also used in similar application like earthquake and volcanic surveying. Accelerometers work by detecting skin surface displacement that is caused by the pressure waves generated by the muscle contractions.

Microphone

Microphone mechanomyography (or phono-myography or acoustic myography) work on similar principles; both are based on the detection of muscle vibration discussed in the previous section. Phonomyography, though similar is not exactly the same, functions by detecting pressure waves that are created by the surface of the skin when the muscle vibrates.

Courteville, Gharbi and Cornu (1998) developed a microphone based MMG sensor for clinic use. Their design was similar to a digital stethoscope, where the microphone was placed in a cylindrical column of air with one end sealed with a thin

elastic membrane, the membrane was then placed in direct contact with the surface of the skin. The sealed volume of air within the air column would allow better transfer of pressure from the membrane to the microphone. For the testing, the subject was placed in a relaxed position, the sensors were placed on his forearms (flexor Digitorum) and he was asked to grasp an object in his hand. The sensor worked satisfactorily and Courteville was able to determine a relationship between grasp strength and acoustic pressure and to a lesser degree, a relationship between frequency and grasp strength.

Later on, Watakabe, Mita, Akataki and Itoh (2001), further looked into AMGs and investigated the mechanical behaviour of the condenser microphones in a MMG application. This was mainly a study into what effects changing the air volume inside the cone would have on the overall performance of the MMG. Using a design similar to Courteville's, air volumes of different dimensions were used; dimensions of 5, 10, 15, 20mm diameters, 15, 20, 25mm lengths and all cylindrical volumes. Tests were repeated twice, in the first test, the sensor was attached to a sinusoidal vibration generator, which allowed a much more accurate result. On the second test, the sensor was attached to a human and measurements taken from the elbow flexion (biceps) during voluntary contractions. During the voluntary contractions test, an accelerometer was attached adjacent to the microphone as a control. Watakabe et al. (2001) determined that with a decrease of either diameter and length of air volume, there was also a decrease in the mechanical response at low frequencies. Furthermore, for correct MMG results it was necessary to have a frequency response of above 5Hz (Orizio, 1993). Consequently, it is suggested that an air volume of 10mm diameter and 15mm length is recommended.

Oster and Jaffe (1980) in order to disprove any claims that the muscle vibrations recorded during Oster and Jaffe's experiment was caused due to the friction between the microphone and the surface of the skin, Oster and Jaffe conducted a separate test which had the sensor and arm being tested submerged under water. By submerging the arm and sensor in water Oster and Jaffe created a water coupled version of an AMG. They stated in their discussion that although there was a decrease in the signal intensity, there was an overall increase in the quality of the

signal, which they attributed to the water having a higher density than that of air and that made water a better transition medium.

Trager, Mchaud, Dechamps and Hemmerling (2006) carried out a comparison of the performances of microphone and ACC MMG. The research was not find out which was better but merely to compare how each concept performed at detecting neuromuscular blockades (i.e. local anaesthetic). In the experiment, 14 subjects underwent general surgery and given a general anaesthesia; this prevents any bias or irregularities in muscle contractions. The sensor placed on adductor pollicis and artificial stimulus applied to the muscle. The muscle was stimulated until a baseline could be established after which the local anaesthetic was injected into the adductor pollicis. The onset, offset times and maximum effects. Trager et al., (2006) concluded that the onset and offset times of the PMG and MMG were in agreement, however the peak effect was severely diminished which was to be expected.

Piezoelectric

A more recent addition to MMG is the use of piezoelectric contact sensors (PEC). Piezoelectric sensors are definitely not a new technology and have been used in a variety of other applications, for example, seismic detection, strain gauges, pressure, acceleration, etc. Watakabe, Itoh, Mita and Akataki (1998), proposed a MMG sensor which utilised a PEC instead of the more traditional ACC or microphones. Watakabe et al., 1998, designed a PEC MMG sensor and setup an experiment which compared the results of the PEC to a traditional ACC result. Two sets tests were created: one with artificial excitation from a sinusoidal excitation system and another where the sensors were applied to a human for a voluntary controlled contraction test. In test 1, the PEC and ACC were tested together, first with the sensor fixed to a rigid wall unaffected by the exciter, after which the sensor was placed on top of the exciter and tested again without being fixed externally. A sinusoidal excitation was applied ranging from 5Hz to 300Hz with a constant magnitude of 0.01G. The signals are amplified by an AC amplifier and filtered through a 1Hz to 1 kHz band pass filter, careful consideration was taken to make sure that the entirety of the transducer bandwidth included in the filter with enough lead to show any anomalies the transducer might have.

Bu, Tsukamoto, Ueno, Shima and Tsuji (2008), proposed a different approach where a thin strip of piezoelectric material was secured to the surface of the skin with adhesive. As the muscle contracts, the radius of the muscle belly increases and stretch the skin and in turn the piezoelectric film, using the stretch of the film sensor, muscle activity can be determined. Bu et al. produced their own custom piezoelectric material with Aluminium Nitride (AlN). The AlN sensor film consisted of an outer electrode layer, AlN layer, Polyimide layer, inner electrode, polyimide layer, AlN, and another outer layer, similar to a basic electrolytic capacitor design and the total thickness was less than 40µm. Since the output of the film sensor is not a typical voltage output, a charge amplifier was added to convert the charge output into a more manageable voltage signal. Signals were put through a probability classifier in order to determine which motion was likely the result of a particular set of contraction. Testing revealed that although it was quite effective at determining the moment of muscle activity, the sensor was very poor at taking static measurements; when a static force is applied to the sensor there is a rapid decay in signal amplitude that was not due to muscle fatigue; for a more functional signal output, the output was integrated. Unfortunately, this sensor system would not be able effective in real time as the output requires integration the signal can be used in a practical manner.

Ultrasonic

Tanaka, Hori and Yamaguchi (2003) proposed a muscle sensor using ultrasonic sensor disks. The sensor is based on the changing shape, stiffness, density and elastic properties associated with muscle activity. Even though the sensor is called an ultrasonic sensor, it is in reality a system of sensors; whereby multiple sensors are used in unison to provide the necessary information to determine muscle activity. Tanaka et al., (2003) utilises 3 different transducers: an ultrasonic sensor, strain gauges, which assists with determining elasticity, stiffness and density, a fibre optic web was implemented which would capture the motion and shape of the muscles. The fibre optic sensors were weaved into a mesh and the ultrasonic sensors were placed in matrix. Two prototypes were made; prototype 2 was a further developed version of prototype 1 with minor improvements in its hardware such as replacing the amplifiers with low noise amplifiers and changing the sampling frequency. An experiment was conducted to compare the performance of the two ultrasonic

sensors against each other; prototype 2 obviously performed far better than the initial prototype. No comparative testing was done against other forms of MMG sensors during the experimentation phase; this meant that the relative effectiveness is known.

In 2005, Tsutsui et al. further developed an ultrasonic sensor component calling it the duplex ultrasonic muscle activity sensor (UMS). Previous to the duplex sensor, the same team had worked on the simplex UMS. The simplex system was a simple single emitter and single receiver configuration, where one emitter had a corresponding receiver. In the duplex version a system of 2 emitters and 2 receivers were set up so that signals from both emitters would be accepted by each receiver. The idea being that by using a system of interconnected sensors this will allow you to use a differential type setup to lower the noise of the system and improve overall accuracy. To prove this, the UMS was applied to a thigh and the test subject was asked to pull against a force to set arm angles. The results showed that there are definite improvements when using the duplex system, particularly at higher angles where more muscle activity is required. At 30° extension, there was a 0.01 improvement on the correlation coefficient and at 90° a 0.06 improvement. This meant that for the simplex system at 90° extension there was far more deviation in results than there was for the duplex method.

Chapter 3 Conceptual Framework

3.1 Bio-Mechanics

Biomechanics is the application of mechanical concepts and principles to living organisms. It explores and analyzes the how the various aspects of biomechanics interact with each other. These aspects include fields such as thermodynamics, mechanical engineering, bioengineering, fluid dynamics, neurology, etc.

3.1.1 Muscles

There are three main types of muscles that can be found in the human body, skeletal, smooth and cardiac.

Cardiac muscles are the muscles found only in the heart to assist in cardiac function. These muscles are incredibly resistant to fatigue due to it having an abundant supply of fresh blood, rich with myoglobin and mitochondria, which is required for healthy muscle activity. Cardiac muscles are controlled by the body's autonomous nervous system, meaning cardiac muscle contractions are completely involuntary.

Smooth muscles are involuntary muscles found in areas such as the stomach, oesophagus and the walls of blood vessels. They assist with things such as pumping blood through the body, swallowing and digestion. Again smooth muscles are governed by the autonomous nervous system and contractions are involuntary.

Lastly you have the skeletal muscles, which facilitate the movement of the skeletal system. These muscles are controlled by brain via the voluntary nervous system. They have the ability to both contract and stretch while still being able to return to its original shape and unlike cardiac or smooth muscles. Skeletal muscles are further separated into three different fibre types, type 1, type 2a and type 2b.

Type 1 fibres are your slow twitch fibres they are highly resistant to fatigue when subjected to low level contractions, which produces a much more steady level of power release from the muscles. Type 1 fibres have a very red colouration; this is due to the high levels of myoglobin and mitochondria that is needed to produce adenosine triphosphate or ATP, which is essentially the muscle's fuel source. Slow twitch muscles produce ATP at a much faster rate however is not able to break it

down as fast thus has a lower power output than fast twitch muscles. The muscle's endurance is based predominantly on the amount type 1 fibres in his skeletal muscles.

Type 2b fibres are known as fast twitch muscles. They are often a white colour as they have relatively low levels of myoglobin and mitochondria, they are able to convert those components very quickly to produce very quick and powerful bursts of energy, however this rapid conversion has the drawback of fatiguing the muscles very quickly. Fast twitch muscles are what gives a person short term strength and speed.

Type 2a are a combination of type 1 and type 2 fibres and has a good balance between both fibre 1 and fibre 2b's attributes. The amount of fibre 2a can be increased by resistance training, which develops the type 2b muscle's aerobic metabolism cycle increasing its resistance to fatigue.

Muscle activity throughout all 3 types can be detected via myographic detection systems, though depending on the situation, focus on a particular type maybe more important.

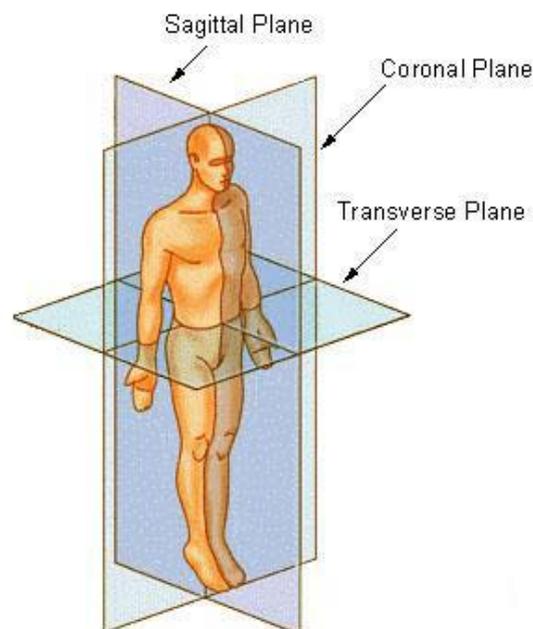


Figure 3.1.1 – Planes of the Body

Muscles are further grouped into groups determined by the actions they perform.

- Flexors/extensors – decreases and increases the angle of joints.
- Medial/lateral rotation – rotation of a limb inward or outward.
- Adduction/abduction – brings limbs closer or further away from the bodies mid line along the coronal plane. Similar to flexion and extension motions but the flexion and extension are on the sagittal plane.
- Elevation/depression – elevation or depression body part, such as shrugging.
- Protraction/retraction – a specialized upper limb motion of parts of the body forward and back anteriorly.
- Supinator/pronator – a motion unique to the limbs which allows for rotation of the forearms and lower leg.

3.1.2 Muscle Fatigue

Muscle fatigue is a very important muscle characteristic to understand when developing any muscle sensor. Muscle fatigue is the loss of muscle strength after extensive and/or continuous muscle activity.

Fatigue will influence the output signal of the sensor and the level of influence will compound the longer and more extensively you use that muscle. In the case of MMGs, Mihai (2003), Oster and Jaffe (1980), Barry et al. (1985), all have documented the lessening of signal amplitudes in relation to increasing muscle fatigue.

A major advantage MMGs have over EMGs are that MMGs more effectively displays the effects of muscle fatigue, i.e. there is a clear degradation in signal amplitude with the increase of muscle fatigue. Whereas in the case of EMGs there can often be an inverse relationship (Barry, Geiringer & Ball, 1985). MMGs which detect muscle deformation are also incapable of detection muscle fatigue. Since muscle deformation is an “after the fact” effect, the muscle deformation is a representation of the combination of force and fatigue.

3.1.3 Muscle Vibration

Muscles are made up of smaller muscle fibres, to facilitate muscle contraction each of those fibres undergo involuntary micro-convulsions the overall effect is that the muscle contracts. The interesting aspect of this process is that these micro-convulsions are believed to cause pressure waves throughout the muscle and eventually appear on the surface of the skin.

Even though many researchers recognize the micro convulsions as the source of the vibrations, there have been some debates as to other origins of muscle vibrations, the majority of which have been ruled out.

The 2 major competing theories are that muscle vibrations are caused by the muscles vascular system or the friction between the muscle lining and muscle fibres. As muscles contract muscle density increases and blood vessel are constricted, allowing a pulse to be easily transmitted throughout the muscle, it would be easy to make the connection between muscle vibration and a pulse from blood vessels. This theory was disproven when experiments were performed while the blood flow was restricted, showing that muscle vibrations still occurred without vascular interference. The second theory, involved the vibration caused by the friction between the muscle fibres and fascia (muscle bundle lining) as muscle expanded, contracted and deformed. This theory was disproven since there was no notable MMG activity recorded during passive muscle actions. Other theories include friction between the sensor and skin, bone oscillations and nerve conduction; all of which were disproven simply by isolating the muscle (Gordon & Holbourn, 1948).

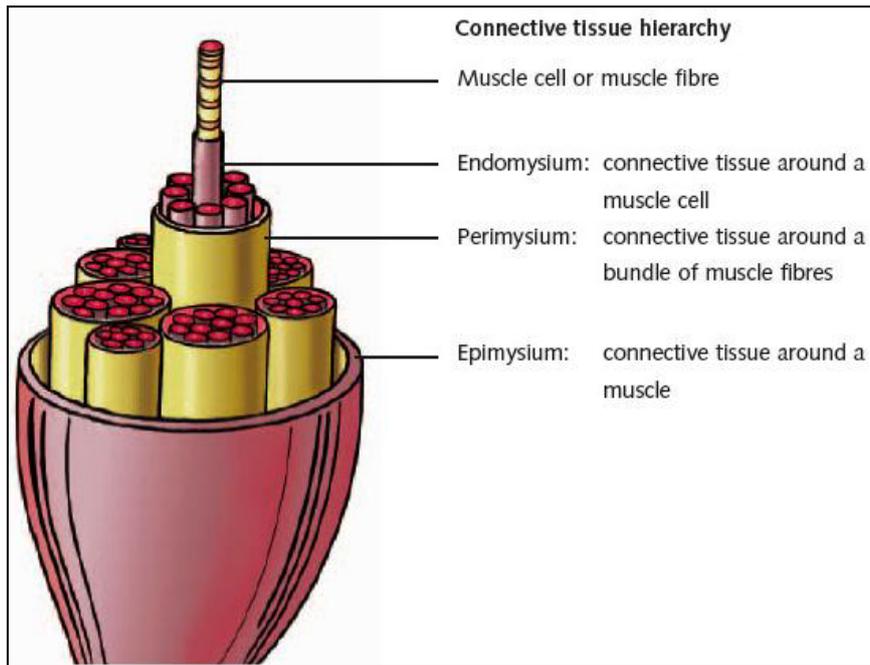


Figure 3.1.2 – Muscle Bundle

Vibration frequency and velocities are governed by the density of the material it is travelling through; naturally any density change will result in a frequency change in the vibration. There are several factors which can affect the density of muscles: temperature, muscle mass, intra muscular pressure (the pressure applied by one muscle to another during periods of activity i.e. contraction or expansion), and viscosity of cellular fluids in the muscle.

Mass, pressure and viscosity will vary from person to person, for this reason; the muscle vibration frequencies will be slightly different for each person. For this reason the filter frequency responses may need to be adjusted to more precisely target the unique vibration frequencies of the user.

Muscles, along with most of the human body, is made up of a large percentage of water, this allows vibrations to travel easily through them. Vibrations from muscle contractions can propagate throughout the entirety of the muscle and can be detected by a wide range of vibration detection devices, such as accelerometers and microphones. It is widely acknowledged that the typical frequency range of muscle vibrations lay between 5 – 100Hz, though the bandwidth can vary from person to person (Oster & Jaffe, 1980).

3.2 Electromyography

Electromyography (or EMG) is a technique for the observation of electrical activity in muscles. EMG detects changes in electrical potential between muscle fibres and cells during activity.

Currently, EMGs are the gold standard of muscle activity detection; it is recognized both in engineering and medical fields as the most reliable and precise method to observe muscle activity. Most commonly used in the medical field to diagnose muscle related disorders and in sports science to provide biofeedback. EMGs are predominantly used as an observational tool as its sensitivity to external interference means it is rarely used outside of a controlled environment. In engineering EMGs are used generally for comparison tool as a control.

There are two methods of EMG, surface and intramuscular (or needle) EMG; both work in a slightly different way and are used in slightly different situations.

Surface EMG

Surface EMGs work by placing sensors on the surface of the skin above the muscle of interest. The sensor disks are able to detect the electromagnetic interference (EMI) created when the muscle nerves fire. By observing both its strength and pattern, it is possible to determine the muscle activity occurring. SEMGs measure muscle activity throughout the entirety of the muscle.

SEMGs are relatively easy to use and operate, no real training is required and the procedure itself is safe and easy; because of this the main users of SEMGs are typically researchers, clinicians and people involved in sports science. SEMGs are often used in biofeedback applications, e.g. monitoring sports performance, diagnosing basic chiropractic conditions, etc.

Intramuscular EMG (Needle EMG)

Intramuscular or needle EMG involves the insertion of two probes, in the shape of needles, directly into the muscle tissue. By measuring electrical potential of the two probes you can determine not only the activity of the muscle but go a step further

and gather information on particular muscle fibres and other information such as nerve conduction.

Typically used in medical situations, like biofeedback, medical diagnosis and advanced muscular research. The precision of needle EMGs makes it very desirable for research, particularly in assistive and interface devices where progress has been hindered by lack of high precision sensors. The obvious drawback of using needle EMGs is that its operation does require a trained operator typically a medical professional. The needles do need to be inserted in the correct places and failure to do so can give wrong measurements and the procedure does involve insertion of needles which warrants the risks of muscle and/or nerve damage, and/or infection. Furthermore, having needles inserted into a muscle, often an array of needles does restrict his/her movement substantially, this limits the range of applications needle EMG are suited for.

The potential of a muscle in a normal state is typical -90mV and the potential difference change can be as high as from 20 to 30 mV (Nigg & Herzog, 2007).

For all its advantages, EMGs are not ideal for all situations. As it's been stated before EMGs greatest disadvantage is its sensitivity to external noise and interference, it limits both its operating environment and the range applications. For example, in tests where artificial muscle stimulation is applied to a muscle via direct electrical impulses to the nerve, EMG measurements are corrupted with large artefacts caused by the stimulus charge (Murphy et al., 2008).

Additionally, an EMG unit is far from simple. Fundamentally, an EMG and MMG sensor are similar, consisting of amplifiers, filters and other signal conditioners. However, because of the greater level of sensitivity it requires to detect nerve activity, the components and techniques are considerably different. EMG sensors need to be made from low noise and very stable components. Furthermore, special considerations need to be made regarding adequate shielding, component placement and board zoning etc; using standard components, techniques and practices is insufficient to produce the required performance. Because of the mandatory requirements for building an EMG, it is not viable to build one, and unfortunately, professionally built EMG are typically made for medical use and are

often very expensive and bulky. EMG sensors for engineering purposes are still in a relatively early stage of development, tending to be fragile, expensive and bulky.

Advantages

- Precise and accurate.
- Can target specific muscle fibres; even in situations where muscles overlap.
- A lot of research regarding EMGs.

Disadvantages

- Requires controlled environment to operate effectively.
- Needle EMGs are invasive.
- Signal processing and analysis is complex.
- Electronic requirements are very high.
- Requires special training to analyse information correctly.
- EMG unit is relatively large and bulky.
- Expensive.

3.3 Mechanomyography

Mechanomyography is the process of capturing mechanical changes of muscle characteristics and properties to determine muscle activity. Mechanomyography is relatively new with extensive research only starting 1980s, though the fundamental ideas and concepts have been around since the 1800s. In comparison, EMG development is estimated to be almost 20 to 25 years ahead of that of MMGs.

There are currently 2 main properties MMGs are based on, muscle vibration and muscle deformation, however recently there has been considerable interest in muscle properties such as density and elasticity. MMG encompasses both muscle deformation and muscle vibration detection techniques but MMG is used typically for muscle vibration only, also phonomyography (PMG), acoustic myography (AMG) and vibro-myography.

Advantages

- Shows the effects of fatigue clearly. Depending on the situation muscle fatigue can be either good or bad. In some situations, it may be useful to have access to information such as fatigue; it could be an additional variable to adjust the performance of an actuator to suit your current physical status.
- Compact and robust.
- Effective MMGs can be made with relatively basic and consumer level components.
- Positioning is not as crucial as EMG since muscle vibration propagates throughout the muscle, so sensors placed anywhere along the length of the muscle will provide similar measurements. Deformation on the other hand will need to be recalibrated for each position.

Disadvantages

- Fatigue affects the measurements. Again, fatigue can either be good or bad; it may simply be an undesired value that needs additional signal processing to compensate for.
- Relatively new with far less research done. MMG sensors technologies are relatively new and majority of resources are rarely purpose built.
- Only surface muscles can be accurately measured, muscles hidden underneath other muscles cannot be targeted.
- Sensors must be attached to the muscle. MMG sensors measure the physical response of the muscle to determine activation, unlike and EMG which measures the electrical response. In a situation involving an amputee or deformation, EMG sensors can be placed in still existing sections of nerve where as the MMG sensors cannot be used since the muscle no longer exists.

3.3.1 Vibration Detection

In the 1980's a lot research was done based on muscle vibrations, this included origins, properties, muscle activity of other species, etc. In the majority of research accelerometers and to a less extent microphones were used.

Vibrations are relatively easy to detect; simply plugging your ears with your fingers will allow you to hear the low rumble of the muscle activity in your fingers. Accelerometers are the standard transducer used for vibration detection in most fields and application. Microphones, however, are application dependant when it comes to vibration detection. Microphones are pressure transducers and as such are only usable in a vibration application is the vibrations cause pressure variations.

Accelerometers

Accelerometers are widely used as a method of vibration detection, from stress testing to seismography. There are several different types of accelerometers, piezoelectric, piezoresistive, capacitive, Hall Effect, etc. Most commonly, accelerometers merely contain a material which is force sensitive, meaning that it is sensitive to any forces which are applied to it, for example, piezoelectric materials like quartz and piezoresistive materials like silicon. Piezoelectric materials emit a low electric charge when a force or pressure is applied to it, a phenomenon found predominantly in crystals and ceramics. Piezoresistive materials have a similar property however rather than changing its electric potential, piezoresistive materials changes in electrical resistance.

In an application such as MMG, the lower cut off frequency is most important; accelerometers have the advantage of being able to measure down to a lower range of zero Hz. As opposed to other vibration transducers like microphones rarely have cut off frequencies lower than 5Hz. Furthermore, accelerometers sensitivity is usually higher than microphones; accelerometers are able to be directly mounted to the measurement surface, whereas a microphone often requires an offset distance to provide the optimal acquisition.

The two main points of interest is that accelerometers measure in Gs (gravities) and orientation affects the output. Accelerometers functions by measuring the physical acceleration on an object by using freefall (Earth's gravity) as a point of reference and as such calculations are needed to adjust for the different ways gravity affects the ACC and any other forces experienced by the ACC that is outside of the system; particularly if the ACC is mounted to a free moving object.

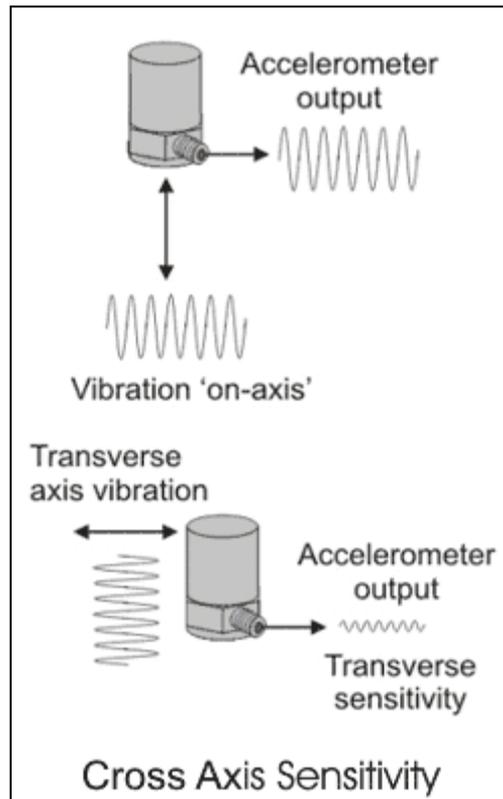


Figure 3.3.1 – Accelerometer Orientation

It is common for accelerometers to have a multi axis capability which means that they are able to simultaneously measure acceleration on two or three axis. As such it is important to note the orientation of the ACC since the effect of tilt will affect the accuracy of the measurements.

A major factor in the decision of what technology to utilise for this sensor was how sensitive accelerometers are, in most situations high sensitivity is a benefit however, in this case it was a hindrance; the ACC would simply acquire too much information, the majority of which would need to be removed or filtered out.

Microphones

Microphones are a relatively old technology, with the first microphone having been invented in 1876 by Emile Berliner; however the development of AMGs only started around the 1980s. There is a large variety of: condensers, carbon, carbon, etc. the majority of which are relatively old concepts but have only recently advanced to the stage where they can be used in a MMG application.

Condenser microphones are one of the older and more evolved microphone concepts. The idea is based on variable capacitors, where there are two plates, one fixed and stationary and the other (diaphragm) moves as sound pressure waves push against it. As the diaphragm moves the distance between the two plates changes and in turn changes the capacitance of the plates.

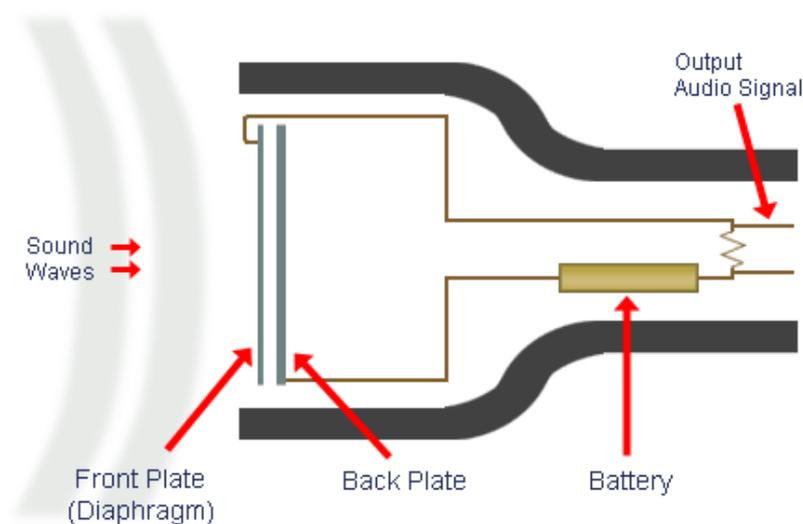


Figure 3.3.2 – Condenser Microphone

Since the condenser microphone is essentially a variable capacitor, a voltage must be applied for the capsule to function. As shown in Figure 3.3.2, a battery is used; however a stable external DC source can also be used, called a phantom power source. A phantom power supply provides a DC voltage (normally 48V) to both outputs of the microphone. By balancing the outputs you are able to negate or lower any noise that may be introduced into the system by the power supply. 48V is applied to the +ve and –ve outputs; potential difference from +ve to GND and –ve to GND will be the same, 48V. However, the potential difference between +ve and –ve will be 0V and since the phantom supply to each output is the same, any AC variations in the phantom supply will simply cancel out.

$$C = \frac{Q}{V} \tag{3.3.1}$$

$$C = \frac{\epsilon A}{d} \quad (3.3.2)$$

The equations above show the basic parallel plate capacitor equation that condenser microphones are based on. Where C is capacitance in Farads, Q is charge in Coulombs, V is voltage in Volts, ϵ is the permittivity, A is surface area of plates and d is the distance between plates.

$$V = \frac{Q}{\epsilon A} d \quad (3.3.3)$$

A simple rearrangement gives equation

(3.3.3, which shows that the voltage output of the microphone is proportional to the distance of the diaphragm; given the charge (Q) across the plates, the surface area of the plates (A) and permittivity (ϵ) remain constant.

Traditionally, the plates are biased externally with a standard 200V or higher DC supply; biasing provides the positive and negative plate charges. In the past there have been issues where the charge across the plates vary, which causes output noise however advancements in microphone technology has drastically lowered the charge variation. Recently, new microphones called electret condenser microphones (ECM), rather than having an external power source provide the charge, one of the plates (normally the diaphragm) is manufactured with a permanent charge.

The polar (directionality) pattern of a microphone shows the pressure sensitivity of the microphone at different angles. This pattern is different depending on the type of microphone, additionally specific patterns can be custom made and matched for a specific task. In applications such as music or general acoustics, a more omni-directional microphone maybe more appropriate. On the other hand applications such as AMG may be more suited for a more directional microphone.

Similar to accelerometers, microphones detect the mechanical changes which occur when muscles contract. However, unlike the ACC, rather than picking up the raw movement (i.e. tilt, roll or translation) of the surface of the skin, microphones pick up

the air pressure waves created by skin vibration; it is similar to having an air coupled ACC with only one axis of measurement. The main advantage of using a microphone is that it eliminates the need for filtering of undesired movement.

3.3.2 Muscle Deformation

Muscle deformation MMG's are a concept based on measurement of the dimensional changes of a particular muscle during activity. Detection method can range from using piezoelectric films to electromagnetic fields, from laser distal displacement detection to using standard strain gauges.

Currently, all the equipment used for this type of muscle detection is custom designed and are not in widespread use. Unlike vibration detection using microphones and accelerometers, muscle deformation detection is not a medically recognized method of detecting muscle activity detection. Additionally, muscle deformation lacks the predictive abilities vibration detection has; reason being that the fundamental principle of the muscle deformation sensor is to detect the result of muscle activity. Even though it may seem like an insignificant amount of time between thought, muscle activation and the resulting action, the smaller the interval between these steps the more fluid the motion can be.

Piezoelectric Film

Piezoelectric film is a very new idea for muscle activity detection. The main concept is to place pieces of PE film laterally or radially on a muscle, any muscle deformation, either extension or expansion, would cause tension or compression on the film and thus cause a change in the potential difference of the film.

PE films are similar to capacitors; they are made of piezoelectric materials that have been rolled out flat. There are several different types of PE film, but usually consist of five or so layers of various materials laminated together; usually consisting of electrode layers, PE material layers, there can be a several different materials in a single film, and insulator layers. Most PE films will consist of these 3 components; however, there can be multiple layers of each depending on the desired properties.

By deforming the film, it is able to alter the charge difference of each PE layer and in turn change the overall capacitance of the film.

The piezoelectric effect is a unique phenomenon found in materials with crystalline structures, though not exclusively in crystals, materials with crystalline structures such as ceramics and polymers can also exhibit similar qualities.

The advantage of using a PE film for MMG applications is that PE films are very light and compact, even though the film would need to be secured to the skin surface via adhesive, it would be no more restrictive than having a piece of tape stuck on you. In cases where the film is applied radially, i.e. round the a muscle as is the case with Bu et al. 2008, the film can be embedded in a elastic material and worn similar to an arm band.

The main disadvantage of using a PE film is that the methods of mounting are far from ideal. Even though the film can be firmly secured on to the skins surface using adhesives, slippages are always a factor. Additionally, by securing it on to the surface of the skin, the skin acts as a coupling medium between the film and muscle deformation and the skins inherent elasticity will introduce some inaccuracies. Radially mounted PE films embedded in materials have also the elasticity of the material it is embedded in to contend with.

Ultrasonics

Ultrasonic sensors are similar to PE films however they rely less on the dimensional muscle deformation but rather more on the internal muscle properties, such as density and elasticity. By looking at the muscle density and elastic properties, in addition to a few other dimensional facts, it is possible to determine the dimensional deformation of the muscle or at the very least forces being applied to it. The ultrasonic sensors do nothing other than gather the time delay between the time when a signal is sent and when it is received.

$$v = f\lambda \tag{3.3.4}$$

$$c = \sqrt{\frac{C}{\rho}} \quad (3.3.5)$$

The equation (3.3.5 shows the equation used to calculate the speed of sound; c is the velocity of sound (ms^{-1}), ρ is the density of the material (kgm^{-3}) and C is the modulus of elasticity (Pa). The modulus of elasticity is a general constant, if a more specific medium is given, the constant would be changed to K (bulk modulus) if a gas or liquid and G (shear modulus) if it is a solid. Tanaka et al. (2003) considered the muscle to demonstrate characteristics of a liquid and so calculation were made with the bulk modulus (K). From equation (3.3.5, muscle activity is expected to change C or ρ ; therefore any change in velocity that the ultrasonic sensors pick up would signify muscle activity.

The main drawback for a system such as this is that it can have a very slow reaction time caused by the amount of processing and calculations required. In a situation such as Tanaka et al. (2003, 2005) multiple samples and averaging was used to provide a more precise result, this significantly lowers the reaction time and can require a considerable amount of processing power, definitely requiring a microprocessor. Furthermore, ultrasonic sensors which pass through multiple different mediums tend to perform poorly. The greatest advantage is that the system would be very robust, both as acoustically and physically, external noise would only affect the sensor if it were at a similar frequency range. Additionally, ultrasonic sensors, particularly these days, are extremely cheap and readily accessible.

3.4 Infrasound

The sound spectrum is divided into 3 main sections, ultrasound, acoustic and infrasound. Acoustic is the standard human hearing range, between 20Hz – 20 kHz. Ultrasound is above the range of human hearing from 20 kHz upward, ultrasound is

used extensively in engineering and medical applications. High frequency sound waves have good reflectivity and directionality; it is often used for imaging applications. Infrasound is at the opposite end of the spectrum to ultrasound, with a range of ~0Hz – 20Hz, with enough sound pressure infrasound waves occasionally infrasound waves are able to be picked up by the ear as a low rumble or felt throughout various parts of the body. Low frequency waves do not require a lot of power to have very wide propagation distance, which means that muscle vibrations from simultaneous multiple muscle activations may interfere with each other.

3.5 Conclusion

The major decision had to be made on what system of muscle activity detection was the most effective and practical; several factors were considered: cost, complexity and robustness/resistance to noise/interference. EMG detection was used as the comparative method, they are very accurate and effective, and factors such as cost and complexity were considerable drawbacks.

The first factor was one of the major objectives of the study, which was to develop a basic and low cost method of muscle activity detection. These two conditions dismissed several of the more advanced methods of detection such as laser distal detection, ultrasonic methods and a few accelerometer variations of the MMG. Ultrasonic's require relative large amounts of signal conditioning and unlike signals from a microphone sensor must be interpreted; i.e. the output signal received will only tell you the time interval between signal sent and signal received, muscle activity must be derived from calculations to determine signal frequencies, signal speeds and muscle density.

Cost was the second major consideration; the obvious reasoning being the more money invested in the device the higher quality of the sensor, and in turn will provide greater performance. A lot of the systems especially the more complex and sophisticated methods such as laser distal, tend to have a higher cost, with the obvious reason being the equipment and materials used are expensive not to mention mere the quantity of that equipment and material. Microphones, accelerometers and ultrasonic's were obvious choice since equipment required for all three are relatively cheap and readily available. Of course there are exceptions,

with microphones and accelerometers, professional high quality, high precision transducers can have very large price tags.

Other less important factors include, ease of use, robustness, previous research, availability of equipment and materials, and possible applications. Ease of use looked into the various circuits and systems that would be required to make a functioning muscle sensor with the chosen transducer; determining the ease of use of the transducer itself how easy it would be to implement. How robust the completed sensor would be was important. How well the sensor would cope with external noise. How well the sensor would cope with harsh treatment and abuse. Even though it's a sensor that can be used in a laboratory environment, it should also be able to achieve a satisfactory level of performance if it was implemented in an application outside of a lab. Equipment and materials requirements needed to implement a microphone/accelerometer sensor are relatively basic. Ultrasonic and piezoelectric systems often will require custom made components such as the piezoelectric film and ultrasonic sensor harness. Consequently, microphone and accelerometer MMG is by far the most extensively and thoroughly researched, with serious research starting in the 1990's. Other methods of muscle activity detection tending to be more novel concepts, with no follow up research to build upon it.

The decision was made to use a microphone as the transducer. Microphones are excellent for the requirements of this sensor; they are cheap, robust, easy to implement, have already relatively large quantity of research done with them and are relatively easy to get acquire. The microphone will need an accompanying preamplifier, power supply, low pass or band pass filter and a sealed air column will also be used to provide better interfacing between skin and sensor.

Chapter 4 Methodology

4.1 Microphone

Selecting the correct microphone for this application is essential. The microphone is the conversion point, where mechanical movement is converted into an electrical signal; the more accurate the conversion the more effective the entire system will be. Since the microphone is at the lowest level of the system, the signal output from the microphone will need to be the highest quality as it will set the base line quality of the signal; the signal quality will only degrade as more processes are added to it.

These days, microphones come in a large variety of specifications, varying from the sensitivity (SPL) to the frequency response, from capsule size to operating voltages. For this application, a high sensitivity, ultra low frequency microphone was needed. A low frequency requirement eliminates most consumer level microphones and industrial acoustic microphones. Manufacturers, typically, will not produce microphones lower than 20Hz since the acoustic frequency range is only 20Hz to 20 kHz. Unfortunately, industrial low frequency microphones are considerable more expensive than the consumer level microphones. Condenser microphones were chosen for this application as it provides good low frequency response and least expensive. Other cheaper microphones were considered however most had a lower frequency cut-off higher than 100Hz; usually infrasonic sensors are used for seismic measurements and detections and are therefore often custom-made for the application, there is rarely an off the shelf solution.

Professional Measurement Microphone (Microtech Gefell MK250)

The Microtech Gefell MK250 was considered initially. The MK250 is an electret condenser microphone meaning no polarization voltage is required. It has a free field frequency of 3.5Hz – 20 kHz (± 2 dB), sensitivity of 50mV/Pa and inherent noise threshold of 15dBA. Since the MK250 is a free field microphone utilized in a pressure field situation the manufacturer has stated that the lower frequency cutoff will be closer to 5Hz. Measurement microphone capsules has a couple major drawbacks, first being the very high price. Measurement capsules commonly cost approximately \$2000 to \$3000 excluding preamp which can also cost along the lines of \$1500 to \$2000; for professional/industry microphone capsules and preamps are paired

together to provide the best results. MG suggests using the MV203/MV204 preamplifier in conjunction with the MK250 capsule.

The MV203/MV204 preamplifier has a frequency response of 1Hz – 1MHz, input impedance of 20GΩ, output impedance of 80Ω and a gain of 1. This preamplifier serves only as an impedance bridge; to decrease the overall output impedance of the microphone.

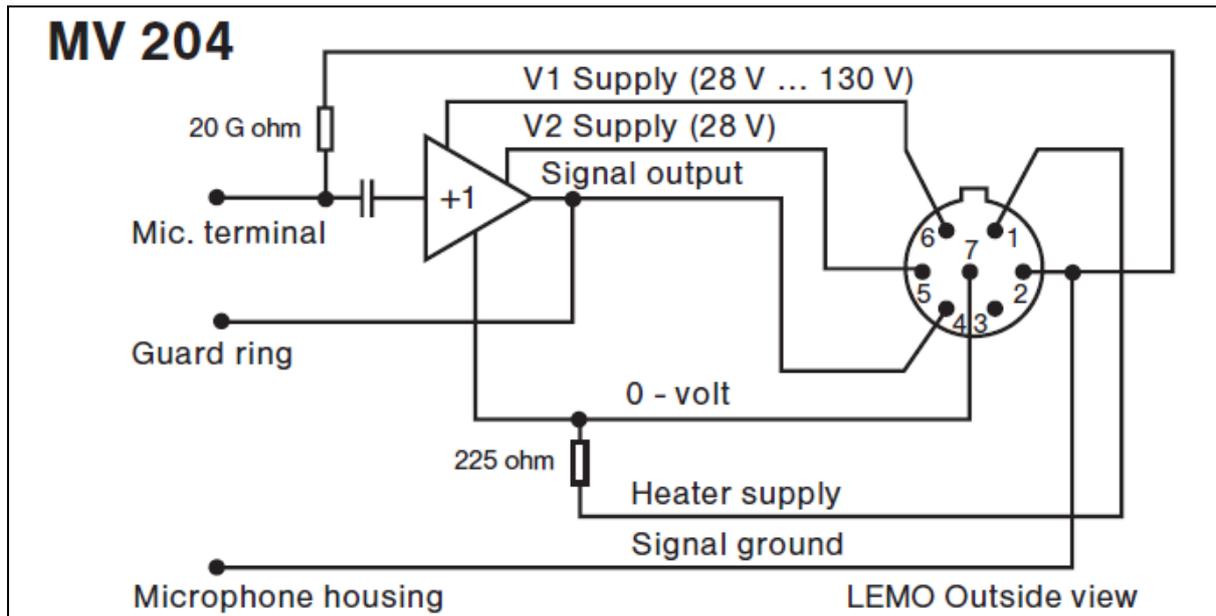


Figure 4.1.1 – MV204 Preamplifier Configuration

Since the MV204 preamplifier was outside of the budget, a preamplifier would need to be built to serve the same function. The drawback of building your own preamplifier is that it is difficult to achieve the same level of performance as that of a professionally built one because professional microphones have their preamplifiers specifically designed and matched for each other. There are often issues such as input/output impedances, technologies and types of components to use; for example, old microphones may require transformer type coupling or use vacuum tubes instead of transistors or BJTs instead of FETs.

Consumer Level Microphone (Panasonic WM-61)

A Panasonic WM61 electret microphone was selected as the consumer level microphone. The WM61 has a frequency range of 20Hz – 20 kHz (tested), sensitivity of 1V/Pa; the manufacturer did not specify inherent noise. The reason the WM61 was selected for this application even though its frequency range only goes down to 20Hz, the lower frequency limit provided by the manufacturer is only the lowest tested frequency, because most ECM applications rarely have lower frequency lower than 50Hz. It has been documented that with the correct preamplifier configuration the frequency range of the microphone could go much lower than 20Hz. The major advantage for using the WM61 is the cost; around \$2 per capsule. Additionally, they are fairly common, being sold by a large group of electronic component stores.

Unlike the MK250 capsule, the WM61 has an accompanying FET driver inbuilt in the capsule which allows plug and play situation with standard ECM preamplifiers. The WM61's preamplifier requirements are far more common, a common instrumentation or low noise preamplifier was all that is needed.

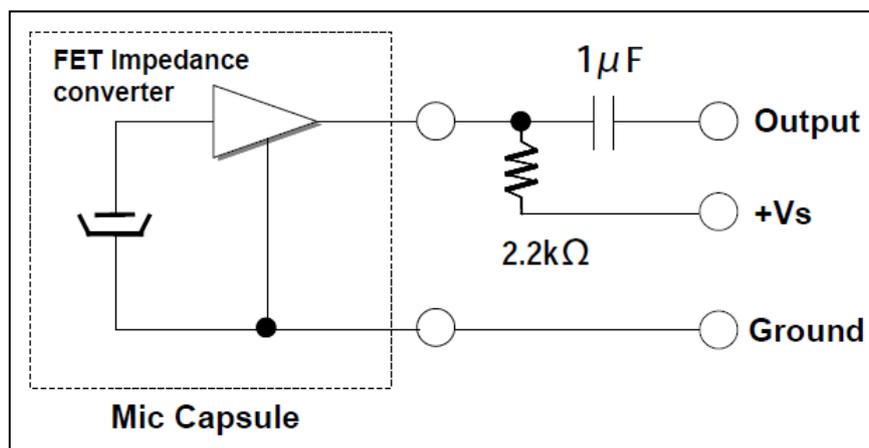


Figure 4.1.2 – Impedance Converter for the Panasonic WM61

A FET impedance converter serves a similar function to the MV204 preamplifier, it is simply a device which converts the mid or high level impedance of the actual microphone element to low impedance. The reason for the conversion is to allow the optimal power transfer between the capsule and preamplifier especially in a case where there is a great distance between the capsule and preamp; power lost by the

line can be significant especially considering the small output signal size coming from the transducer.

4.2 Microphone Holder Assembly

Research done by Courteville et al. (1998) showed that a sealed column of air allowed for a better mechanical response as it concentrates or confines the pressure waves which more effectively transfers the force generated from the pressure waves to the diaphragm of the microphone.

Rather than use the basic cylinder design, a conical design was decided on because of the low sensitivity WM-61A, steps maybe needed to boost the pressure. Using a cone design will concentrate the force from a large surface area to a smaller one and therefore increases the pressure.

$$P = \frac{F}{A} \quad (4.2.1)$$

$$P = \frac{\partial F}{\partial A} \quad (4.2.2)$$

The drawback of using a large contact surface is that the precision is lowered; depending on the situation this may or may not be relevant. If the sensor were to be used on a smaller muscle, fingers or jaw for example, the larger surface area may cover several individual muscles; consequently the output signal would contain vibrations all the muscles making specific muscle activation detection difficult. However, for larger muscles such as the arms or thighs the drawback is not as significant.

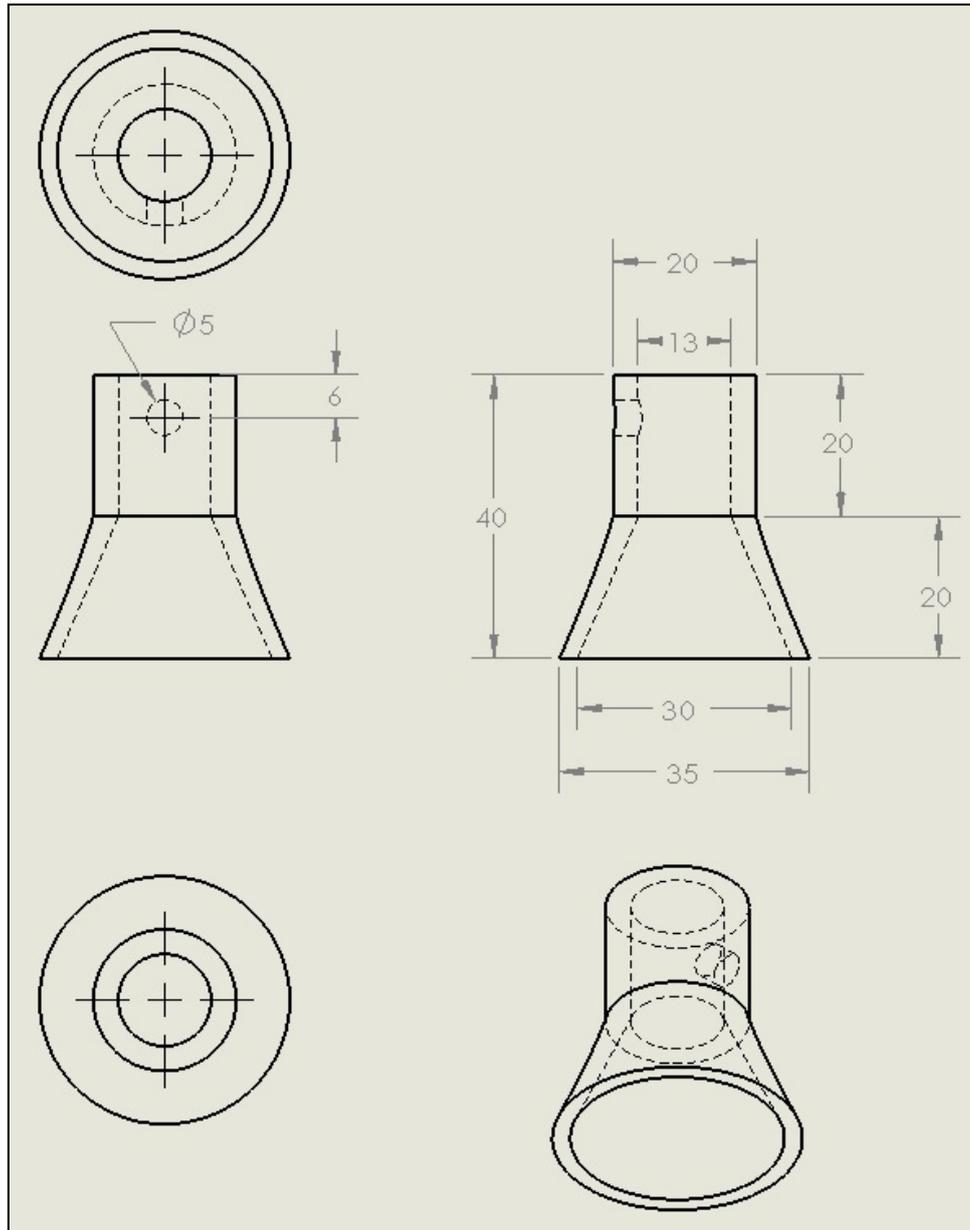


Figure 4.2.1 – Microphone Holder Assembly

Figure 4.2.1 shows the design of the microphone assembly. The microphone is wrapped in a strip of latex rubber and slotted in to the narrow end and secured with a single M5 bolt; the latex strip will help form an air tight. It is important to note the position of the vent on the capsule, commonly there are either rear vented or side vented microphones. The vent must be exposed to atmosphere in order to provide a pressure differential either side of the diaphragm. If the vent position were inside the sealed air volume and there would be a significant decrease in the pressure differential and would in turn decrease the mechanical response. Accommodations

can be made if a side vented microphone is all that is available simply by drilling a ventilation hole in the side of the holder, but the simplest method is to select a rear vented microphone.

The material used for the elastic membrane is common latex rubber; it provides the greatest combination of elasticity, strength and availability. The membrane was stretched to a specific surface tension maintained at that value throughout all tests. Because the surface tension of the membrane would affect the mechanical sensitivity of the sensor it was important to maintain the same tension throughout all the tests so as not to bias the results, the actual value can be arbitrary as long as the tension remain consistent throughout the experimentation. However, sensitivity could be improved with testing to determine the optimal surface tension which would allow for maximum spring back and minimal hindrance to skin displacement.

Oster and Jaffe (1980) determined that water was a far better acoustic coupler than air due to its higher density, even though there is a considerable loss in signal intensity, the quality of the signal is far greater since there is better transition between mediums i.e. the transition from skin to water rather than skin to air, which tends to cause a lot of reflective noise. A water coupler is a good method of removing the interference produced between the sensor and skin; however this would require a hydrophone rather than a standard microphone so the idea was rejected.

Consideration was given to the idea that constant compression and expansion of the air column may cause temperature fluctuations which may in turn affect the results especially after prolonged use. However, the pressure changes can be viewed as adiabatic as the period of compression is too short and the pressure is too small to produce any significant quantity of heat (Courteville et al., 1998; Watakabe et al., 2001).

Using Watakabe's 2001 research as a basic guideline of the dimensions of the assembly the receiver end of the assembly has a diameter of 15mm with an additional 2mm of clearance to allow for the air tight packing. The assembly's length was set at 15mm which allowed for 5mm of the microphone to be exposed. The

larger contact end of the assembly was set 20mm; Aluminium was used as the construction material.

Fastening

Ideally, the optimal fastening method would be to use an adhesive which would allow the sensor to simply be glued on to the surface of the skin, free from any external fastenings i.e. straps. Unfortunately, the size and weight of the sensor makes this approach impractical.

Instead with no better alternative straps will need to be used. The initial plan for fastening the sensor was to simply use elastic straps to strap the sensor to the muscle. Preliminary tests showed that the increase in the cross sectional area of the muscle will cause the sensor to embed itself in the muscle since the strap will prevent the sensor moving in relation to the skin surface.

Courteville, Gharbi and Cronu, (1998), faced a similar problem; a method of attaching their sensor to the surface of the skin without restricting its ability to displace its position in relation to normal skin displacement. Their solution was a compromise rather than a complete solution, the sensor was embedded in a foam arm band. The foam would allow for a moderate level of skin/sensor displacement while at the same time providing a relatively consistent level of force to keep the sensor on the surface of the skin. A similar fastening method is used for this sensor.

4.3 Preamplifier

Majority of transducers have a very low voltage output so in most cases transducers are coupled with a preamp which helps boosts the signal power to an easier to use level; for example microphones will often use a preamp to boost the signal to line strength to allow for transmission across cable to avoid loss of signal.

Preamplifiers are often located close to the transducer to lower the effect the impedance of the line or cable has on the signal. Preamps serve only to boost control signal power, so even though it can provide relatively high voltage gains, it provides no significant current gains. Preamps typically only use a very small 1.5V to

9V supply, often a battery source. In some cases the preamp does not amplify the signal at all but rather acts as an impedance bridge.

The most important properties for a preamplifier to have are high input impedances and low output impedances; this allows to the maximum transfer of signal power. In addition to have good power transfer, it is important that the preamp provide stable signal amplification, fluctuations at this point will cause high levels of noise in the system which cannot be removed.

4.3.1 Implementation

Line Preamplifier with Phantom Power Supply

INA103 is a low cost instrumentation amplifier with both low noise and distortion properties.

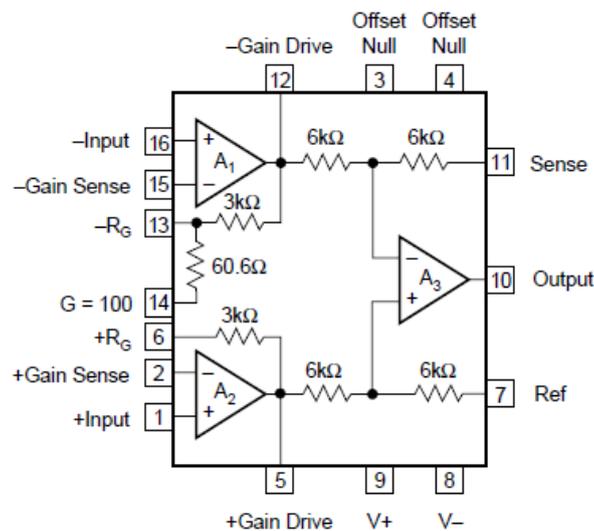


Figure 4.3.1 – INA103 Internal Schematic

Typical specifications:

- Supply Voltage: $\pm 9V$ to $\pm 25V$
- Input Voltage: $\pm 11V$ to $\pm 12V$
- Output Voltage: $\pm 21V$ @ $\pm 25V$ supply
- Input Impedance: $60M\Omega$
- Gain: 1 to 1000

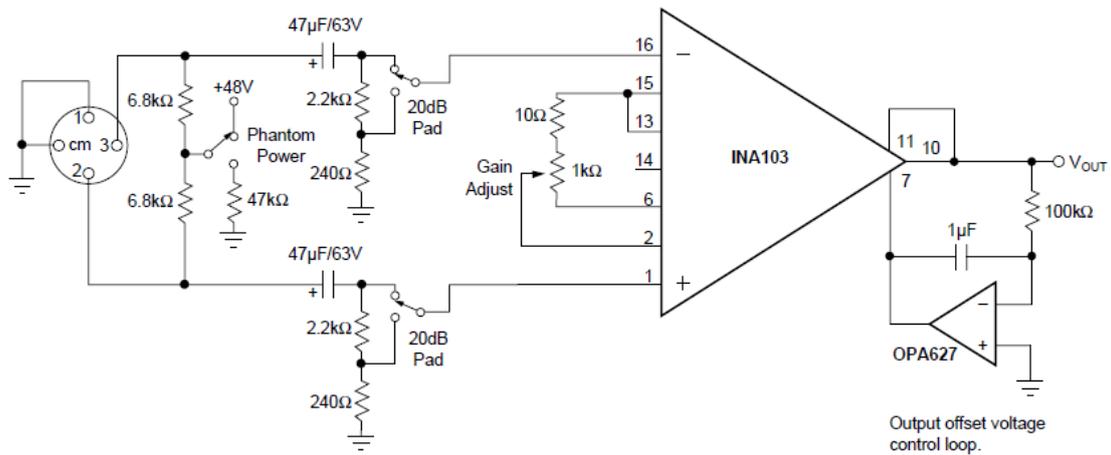


Figure 4.3.2 – Microphone Preamplifier with Phantom Power Provision

Panasonic WM61 Preamplifier

The preamplifier design was taken from an existing preamplifier design of the Behringer ECM8000 microphone which used a microphone capsule with similar characteristics to the WM61. Figure 4.3.3 shows the schematic of the preamplifier. The preamplifier requires a phantom power supply of 48V.

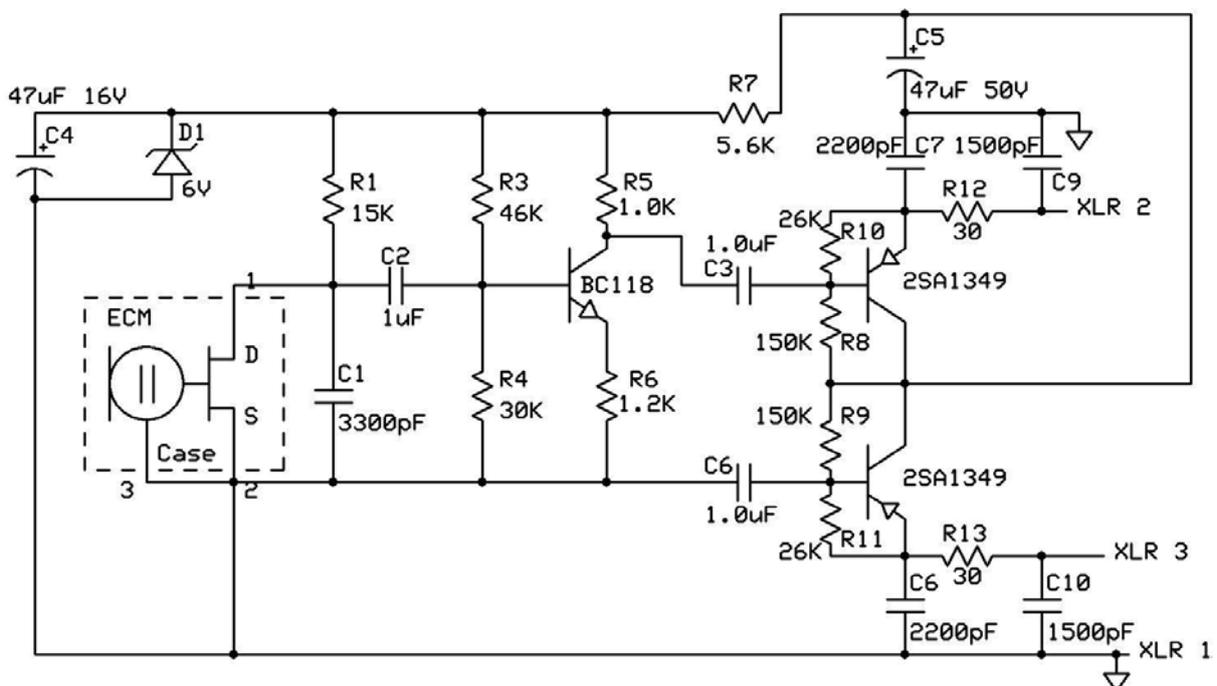


Figure 4.3.3 – Preamplifier for Panasonic WM61

The current preamplifier design shown in Figure 4.3.3 only has a lower frequency range of 20Hz, however by increasing the values of the capacitors C2, C3 and C6 to ~10 μ F, the lower limit should go even lower to around 10Hz – 15Hz.

4.4 Filter

There are several types of signal filters, the three predominant ones are Butterworth, Bessel and Chebyshev. Each filter has its own set of unique responses which make them suited for various situations.

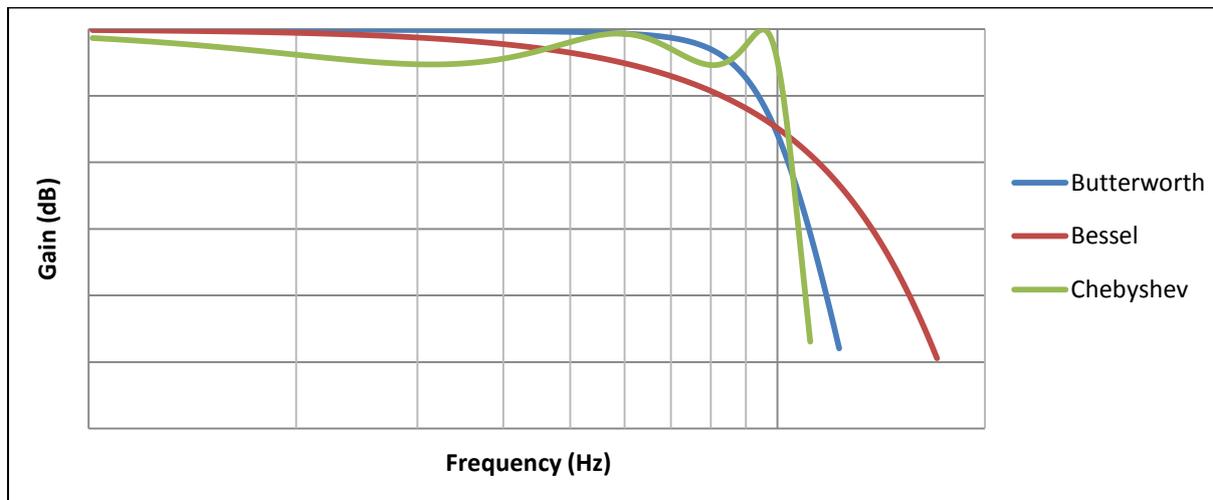


Figure 4.4.1 – 5th Order Filter Responses

The above Figure 4.4.1 shows the frequency responses of 5th order low pass filters to highlight the unique characteristics which each filter displays.

Bessel filters have a good flat pass band; however it has a slow attenuation rate causing signal attenuation far before the cut-off frequency. Bessel filters would not be effective for any applications which require clear and precise frequency cut-offs. Chebyshev filters are almost the opposite of Bessel filters; they have high attenuation rates giving very sharp and clean cut-offs however are not maximally flat in the pass band, particularly near the cut-off regions. Butterworth filters are a good compromise between Bessel filters maximally flat pass band and Chebyshev's high attenuation rate.

Topography

Topography concerns the arrangement of the filter and is usually only considered in realizing analogue filters. Similar to having various filter methodologies, topology also has several designs again there are three common ones the Cauer, Sallen-Key and multiple feedback (MFB). Sallen-Key topology is by far the most commonly implemented of the topologies because of its simplicity. Performance wise the Sallen-Key is decent, however the design is prone to stop band leakages where undesired signal fluctuations occur in the stop band due to imperfect op-amp impedances; in an effort to remedy this issue the MFB topology was created. For the MFB topology an additional feedback loop was added running in parallel with the first feedback loop, which increased the complexity of the filter but fixes the stop band leakage. Active filters utilise op-amps in their circuits which allow the filter itself to have some kind of signal gain and without the inclusion of inductors, signal degradation is significantly less than that of a passive filter.

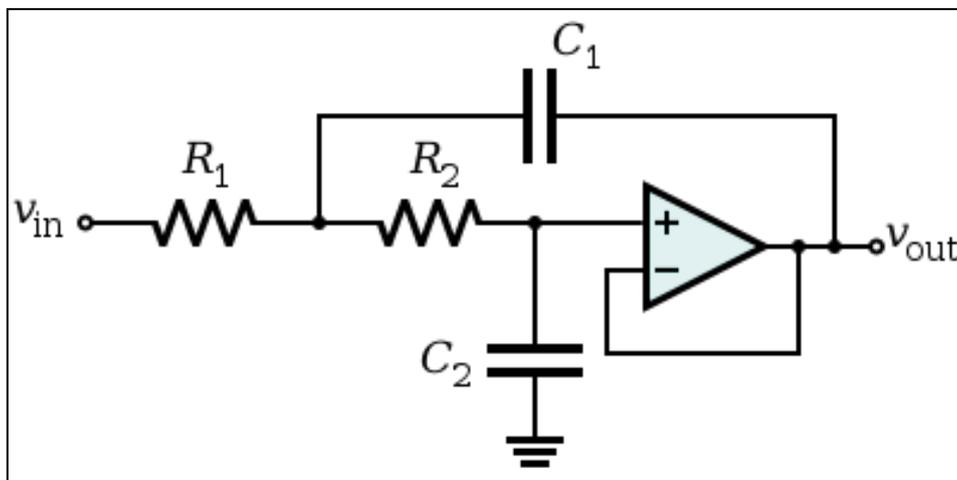


Figure 4.4.2 – Common Low Pass Sallen-Key Topology

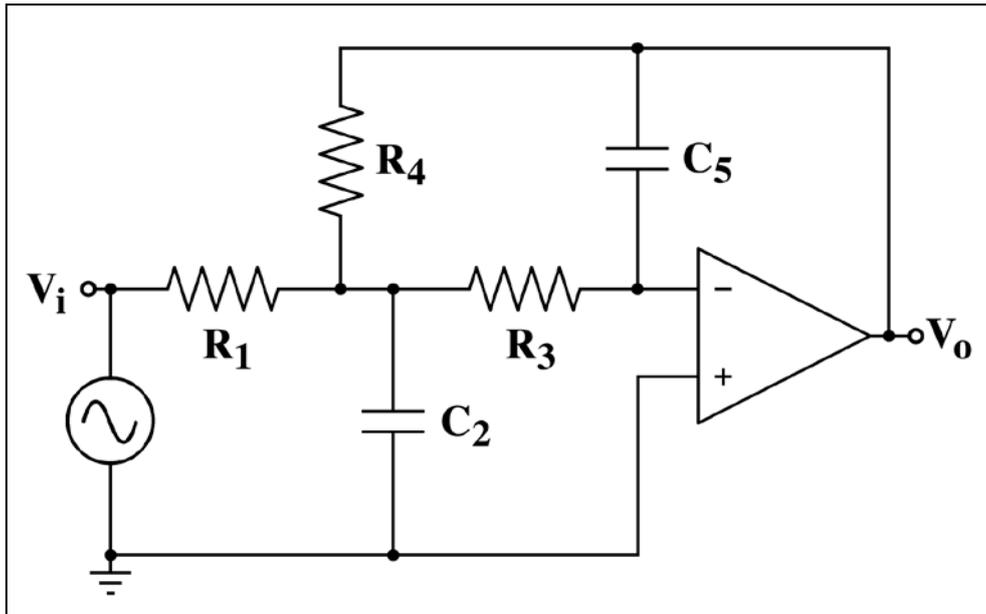


Figure 4.4.3 – Common Multiple Feedback (MFB) Topology

The Cauer (commonly known as ladder) topology is a popular passive filter topology which rather than using active components such as op-amps like the Sallen-Key and MFB, only use passive components like capacitors, inductors and resistors. The obvious advantage of a passive filter is that they are extremely easy to design and implement however does lack the level of performance granted by the more sophisticated active filters. Passive filters are heavy reliant on their components tolerances, in particular in very high or very low frequency applications, components values can be impractical (either being too large or too small), passive filters provide no gain and can degrade the signal power, and finally even though a low order filter is easy to design and implement, passive filters with similar response characteristics of an active filters need to be much higher order filters.

4.4.1 Implementation

The majority of muscle vibration frequencies fall within 1Hz to 100Hz. A hardware filter was designed and implemented. Additionally, a software filter will be used for some tests and/or if the hardware filter is inadequate.

A low pass filter is implemented to allow only the infrasonic vibrations to pass; any vibrations of higher frequencies are treated as noise and filtered out. The cutoff frequency is set at 100Hz. Courteville et al. (1998) states that muscle vibrations can

occur between a range of 2Hz to 100Hz and Murphy et al. (2008) mentions that “no signals of interest occur beyond 70Hz”.

Murphy et al. (2008) suggested in their research that for the low pass component of the filter to use a 5th order Butterworth filter. This would provide the greatest maximally flat response in the pass band; both Bessel and Chebyshev are ill-suited for this application as the Butterworth has a far better attenuation rate than the Bessel filter and a better pulse response than the Chebyshev.

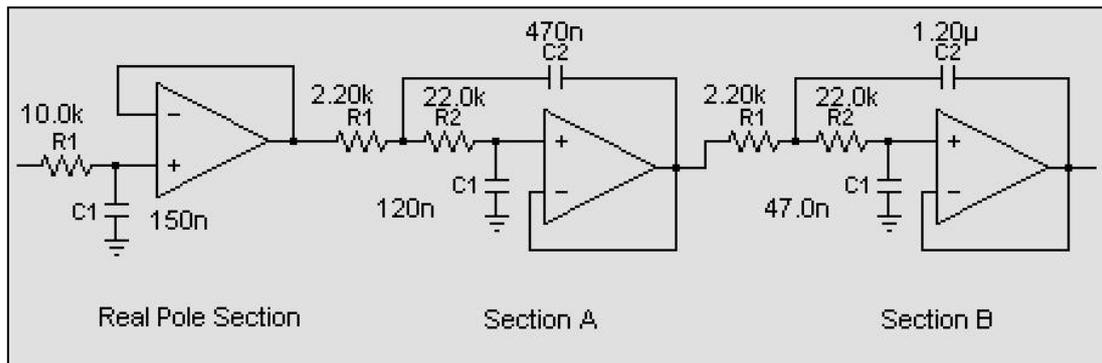


Figure 4.4.4 – 5th order Low Pass Butterworth Filter

The design implemented in the above Figure 4.4.4 is a 5th order Butterworth filter arranged in a Sallen-Key topology. The Op-Amps used in the design are OPA27GP ultra low noise precision op amps.

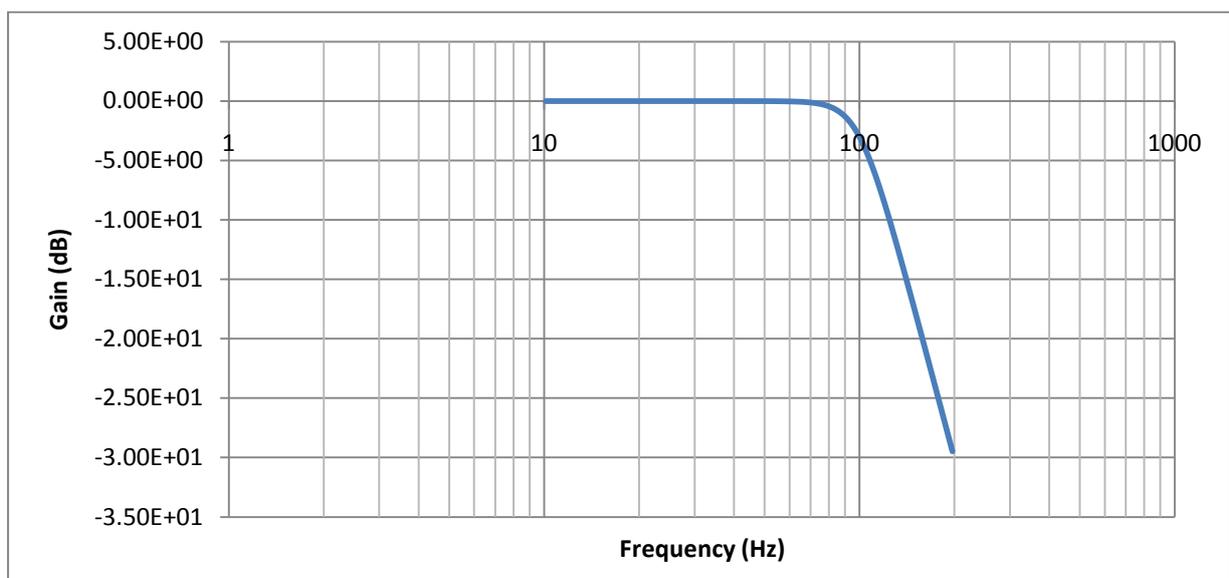


Figure 4.4.5 – Frequency Response of 5th order Filter

The expected frequency response of the filter is shown in Figure 4.4.5. The frequency response model was determined using Texas Instrument's filter design program FilterPro V2.0.

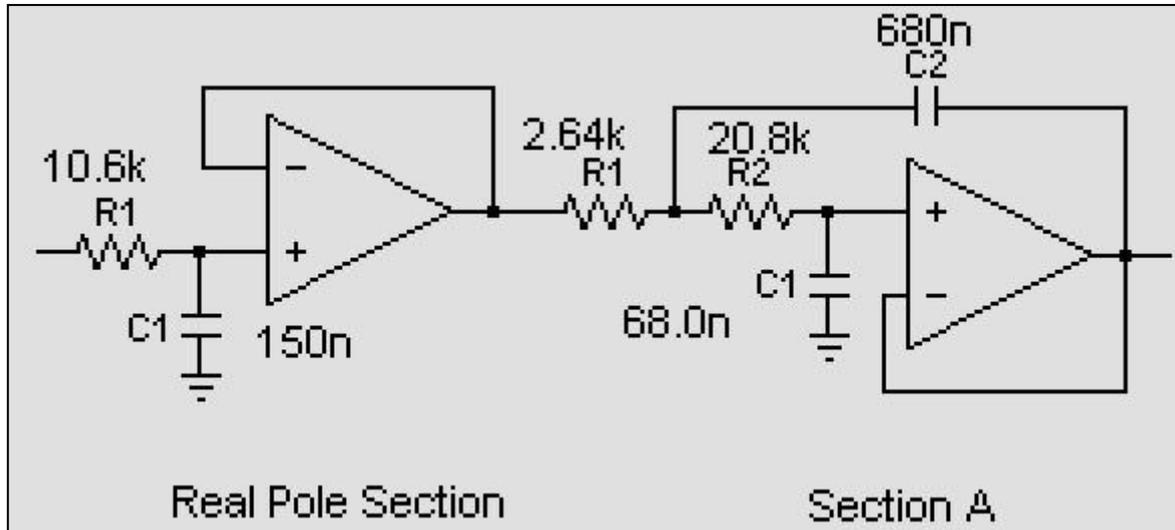


Figure 4.4.6 – 3rd order Low Pass Butterworth Filter

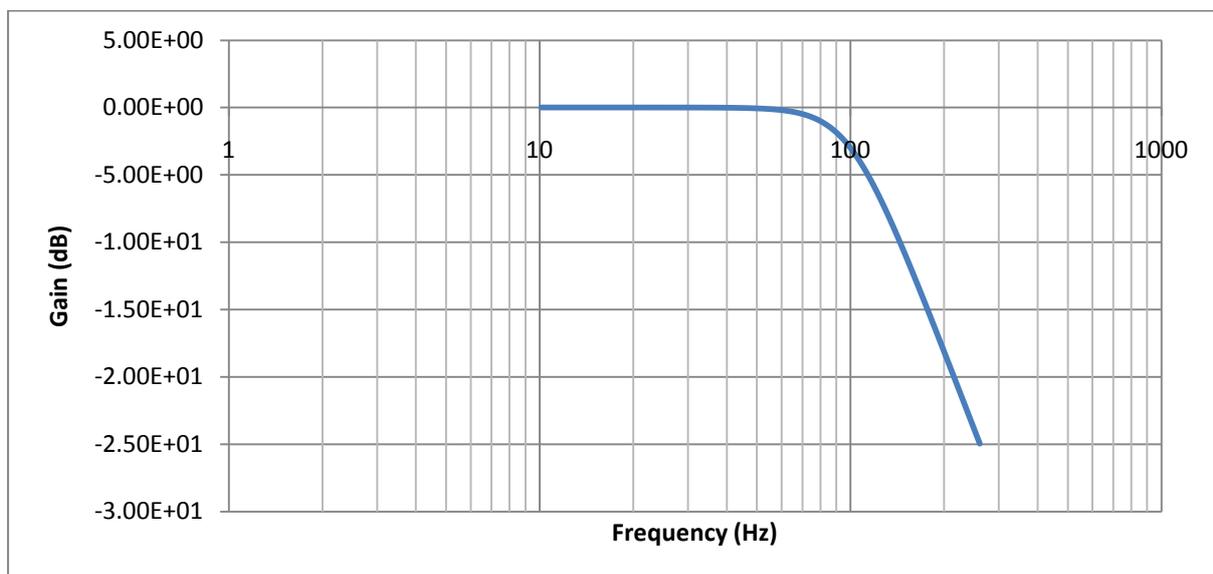


Figure 4.4.7 – Frequency Response of 3rd order Filter

A 3rd order version (Figure 4.4.6 and Figure 4.4.7) of the filter was modelled to compare performance. As you can see the 3rd order attenuation begins near the 50Hz mark where the 5th order attenuation begins around 70Hz. Even though the

difference may seem small the entire bandwidth is only 100Hz so a difference of 20Hz is a loss of 20% of the signal.

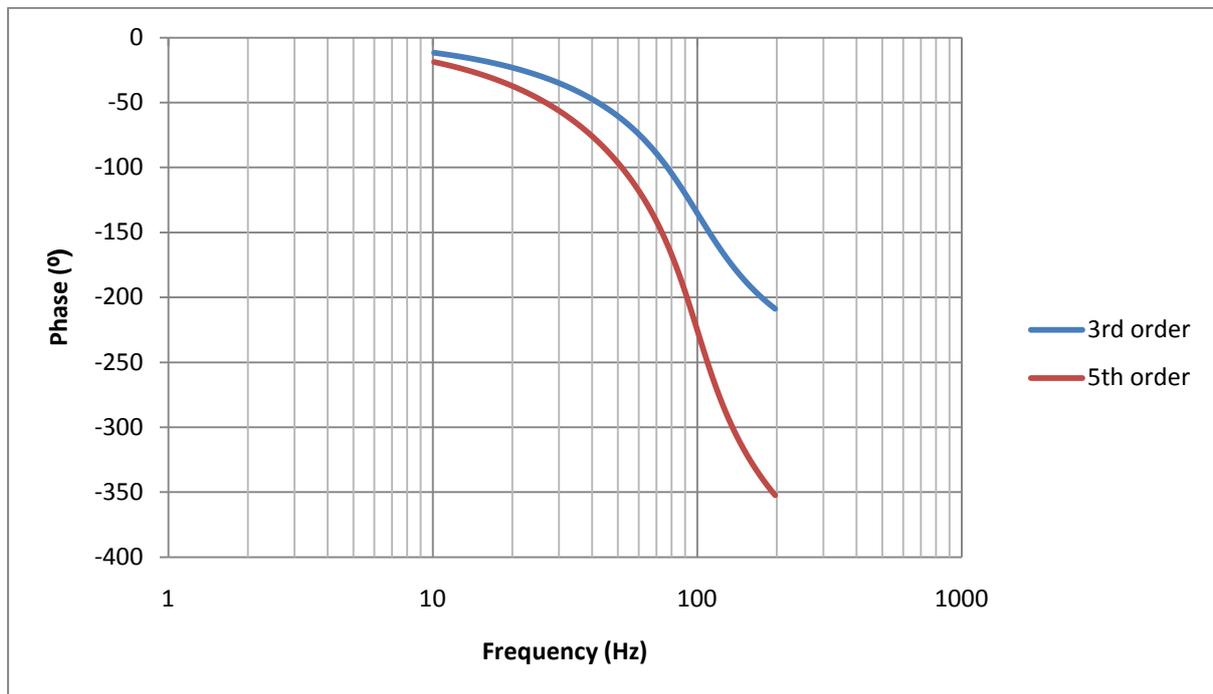


Figure 4.4.8 – Phase Response

The phase shift for the 5th orders it relatively large especially toward the higher end of the frequency range; shifting upwards of 200° at 100Hz. The 3rd order filter, on the other hand, only has around 130° phase shift and average phase shift is considerably lower than the 5th order filter. Despite the larger phase shift, it is not expected to cause any significant detriment to the sensors overall performance. The advantages provided by the increased precision of the 5th order filter far outweigh the disadvantage of a larger phase shift and slower settling time.

In a situation where you are using an accelerometer, a high pass filter would be needed to filter out the static DC signal (gravity), which would usually lead to further degradation of the signal (Murphy et al., 2008). The advantage of using a microphone is that the static DC signal is always 0V as long as no external forces are applied to the sensor.

Matlab Simulink

In addition to using Texas Instruments' FilterPro, the filter design was also modelled on Matlab R2009a using the Simulink component. Modelling the filter on Matlab made it possible to better understand the behaviour of the filter and test the filter without having to build it.

Matlab is a numerical computing environment, used predominantly in mathematical applications such as matrix manipulation, function plotting, algorithm manipulation, etc. Although text based there are a number of add-on toolboxes which can provide graphical support. One such add on is the Simulink toolbox, which provides a graphical simulation environment. Simulink is similar to other graphical programming environments such as National Instruments LabView™ and Microsoft Robotics Studio. Simulink itself contains several different components (or Libraries). Libraries can be filled with component blocks which replicate mechanical bodies and joints, electronic components, neural networks, etc.

To simulate the hardware filter design the Simulink and SimScape/SimElectronics libraries were used. SimScape blocks are programmed to simulate real physical components i.e. resistors, capacitors, etc. Simulink contains general blocks which can be used by essentially all other libraries, these include display scope, signal generator, multiplexors, etc. The resulting models shown below, Figure 4.4.9 and Figure 4.4.10.

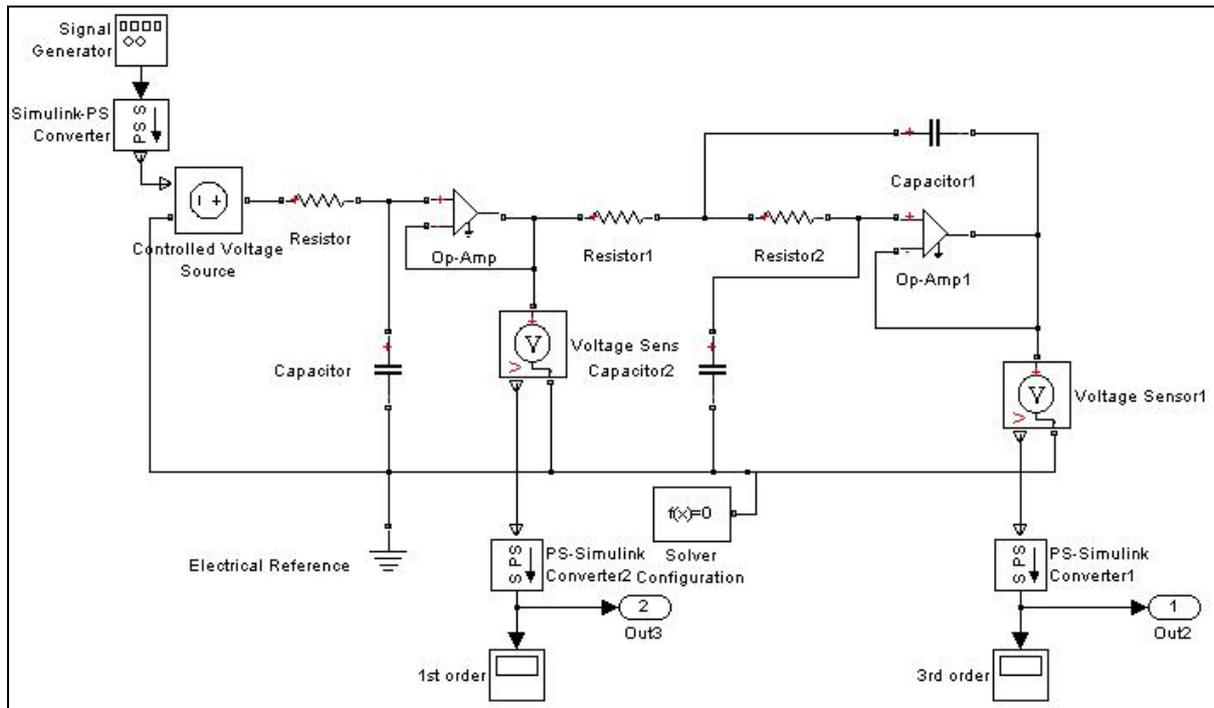


Figure 4.4.9 – 3rd Order Filter Model

The *controlled voltage source* is needed to provide an ideal voltage input, typically not always needed when SimElectronics and SimScape are used independently, but when Simulink and SimScape blocks are used together the *controlled voltage source* block is required.

For each output, an *output port* block was added. In some cases where scope blocks are not used the output should not be left floating so an output port block can be used to tie down the output however in this case because the output is tied to a scope the output port is not necessary.

The *Solver configuration* block is similar to the *electrical reference* block, where the electrical reference (ground) block defines the physical/electrical environment of the circuit, the *solver configuration* block defines the mathematical environment or rather defines the solver settings which will be implemented during the simulation of the model. These settings include options like the tolerances, iterations, steady state simulation, etc. For both filter models, the settings were left on default; 1×10^{-9} tolerance and start simulation at steady state. By starting the simulation at steady state you don't have to contend with the instability at the start when the capacitors

are just powering up. If the simulation isn't started at steady state, initial voltage values can be set for the capacitors.

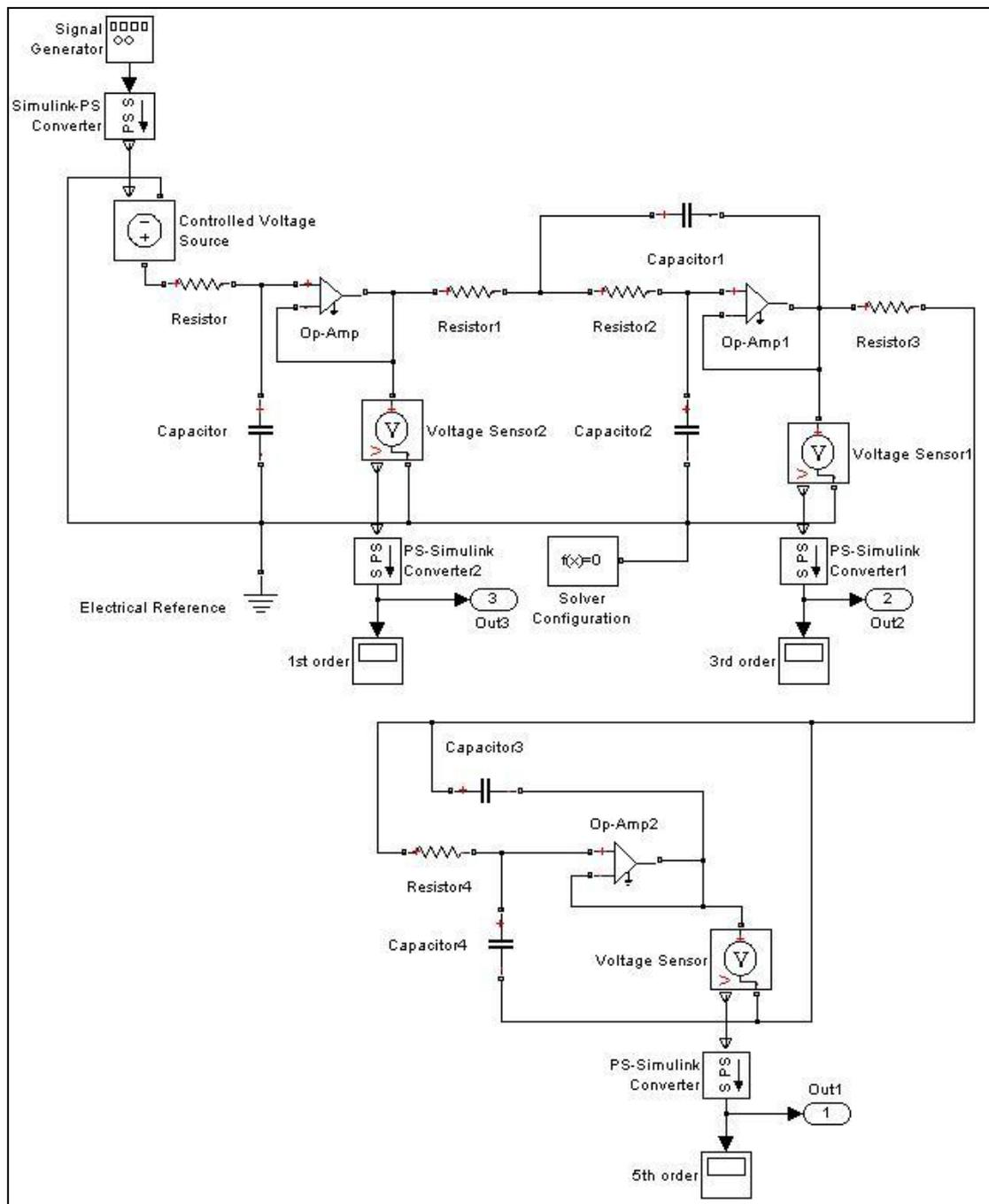


Figure 4.4.10 – 5th Order Filter Model

A signal generator block was used to produce various frequency signals to test the performance of the filter, even though the noise generator block is available, the

signal generator provided far more control over the test signals required. A noise generator was used only as basic gauge of performance.

In both Figure 4.4.9 and Figure 4.4.10, you can see that both the input must be linked using a Simulink converter. The reason for this is because the signal generator is not actually part of the SimScape/SimElectronics library so there are some cross compatibility issues, the main problem being that SimScape blocks are designed to replicate real physical components where as Simulink blocks are typically ideal/ theoretical blocks. Therefore the Simulink signal must be converted into a “physical” signal before it can be applied to the physical SimScape blocks. Similarly, the output of the filter signal must be again converted back into a Simulink signal before it can be used by the Simulink Scope block.

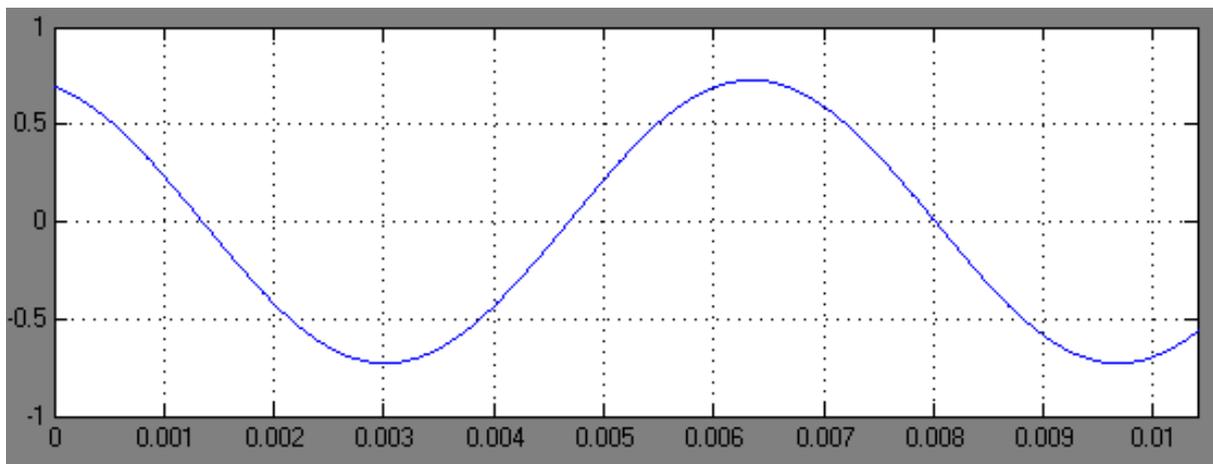


Figure 4.4.11 – Matlab Simulation 1st Order Filter

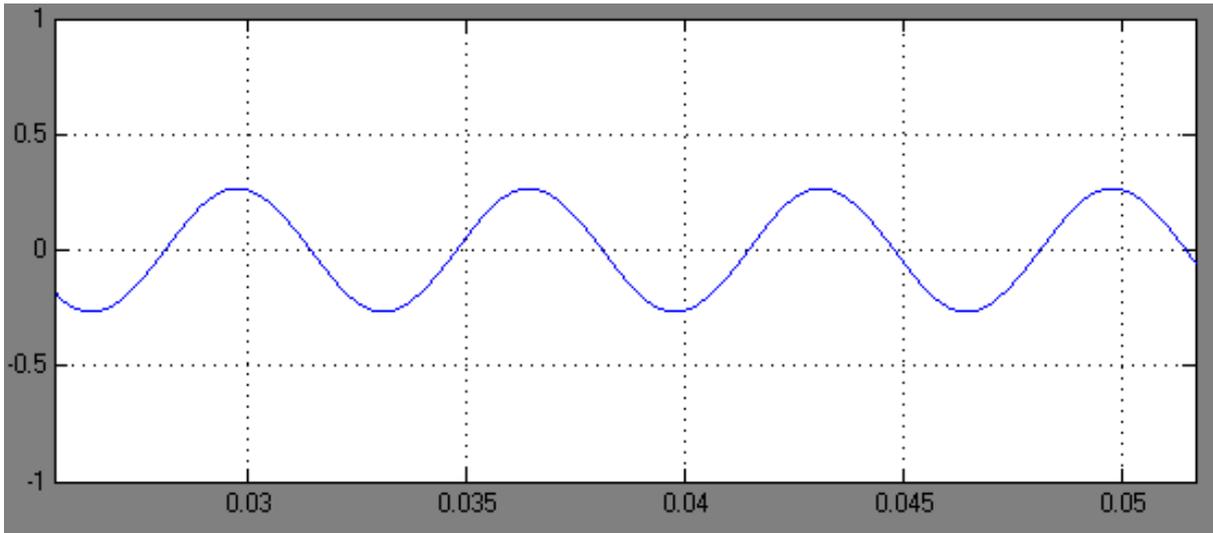


Figure 4.4.12 - Matlab Simulation 3st Order Filter

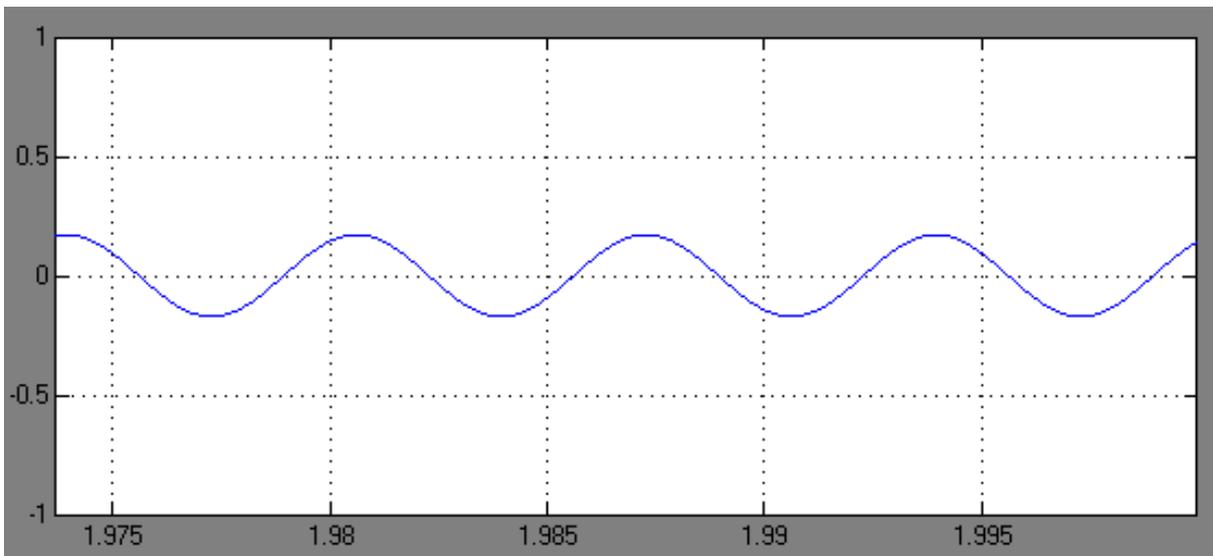


Figure 4.4.13 - Matlab Simulation 5st Order Filter

Figure 4.4.11, Figure 4.4.12 and Figure 4.4.13 are simulations of the filters at 100Hz.

4.5 Cadsoft Eagle (Easily Applicable Graphical Layout Editor)

Eagle is a simple program used to design and develop printed circuit boards (PCBs). It is a simple process of laying out a basic electronic schematic and converting it into a physical board layout.

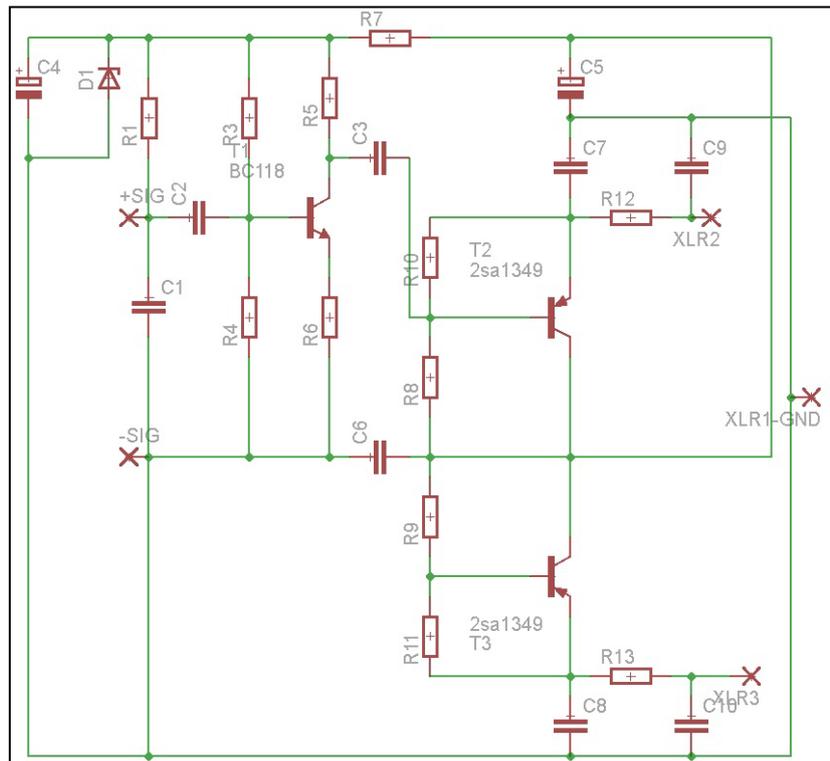


Figure 4.5.1 – Microphone Preamp Schematic

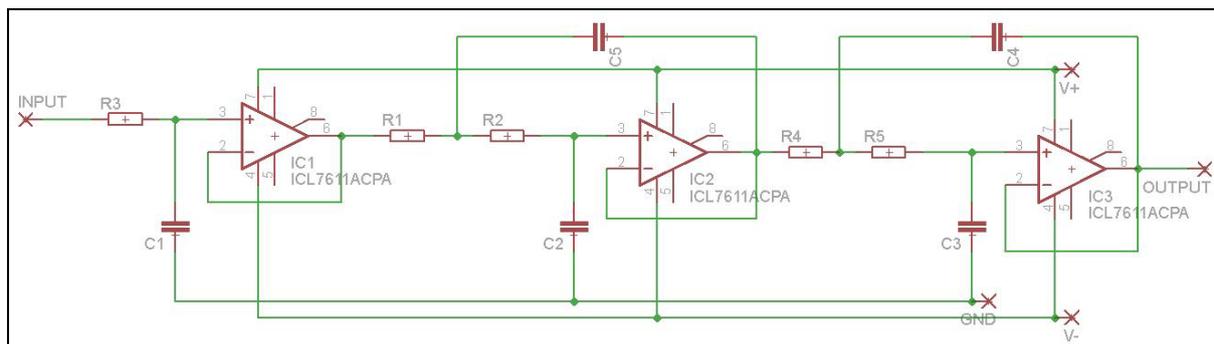


Figure 4.5.2 – Filter Schematic

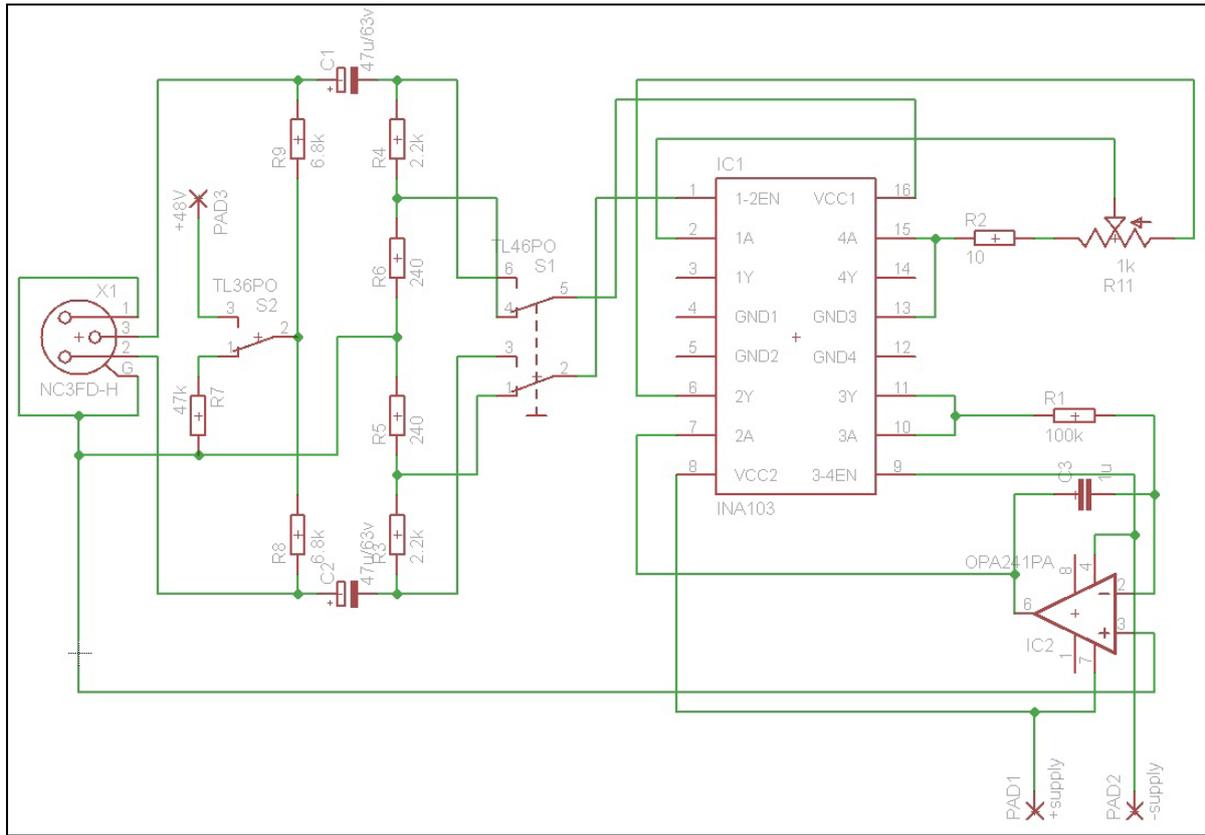


Figure 4.5.3 – Line Preamp Schematic

As shown above both in Figure 4.5.3 and Figure 4.5.2, schematics are laid out in a similar fashion to a standard circuit diagram. The parts locations and placement is unimportant and will normally be placed in a similar or identical fashion to the circuit diagram. The placement becomes important later, when the tracks and holes are being laid out. As you design the circuit on the schematic side it is important to pick appropriate sized parts, there will be several different sizes for each part even though the size will not affect the schematic drawing the board layout will be in accurate.

Eagle has an auto route function which automatically routes tracks to each of the pads, several routes maybe tested to allow for the optimal track route, increased complexity of the schematic will increase the routing time. Furthermore, occasionally some tracks are un-routable with the auto route function so they will be left to be routed manually. Refer to Appendix 8.2 below for PCB board layouts.

4.6 Data Acquisition

Data acquisition (DAQ) is a process of capturing analogue signals from the physical world and converting them into digital format for analysis or manipulation by computer or microprocessor. A fully implemented data acquisition system (DAS) typically consists of:

- Sensors – not exclusively part of the DAQ since a generated signal can be also be captured. However, a DAS typically contains a sensor of some kind.
- Signal Conditioners – the signal conditioning circuit only serves as a preamplifier to boost the signal to a useable state. Furthermore, advanced signal conditioning is typically done via software.
- Analogue to Digital Converter – most important component of the DAQ.

4.6.1 Hardware

For hardware, a National Instruments data acquisition card (PCI-6024E), a third party 50 pin screw terminal block with 50 pin ribbon cable and 68 pin to 50 pin adapter. The PCI-6024E has 16 analogue input channels. Should the need arise for an array of muscle sensors, a maximum of 16 sensor will be useable. The DAQ has a resolution of 12 bits and has a maximum input voltage of 10V. Additionally, the PCI-6024E has 8 bidirectional digital input/output channels, should the need arise for digital inputs.

Input Range	Gain	Resolution	
		NI 6023E/6024E/6025E	NI 6034E/6035E/6036E
-10 to +10 V	0.5	4.88 mV	305 μ V
-5 to +5 V	1	2.44 mV	153 μ V
-500 to +500 mV	10	244 μ V	15.3 μ V
-50 to +50 mV	100	24.4 μ V	1.53 μ V

Table 1 – Input Ranges for NI 6023E/6024E/6025E

A 50 pin screw terminal block was used because of its availability, the drawback being that an adapter was needed to convert the 68 pin socket of the PCI card to the 50 pin configuration of the ribbon cable. Additionally, with the missing 18 pins from

the terminal block, not all of the PCI-6024E's functions can be used, but only the analogue input terminals were needed in this particular situation.

AI GND	1	2	AI GND
AI 0	3	4	AI 8
AI 1	5	6	AI 9
AI 2	7	8	AI 10
AI 3	9	10	AI 11
AI 4	11	12	AI 12
AI 5	13	14	AI 13
AI 6	15	16	AI 14
AI 7	17	18	AI 15
AI SENSE	19	20	AO 0 ¹
AO 1 ¹	21	22	AO EXT REF ¹
AO GND ¹	23	24	D GND
P0.0	25	26	P0.4
P0.1	27	28	P0.5
P0.2	29	30	P0.6
P0.3	31	32	P0.7
D GND	33	34	+5 V
+5 V	35	36	AI HOLD COMP
EXT STROBE	37	38	PFI 0/AI START TRIG
PFI 1/AI REF TRIG	39	40	PFI 2/AI CONV CLK
PFI 3/CTR 1 SRC	41	42	PFI 4/CTR 1 GATE
CTR 1 OUT	43	44	PFI 5/AI SAMP CLK
PFI 6/AO START TRIG	45	46	PFI 7/AI SAMP CLK
PFI 8/CTR 0 SRC	47	48	PFI 9/CTR 0 GATE
CTR 0 OUT	49	50	FREQ OUT

¹ No connects appear on pins 20 through 23 of devices that do not support AO or use an external reference.

Figure 4.6.1 – I/O Block Layout

MIO-16 Connector			Extended I/O Connector				
AI 8	34	68	AI 0	AI 24	34	68	AI 16
AI 1	33	67	AI GND	AI 17	33	67	AI 25
AI GND	32	66	AI 9	AI 18	32	66	AI 26
AI 10	31	65	AI 2	AI 27	31	65	AI 19
AI 3	30	64	AI GND	AI 20	30	64	AI 28
AI GND	29	63	AI 11	AI 21	29	63	AI 29
AI 4	28	62	AI SENSE	AI 30	28	62	AI 22
AI GND	27	61	AI 12	AI 23	27	61	AI 31
AI 13	26	60	AI 5	AI 32	26	60	AI 40
AI 6	25	59	AI GND	AI 41	25	59	AI 33
AI GND	24	58	AI 14	AI 34	24	58	AI 42
AI 15	23	57	AI 7	AI 35	23	57	AI 43
AO 0 ¹	22	56	AI GND	AI GND	22	56	AI SENSE 2
AO 1 ¹	21	55	AO GND	AI 44	21	55	AI 36
AO EXT REF ¹	20	54	AO GND	AI 37	20	54	AI 45
P0.4	19	53	D GND	AI 38	19	53	AI 46
D GND	18	52	P0.0	AI 47	18	52	AI 39
P0.1	17	51	P0.5	AI 48	17	51	AI 56
P0.6	16	50	D GND	AI 49	16	50	AI 57
D GND	15	49	P0.2	AI 58	15	49	AI 50
+5 V	14	48	P0.7	AI 51	14	48	AI 59
D GND	13	47	P0.3	AI 52	13	47	AI 60
D GND	12	46	AI HOLD COMP	AI 61	12	46	AI 53
PFI 0/AI START TRIG	11	45	EXT STROBE	AI 54	11	45	AI 62
PFI 1/AI REF TRIG	10	44	D GND	AI 55	10	44	AI 63
D GND	9	43	PFI 2/AI CONV CLK	NC	9	43	NC
+5 V	8	42	PFI 3/CTR 1 SRC	NC	8	42	NC
D GND	7	41	PFI 4/CTR 1 GATE	NC	7	41	NC
PFI 5/AO SAMP CLK	6	40	CTR 1 OUT	NC	6	40	NC
PFI 6/AO START TRIG	5	39	D GND	NC	5	39	NC
D GND	4	38	PFI 7/AI SAMP CLK	NC	4	38	NC
PFI 9/CTR 0 GATE	3	37	PFI 8/CTR 0 SRC	NC	3	37	NC
CTR 0 OUT	2	36	D GND	NC	2	36	NC
FREQ OUT	1	35	D GND	NC	1	35	NC

NC = No connect

Figure 4.6.2 – Cable I/O Layout (Extended Version Used)

4.6.2 Software

For software, National Instruments' LabView™ was used, and more specifically LabView™ Signal Express 2009. LabView™ is development environment software for visual signal and instrumentation development. It is used quite extensively in data acquisition applications, signal processing and control system designs.

SignalExpress is a simpler cut down version of LabView™, with heavy emphasis on DAQ applications, where data can be captured and displayed quickly. Signal Express also allows for basic to intermediate signal processing and analysis, for example filtering and fast Fourier transforms for power spectrum graphing.

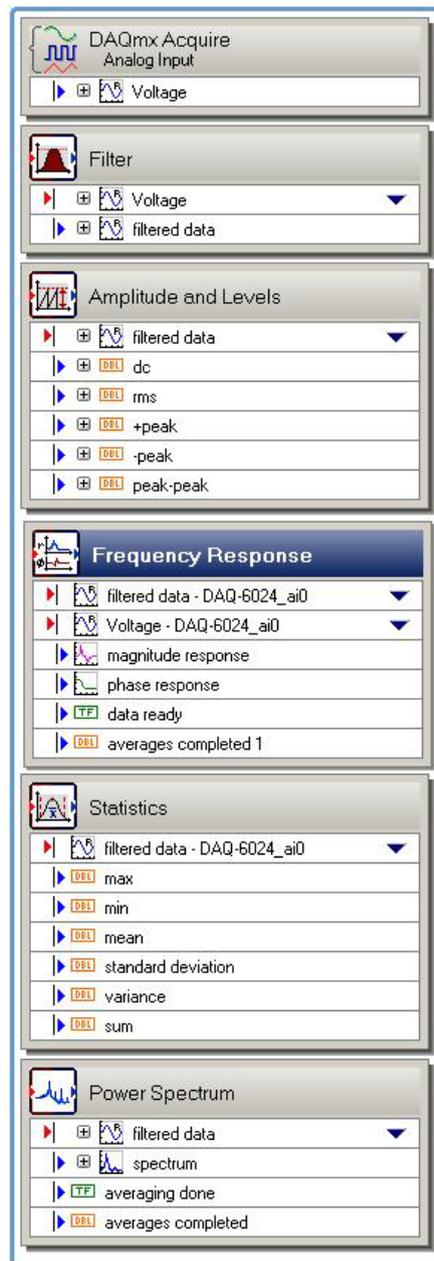


Figure 4.6.3 – Signal Express Capture and Display Program

These stages (blocks) are in sequential order. Process runs from top to bottom through each stage.

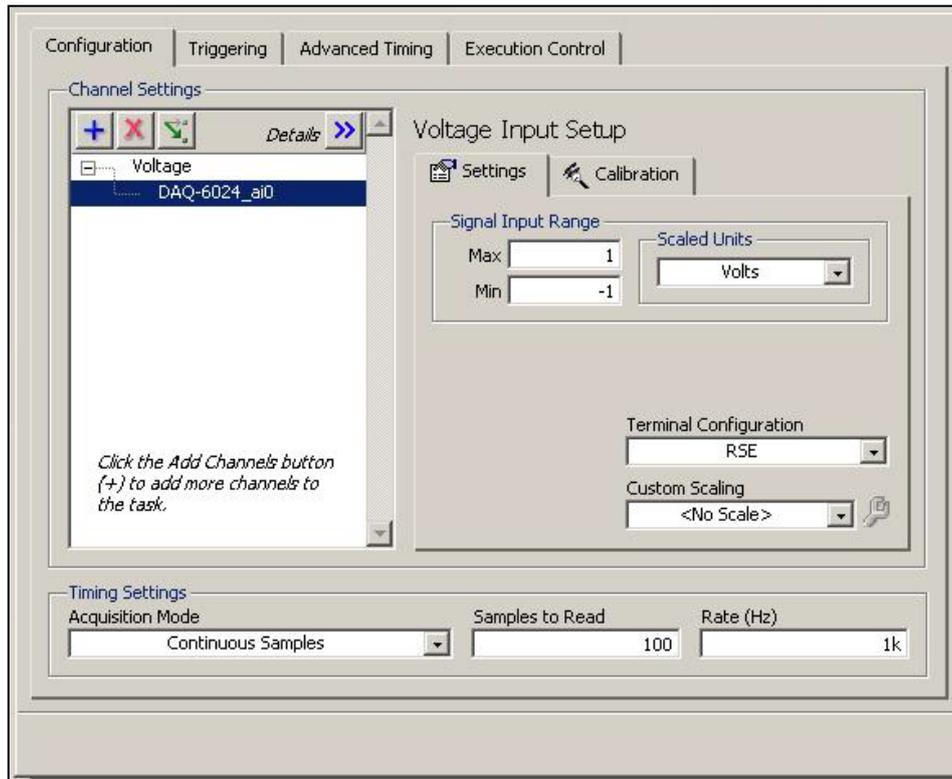


Figure 4.6.4 – DAQmx Acquire Block

The first stage (block) is the acquisition block. A standard analogue input was used for input source. Analogue input channel 1 (AI01) has its own 16-bit ADC and can handle input voltages up to +10V unipolar and $\pm 10V$ bipolar.

The output levels of the preamp were maintained well within a $\pm 1V$ range so as not to cause any clipping of the signal. High resolutions would allow you to be more precise with taking measurements however in this situation it was decided that excessively high resolutions would be redundant and/or detrimental when trying to determine muscle activity. With real time applications in mind continuous samples were taken at 200Hz and a 100 sample buffer size; the buffer size and sampling rate are National Instruments recommended value.

A digital input could have been used, but would have required an external/independent ADC, to convert the analogue signal to a digital one before inputting into the DAQ. Since the PCI-6024E's onboard ADC had a sufficient level of resolution an independent ADC was not required.

There are 3 terminal configurations: differential (DIFF), referenced single ended (RSE) and non-referenced single ended (NRSE). RSE was used, where the positive output of the preamp was connected to AI line and the negative is tied to ground (AIGND).

Filter

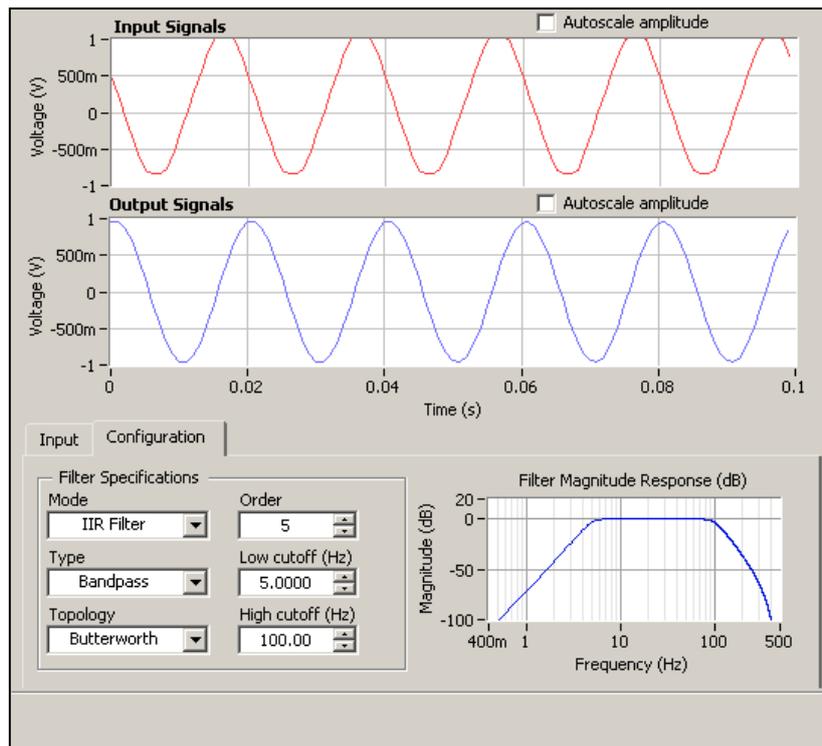


Figure 4.6.5 – Filter Block

After source acquisition the signal is filtered. The filter is a 5th order band pass Butterworth filter with a 5Hz lower cutoff and a 100Hz high cutoff frequency. A Butterworth filter provides the best flat response and roll off.

Since LabView™ was used for testing a software filter was able to be implemented. However, this sensor is designed to be used as a standalone embedded system without computer software assistance and in that case a hardware filter will be used. A higher order filter provided the most desirable frequency response, however something like a 100th order hardware filter is completely impractical to design and implement, as such a 5th order Butterworth will be used instead.

A test was completed with a 100th order software filter and then with a 5th order software filters and then finally against realised 5th order hardware filter. The results were compared to determine the variation in performance particularly between hardware and software. Refer to Appendix 8.3.

Amplitudes and Levels

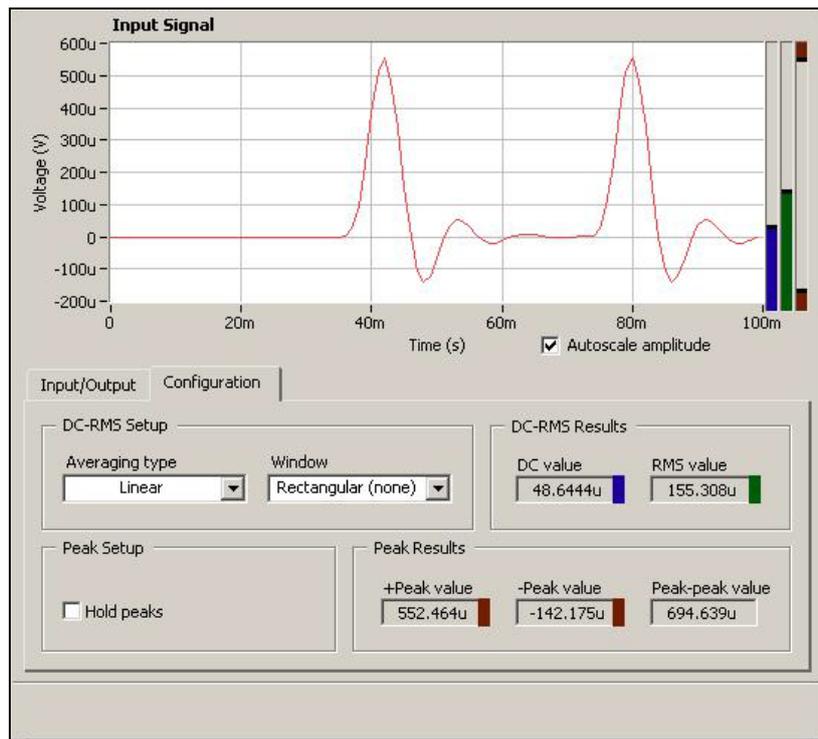


Figure 4.6.6 – Amplitude and Levels Block

The amplitudes and levels block is a simple data display block; it is used to display the filtered data. The amplitudes and levels block contains an average, range and deviation bar for visual analysis. Additionally, it has a “peak hold” and numerical peak max and peak min functions which is very useful for any real time analysis.

Frequency Response

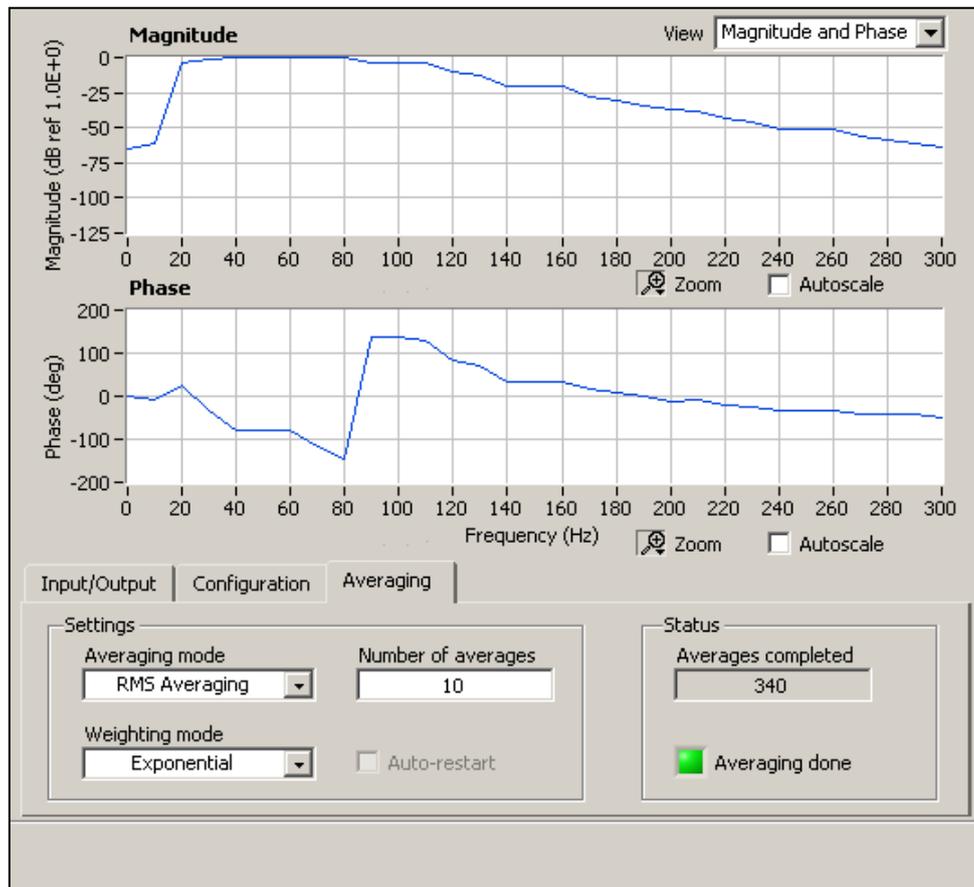


Figure 4.6.7 – Frequency Response Block

The frequency response block allows for real-time analysis of the frequency and phase response of 2 signals. For analysis of the internal signal, i.e. signals that exist purely within SignalExpress' program loop, simply set the input as the signal before entering the filter (Voltage-DAQ6024_ai0) and set the signal output as (filtered data-DAQ6024_ai0). To determine the frequency response of hardware, this would require the use of 2 input ports on the DAQ. One input port (ai1) is connected to the input channel of the hardware device and other input port (ai2) is connected to the output channel. In the frequency response block, the input would be set to Voltage-DAQ6024_ai1 and the output would be set to Voltage-DAQ6024_ai2. RMS averaging was done on the output to provide a more stable graph.

Statistics

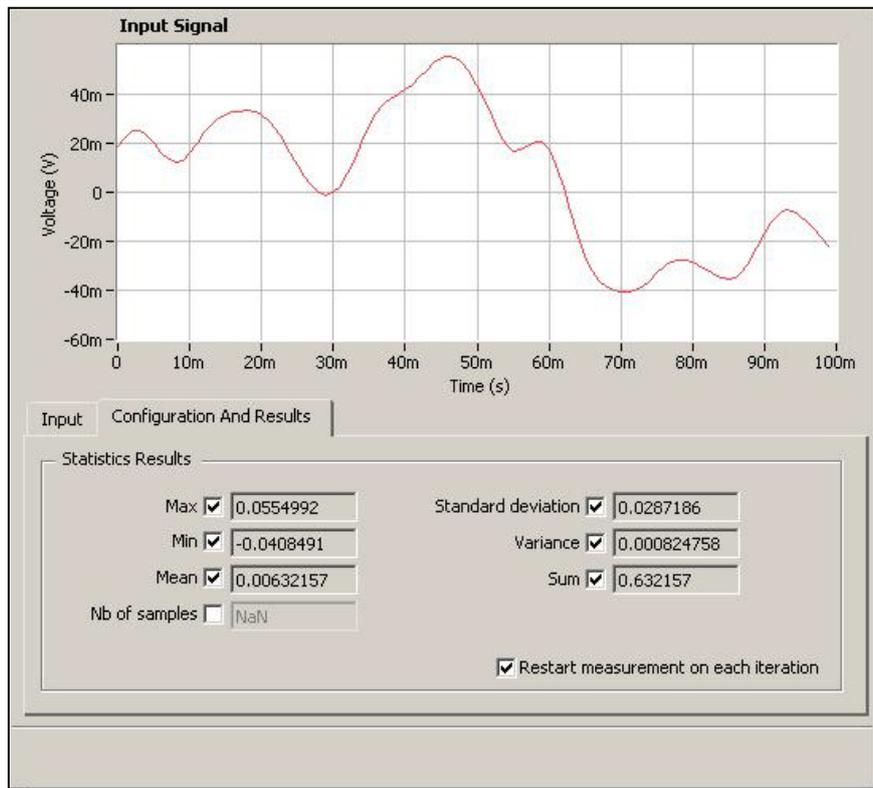


Figure 4.6.8 – Statistics Block

This statistics block provides basic numerical display of various information, i.e. mean, variance, max, etc. The statistics block is particularly useful for any statistical analysis which may be required.

Power Spectrum

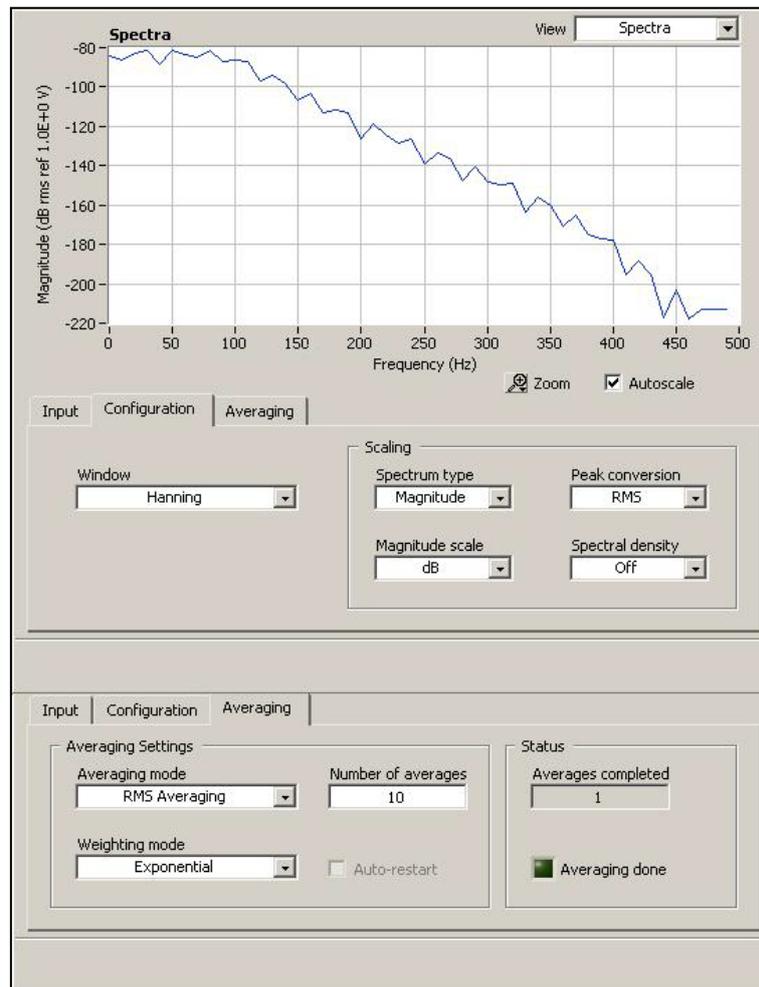


Figure 4.6.9 – Power Spectrum Block

Signal Express is able to do FFT's to provide a simple frequency domain graph which will help in analysing the frequency properties of the input signal. The power spectrum was set to Hanning window and with a linear magnitude scale. The power spectrum is particularly useful during development and testing since it allows you use the output to calibrate the cut off range of your filters. Since the muscle vibration frequency varies from person to person it would be a valuable asset to be able to calibrate the filter to the specific user and that will allow for much more precise muscle measurement.

Input averaging was turned off as it would introduce more lag in the output, this obscures any important vibration spikes or fluctuations that may be of interest during testing.

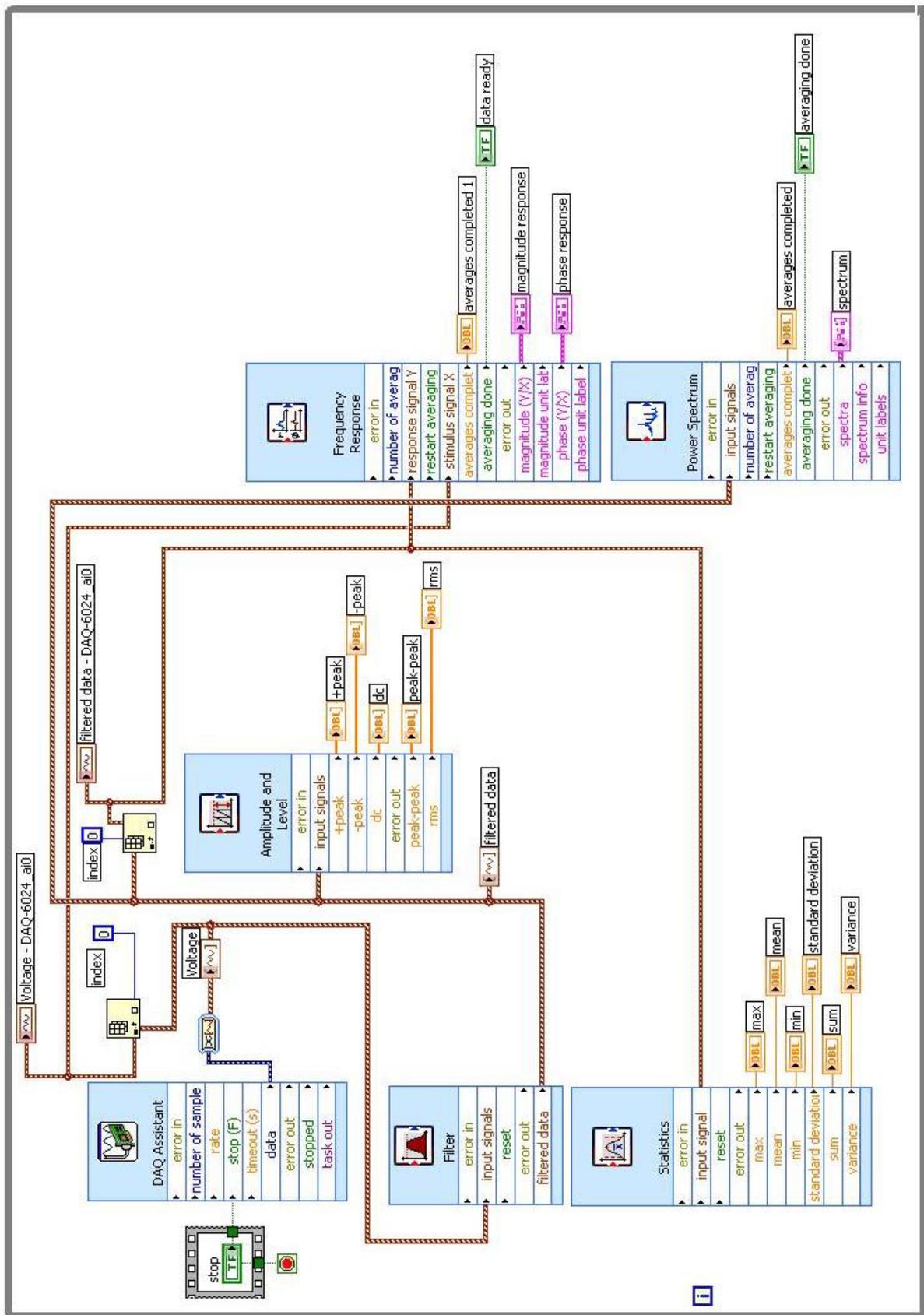


Figure 4.6.10 – LabVIEW Representation

Chapter 5 Testing

The main objective of testing at this stage is to attempt to replicate and observe various phenomena and events which were discussed in previous research. The two main phenomena include instance of muscle fatigue, the relationship between force and signal and to observe any unusual activity not discussed previously or unique to this sensor design. Additionally, the tests will also be used as a basis to determine the effectiveness of the sensor in relation to detection of muscle activity.

5.1 Procedure

Environment and Preparation

- Quiet location will be needed.
- Tests will be conducted in room temperature (20°C - 25°C).
- Subject should not have done any strenuous physical activity before the testing which may bias the results.
- Area the sensor will be applied to will need to have hair removed to reduce interference caused by the friction between the membrane and surface of the skin.
- The pressure of the air volume that couples the skin to the microphone will be the same as atmospheric pressure.
- The latex membrane will again be set to an arbitrary but suitable surface tension.

Test Subject

The same test subject was used throughout all of the tests in order to eliminate any unique variables the multiple subjects might introduce i.e. different sized muscles, different fatigue thresholds, etc.

The subject chosen to test the sensor is a 20 year old Asian male, approximately 182cm tall and 66kg. He has no pre-existing medical conditions and is relatively fit, exercising regularly on a daily basis. The subject is left handed and all tests will be done on the dominant side.

Tests will be done on the Biceps Brachii and Flexor Digitorum. The Bicep Brachii is one of the largest and strongest muscles along the arm. Since this is a relatively new and untested prototype, a larger and easier to access muscle was decided on. Additionally, because of the location of the muscle, both on the surface and relatively isolated there is a far lower chance of interference from other muscles. The Flexor Digitorum is located on the outer forearm. Its primary function is the flexing of the wrist and finger manipulation, as the name extensor suggests, the Flexor Digitorum is the opening/ extending component of finger manipulation.

Applying Sensor

1. Thin layer of glue will be applied to the rubber of the acoustic cone.
2. Acoustic cone will be placed on the muscle of interest.
3. The device straps will be attached firmly and securely.
4. Insert microphone allowing the top of the microphone to protrude approximately 5mm inside the sealed volume.
5. Tighten the locking nut until it holds the microphone assembly securely. Make sure not to over tighten it as to prevent damage to the microphone assembly itself.

5.2 Tests

The sensor will be put through several tests; two main muscles will be targeted: Biceps Brachii and Flexor Digitorum. Lift test will target the Biceps Brachii, the test will be conducted twice; once with a 2kg mass and a second time without to represent the difference between heavy and delicate activity respectively. A squeeze test will also be conducted; the squeeze test will target the Flexor Digitorum. Furthermore, an extra set of miscellaneous test will be conducted, consisting of increasing mass test, sensor placement test, muscle interference test and fine motor control test.

Lift Test (Test A w/ mass and Test B w/o mass)

The lift test will be conducted on the Bicep Brachii.

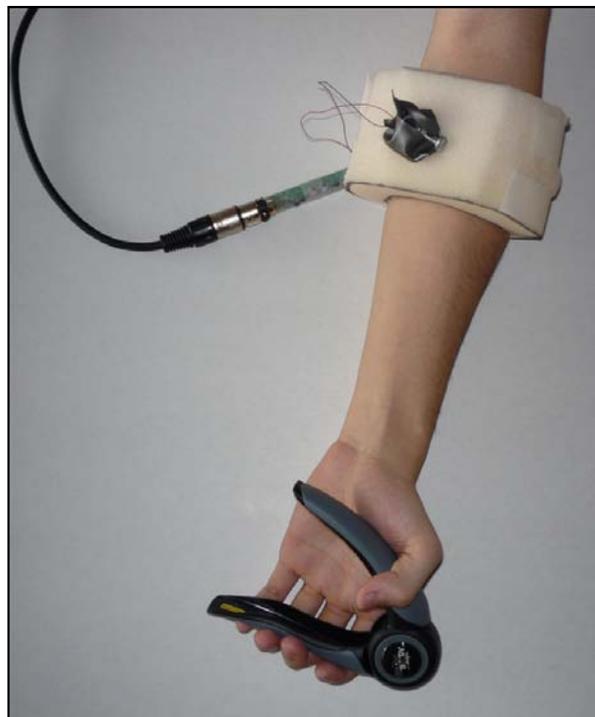
1. Subject will be standing and remain standing throughout the test.
2. The test arm will rest by the subjects side full extended and relaxed.
3. When instructed to at the appropriate time interval, the subject will smoothly flex the elbow 90° until the forearm is parallel to the floor and perpendicular to the upper arm.
4. The position will be held until the appropriate time interval has passed, while trying to remain as still as possible. After which the forearm will be moved back to its original location as smoothly as possible.



Squeeze Test (Test C)

Test will be conducted on the Flexor Digitorum.

1. The subject will be seated on a comfortable chair for the duration of this test.
2. The subject arm will be resting on a flat surface in a comfortable position.
3. When instructed to the subject will squeeze the grip until it is fully compressed and hold until the contraction time is over.
4. When the contraction time is up the subject will release the grip in a quick but controlled manner.



Additional Tests (Test D)

1. Increasing Mass (Sequence 8)

Sensor will be placed in the standard lift test position. The arm will start at a 90° flexion. A steadily increasing force will be applied to the hand until a total of 8kgs (26.68Nm) is reached.

The object of this test is to determine whether or not force and vibration amplitude are related. By eliminating any significant arm movement, vibration due to movement of the sensor is no longer a factor.

2. *Muscle Interaction (Sequence 9)*

This test will have 3 stages. The sensor will be placed on the Bicep Brachii, a standard squeeze test performed, without Bicep contraction, after which another squeeze test will be performed during a Bicep contraction.

This test will highlight any interactions between two different though not necessarily independent muscles.

3. *Focus Filtering (Sequence 6)*

A simple Bicep Brachii lift test will be conducted 6 times, each with a different frequency range.

- 5 – 100Hz (default)
- 5 – 20Hz
- 5 – 50Hz
- 20 – 50Hz
- 50 – 100Hz
- 80 – 100Hz

Muscle vibration frequencies occur around the 1 – 100Hz region. However, the actual bandwidth of an individual's muscle vibrations will be slightly different from person to person, depending on the individual's muscle properties. This test is an attempt to isolate the unique muscle vibration of the user, if possible at all, and consequently improve the test results and the sensors performance.

Notes:

- A complete run through of the test sequence will be completed twice for the lift test (no weight and 2kg weight) and once for the squeeze test, allowing for 10mins rest between each test to eliminate any accumulating fatigue influence.
- After each full test sequence, the subject should rest for 10 – 20min to eliminate any fatigue that may affect the results of the next test.

All sensor outputs will be measured and recorded by the DAQ and SignalExpress software.

5.3 Test Sequence

Tests A, B and C

Sequence 1 – Ambient no contact

Readings will be taken without being attached to muscle.

- Duration – 1min

Purpose: to establish an ambient measurement, this will show the inherent noise of the system.

Sequence 2 – Ambient contact

Reading taken while device is attached to the subject, he/she will try to keep as relaxed as possible.

- Duration – 1min

Purpose: to establish the ambient noise introduced by the muscle.

Sequence 3 – Single contraction

Readings will be taken of a single contraction.

- Relax – 20sec
- Contract – 10sec (0:20)
- Relax – 20sec (0:30)

Purpose: simple test to show a basic muscle vibration waveform.

Sequence 4 – Multiple uniform contractions

Readings will be taken of the subject performing 5 uniform contractions and uniform rest intervals.

- Relax – 10sec
- Contract – 10sec (0:10)
- Relax – 5sec (0:20)
- Contract – 10sec (0:25)
- Relax – 5sec (0:35)
- Contract – 10sec (0:40)
- Relax – 5sec (0:50)
- Contract – 10sec (0:55)
- Relax – 5sec (1:05)
- Contract – 10sec (1:10)
- Relax – 10sec (1:15)

Purpose: shows the behaviour of muscles during continuous activity

Sequence 5 – Multiple un-uniform contractions

Readings will be taken of a series of 5 contractions of varying time intervals with uniform rest intervals.

- Relax – 10sec
- Contract – 5sec (0:10)
- Relax – 5sec (0:15)
- Contract – 10sec (0:20)
- Relax – 5sec (0:30)
- Contract – 15sec (0:35)
- Relax – 5sec (0:50)
- Contract – 20sec (0:55)
- Relax – 5sec (1:15)
- Contract – 25sec (1:20)
- Relax – 10sec (1:55)

Purpose: builds on sequence 4 showing the behaviour of muscles during continuous and irregular activity.

Sequence 6 – Single long contraction

Readings will be taken of a long contraction.

- Relax – 10 sec
- Contract – 40sec (0:10)
- Relax – 10sec (0:50)

Purpose: shows the effect of muscle fatigue.

Sequence 7 – Multiple uniform long contractions

Readings will be taken of 2 long contractions with a short rest interval.

- Relax – 10sec
- Contract – 40sec (0:10)
- Relax – 5 sec (0:50)
- Contract – 40sec (0:55)
- Relax – 10sec (1:35)

Purpose: shows the effect of muscle fatigue on subsequent muscle activity.

Test D

Sequence 8

- Relax – 10sec
- Contract – 20sec (0:10)
- Relax – 10sec (0:30)

Sequence 9

Bicep Brachii

Flexor Digitorum

- Relax – 30sec Relax – 10sec
- Contract – 10sec (0:10)
- Relax – 20sec (0:20)

- Contract – 30sec (0:30)
- Contract – 10sec (0:40)
- Relax – 20sec (0:50)
- Relax – 10sec (1:00)

Sequence 10

- Relax – 10sec
- Contract – 20sec (0:10)
- Relax – 10sec (0:30)

5.4 Results, Analysis and Discussion

The results are recorded using the National Instruments PCI-6024E data acquisition card and used in conjunction with the National Instruments LabVIEW™ SignalExpress™. Analogue input port *ai0* was used for all the tests.

Only the *filtered data* was recorded fully recorded, other data was either impractical to record or the data irrelevant; this included *frequency response*, *power spectrum*, and *amplitudes and levels*. The voltage axis is set to $\pm 1V$ for ease of comparison and the time axis is set to relative time scales.

5.4.1 Lift Test Results (Biceps Brachii)

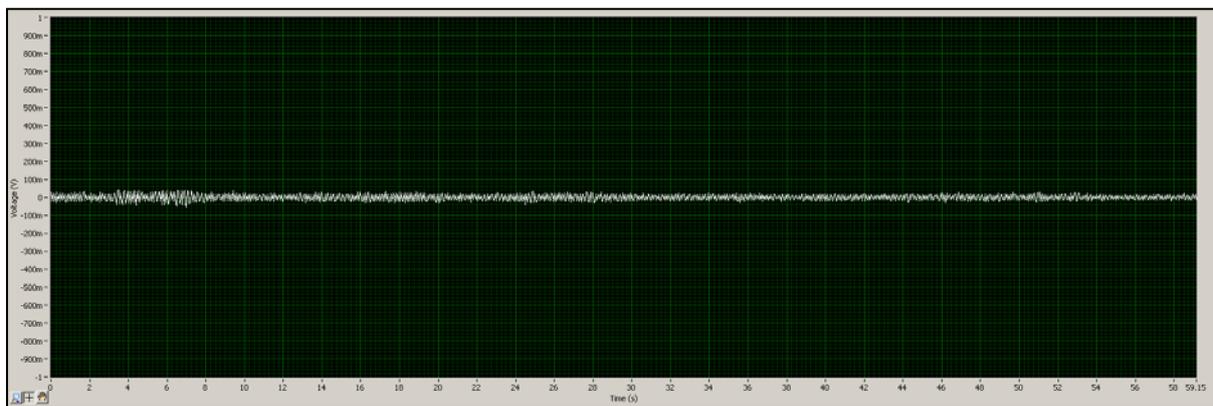


Figure 5.4.1 – Test 1

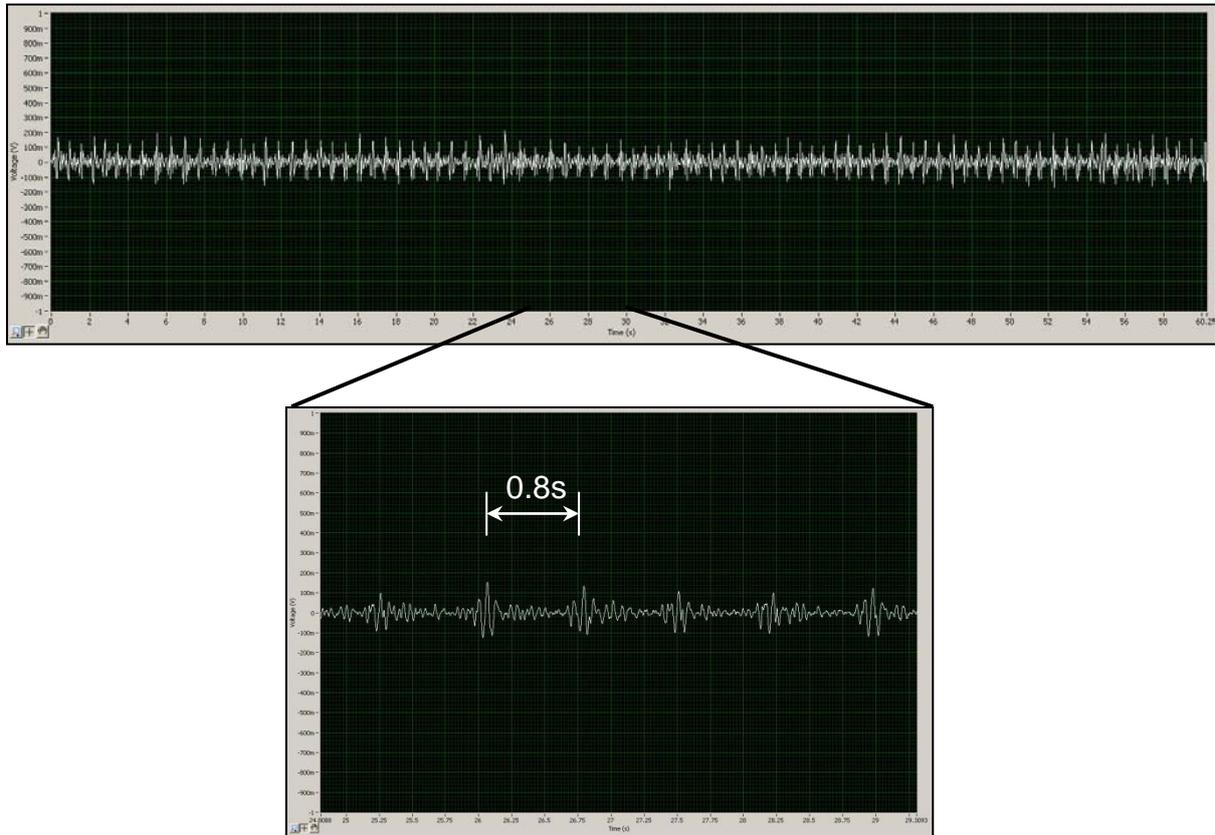


Figure 5.4.2 – Test 2a

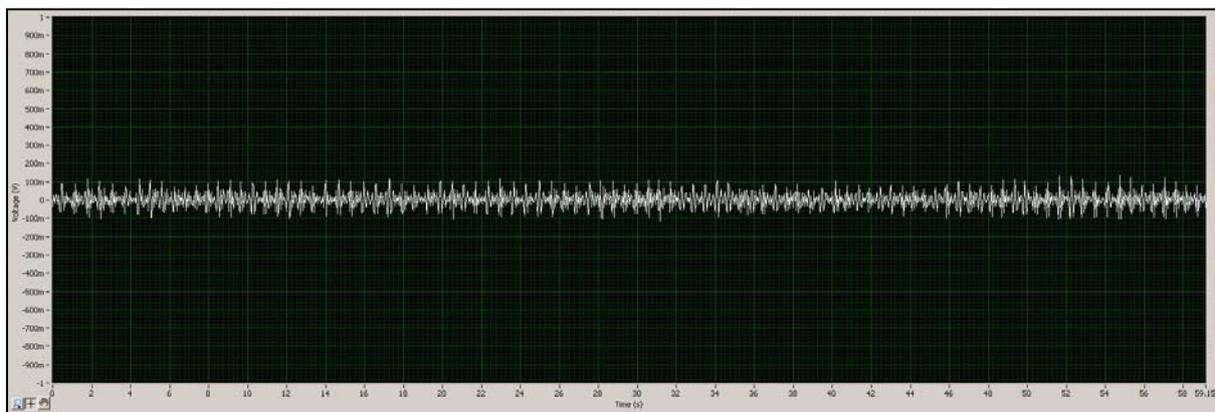


Figure 5.4.3 – Test 2b

Test 1, 2a and 2b were conducted to establish a reference. Test1 had no contact with the surface of the skin and by comparing that to test 2a and 2b where there was contact, the minimal noise level can be established. Test 1 showed an average signal peak levels were approximately 25mV, however, in tests 2a and 2b signal peaks averaged 50mV – 60mV and often peaking to as high as 100mV. The difference in the signals between Test 1 and 2 shows that even though there is no

deliberate muscle activation, the increase in vibrations show that involuntary activations occur even when the muscle is in a relaxed state. Additionally, a regular artefact is also detected, the artefact occurs every 0.8 sec, equating to a frequency of approximately 1.25Hz, or 75 bpm, consistent with the test subjects cardiac pulse rate (78bpm).

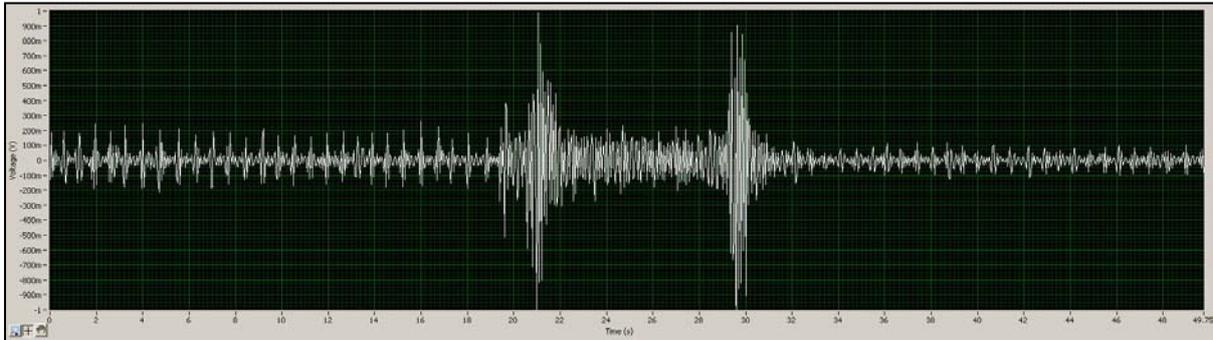


Figure 5.4.4 – Test 3a

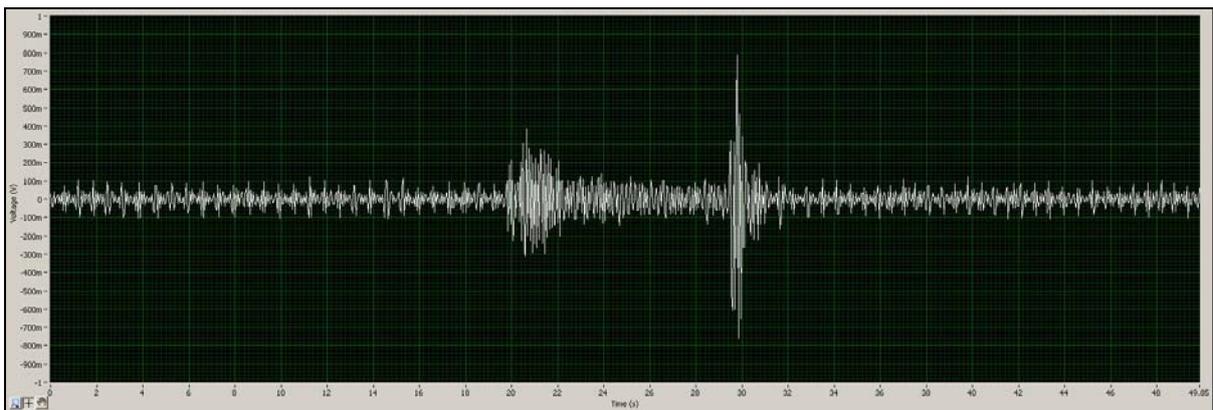


Figure 5.4.5 – Test 3b

Test 3 involved the subject producing a single, short contraction of the Bicep. The contraction was a total of 10 seconds long.

Test 3a and 3b, showed that an increase of 2kgs is not enough to significantly influence the signal levels; this was investigated by Barry and Geiringer, 1985. Barry and Geiringer showed with their research that the amplitude has an exponential relationship with the mass load applied to the muscle. Between 0 – 7lb (0 – 3.2kg) range there was very small amplitude difference. However, in their research it also showed that the larger the mass load applied the large deviation in their trial sets,

which shows that although the difference is much more significant, the results may not be completely reliable.

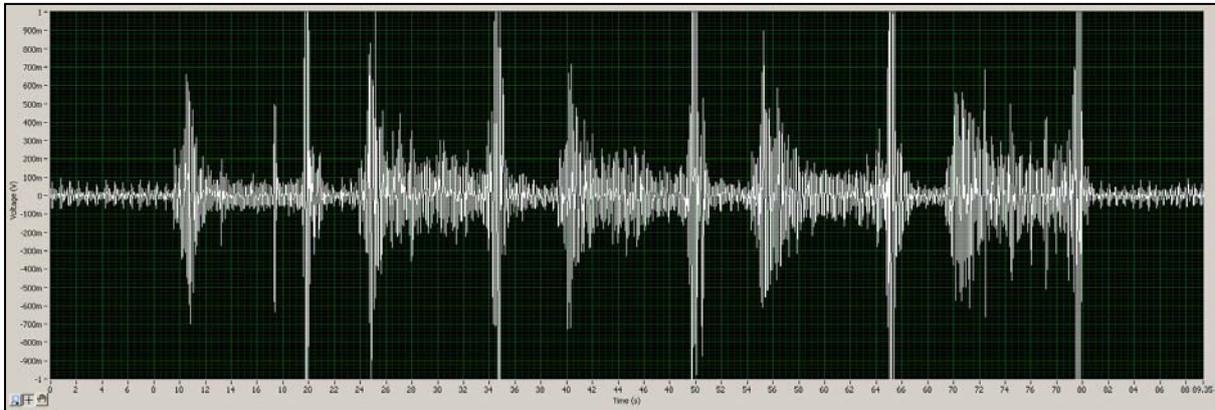


Figure 5.4.6 – Test 4a

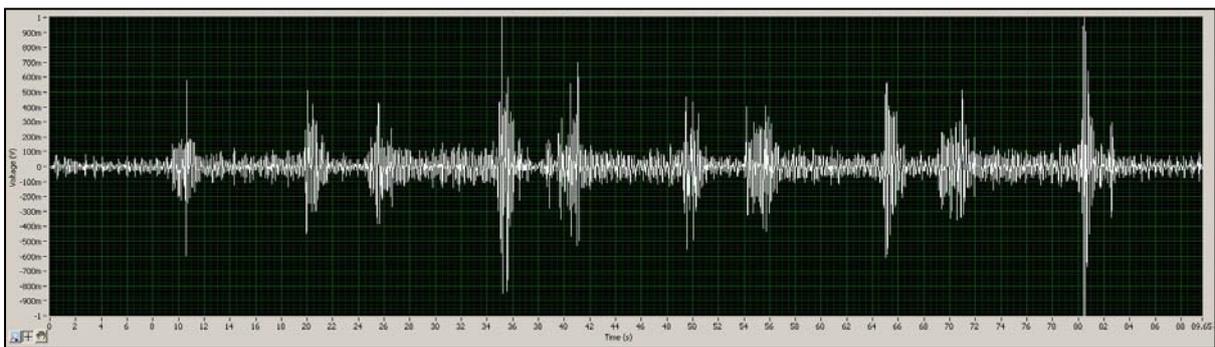


Figure 5.4.7 – Test 4b

The aim of test 4 was to determine whether or not, consistent measurements can be taken using the sensor. As shown, the results are relatively inconsistent. The major abnormalities occur in both the first and final contractions; the second, third and fourth contractions seem relatively consistent. The first contraction seems to be considerably smaller than the others, in both the weighted and un-weighted tests. It is unclear as to the reason for the amplitude abnormality.

Contraction 5 is consistent in test 4b with the middle 3 contraction, however in 4a the signal is slightly larger; the most likely reason for this interference from an external source such as vehicles, people talking, footsteps, etc.

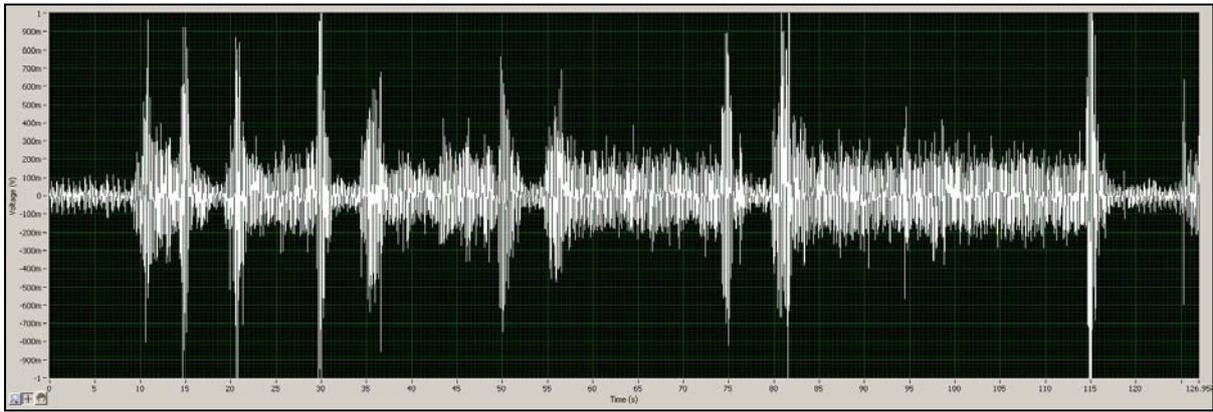


Figure 5.4.8 – Test 5a

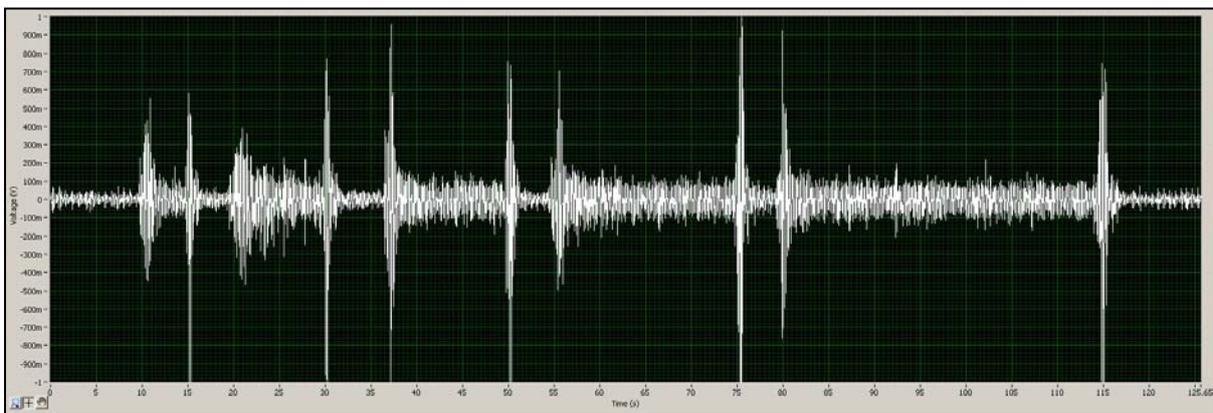


Figure 5.4.9 – Test 5b

Test 5a and 5b clearly demonstrates the difference between tests with mass and without mass. The signals amplitudes clearly get larger with the increase in force. The average peaks in the weighted test is $\pm 215\text{mV}$ compared to the un-weighted result approximately $\pm 120\text{mV}$.

As the first stage of the test illustrates, the 5 second contraction lacks clarity because the sensor is not allowed enough time to stabilize from its initial displacement. Most evident is test 5a, where the signal remains high until the last 2 seconds of the contraction before it begins the extension. Additionally, there is an unusually longer settling time for the final artefact of the first contraction.

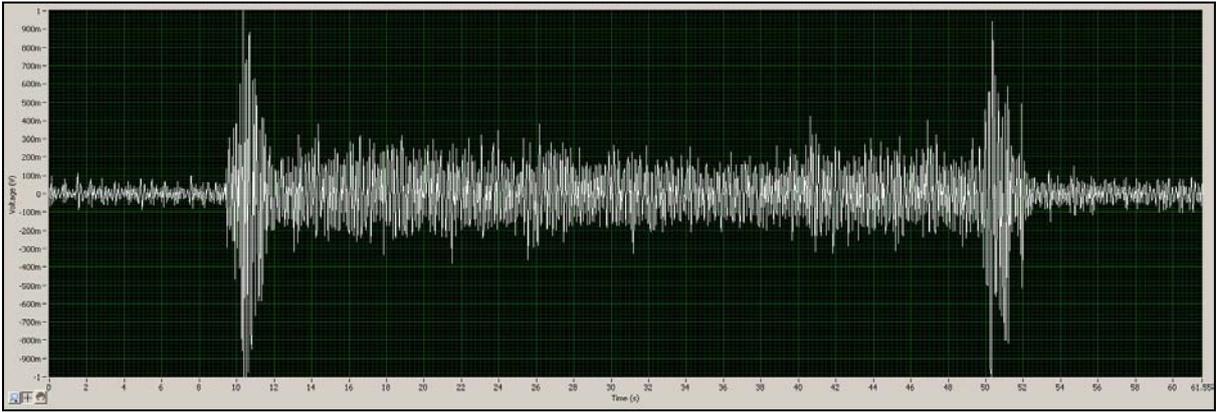


Figure 5.4.10 – Test 6a

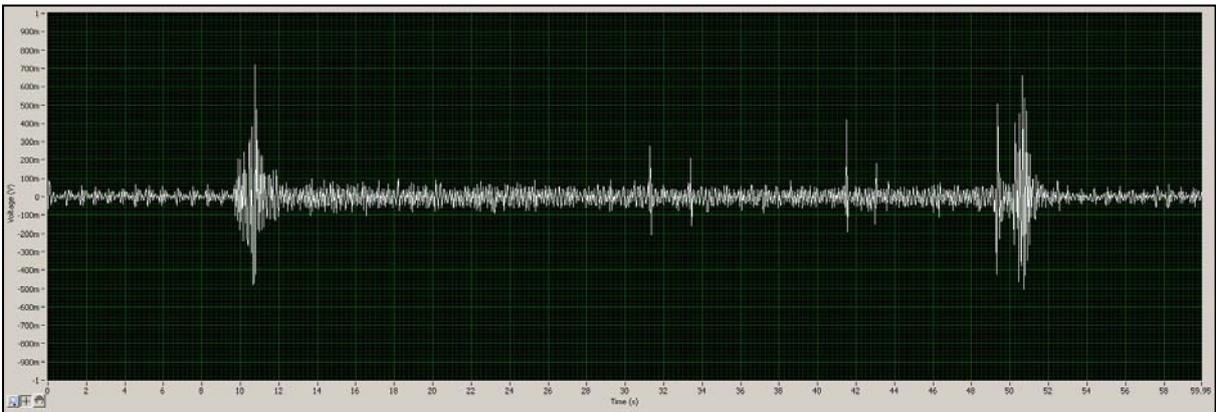


Figure 5.4.11 – Test 6b

Again, similar to test 3b and 4b, the reduced mass makes it difficult to distinguish between activation and ambience. In an application where the sensor was used on a person performing light work, the sensor would not work effectively.

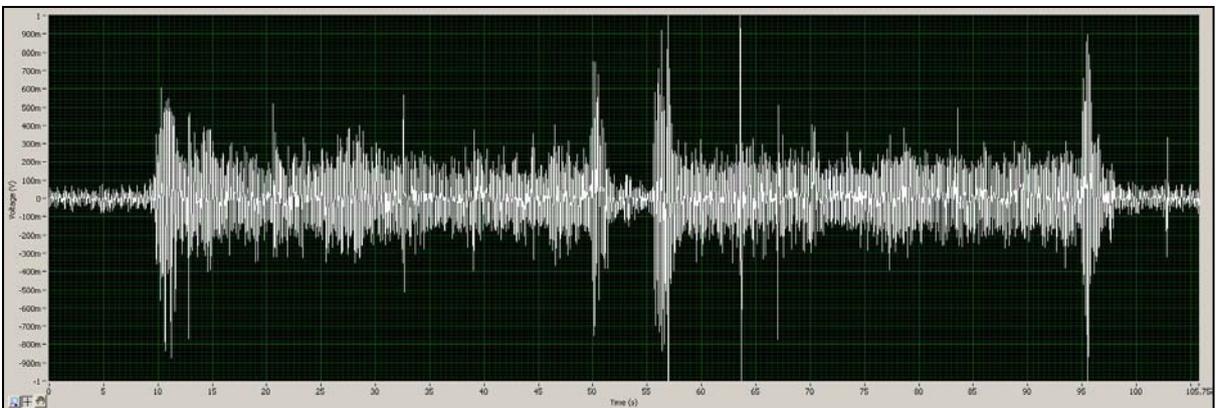


Figure 5.4.12 – Test 7a

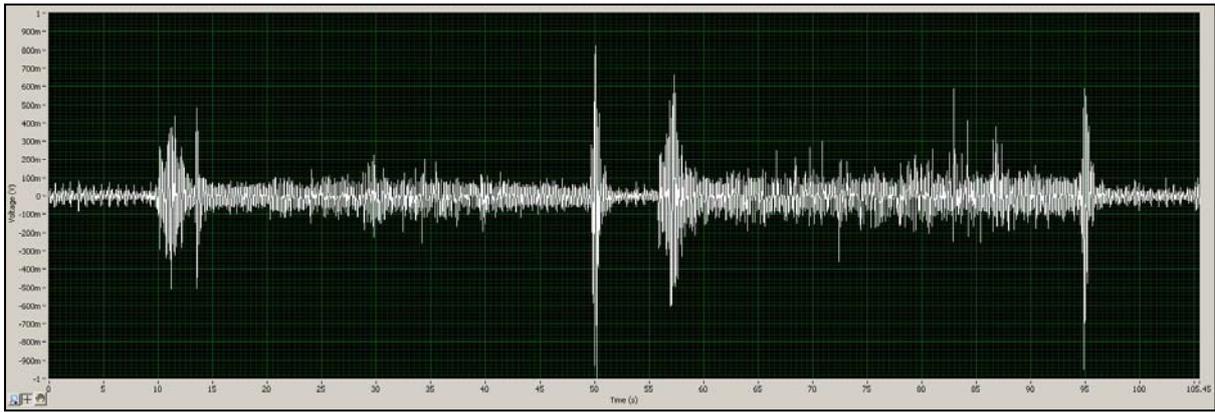


Figure 5.4.13 – Test 7b

5.4.2 Squeeze Test Results (Flexor Digitorum)

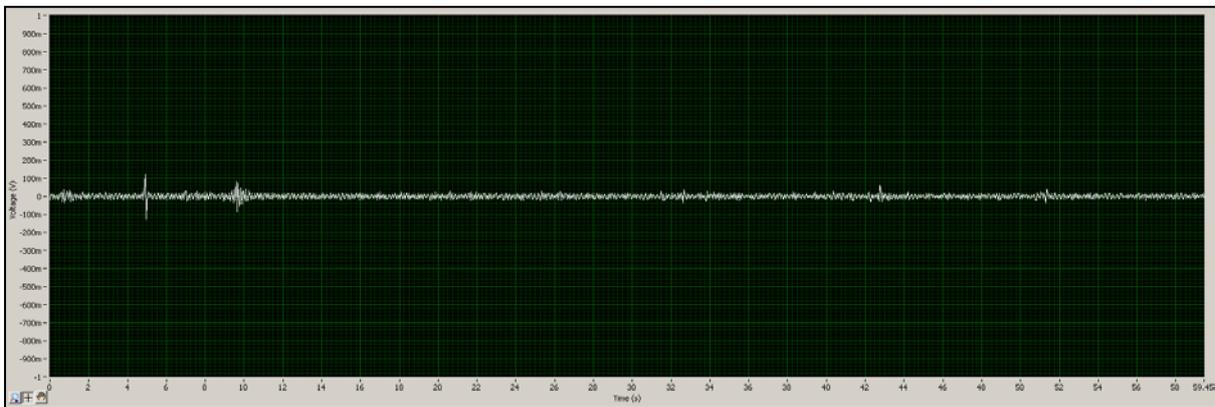


Figure 5.4.14 – Test 2c

Comparatively, the vascular activity in the Flexor Digitorum is far less than that of the Bicep Brachii; no vascular activity was observable during the squeeze tests. The blood flow is not purposefully restricted in anyway. The most probable reason for this is there simply aren't any blood vessels that are located close enough to the surface of the skin in that particular region. If that is the case, more focus would have to be given to the placement of the sensor, to allow for optimal measuring conditions.

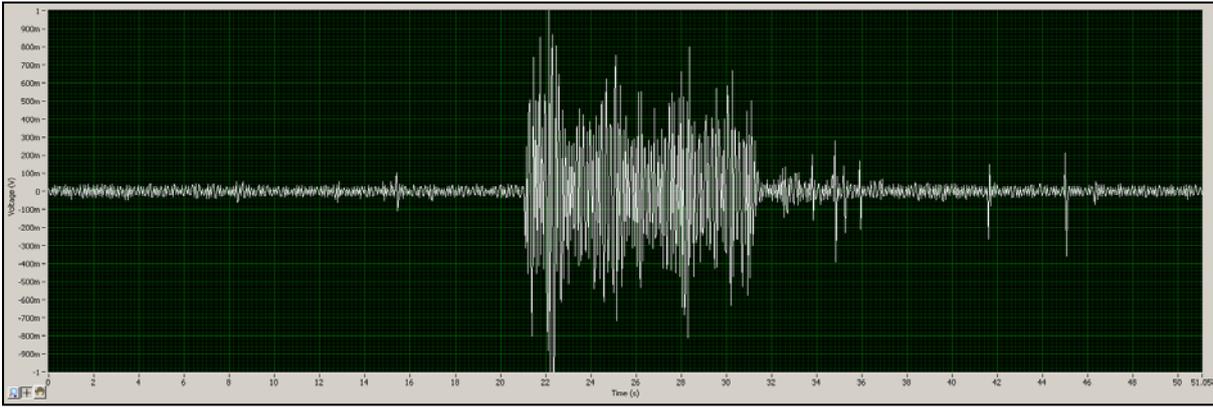


Figure 5.4.15 – Test 3c

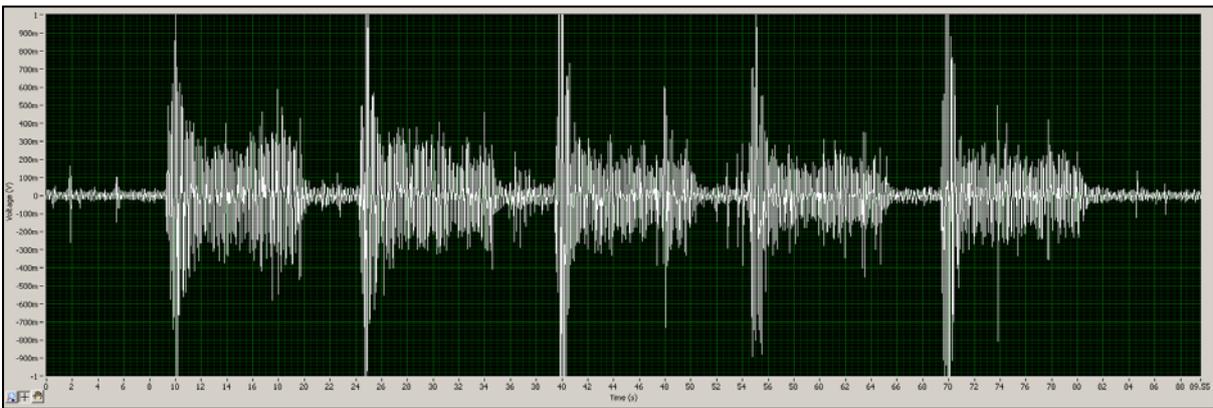


Figure 5.4.16 – Test 4c

Shown clearly in Figure 5.4.16, the muscle deformation artefacts are shorter and occur only in the beginning of the contraction. However, the amplitude of the artefact is far stronger than that of the lift tests.

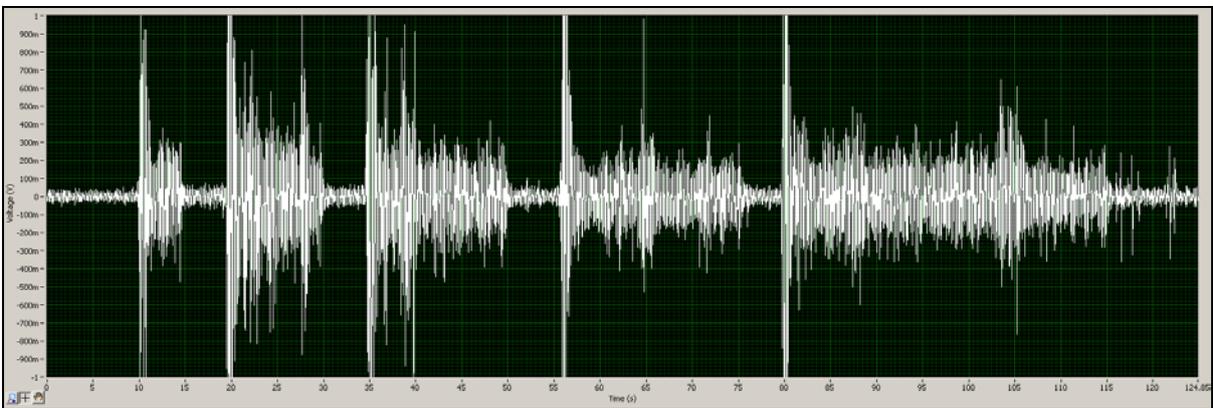


Figure 5.4.17 – Test 5c

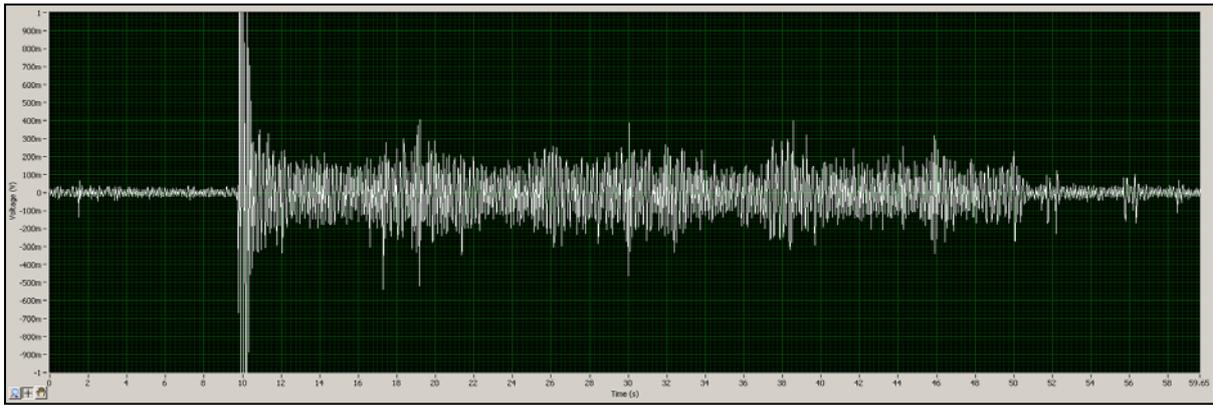


Figure 5.4.18 – Test 6c

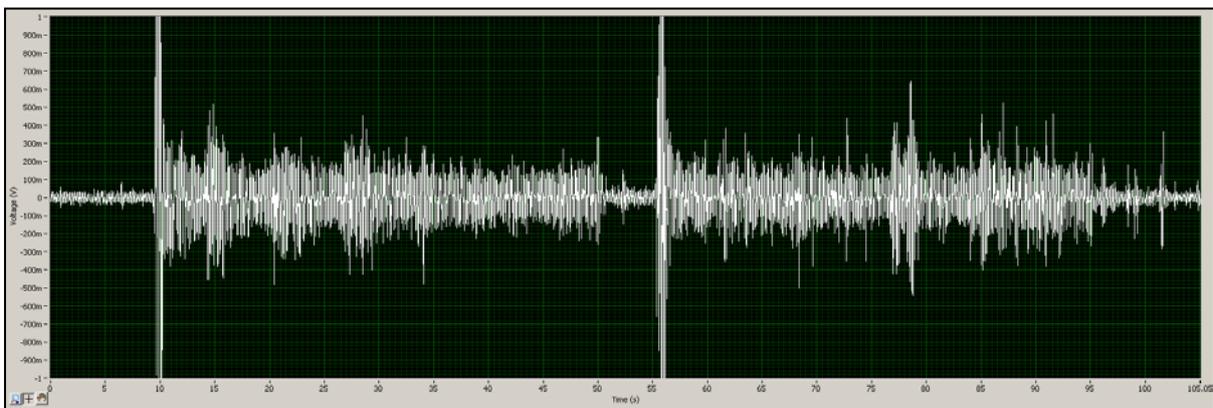


Figure 5.4.19 – Test 7c

Comparing the first test 3c (Figure 5.4.15) to test 7c (Figure 5.4.19), there is a clear decrease in amplitude which could indicate muscle fatigue. However, as shown in the lift tests the consistency of the sensor is very low so there is no conclusive evidence that the drop in signal amplitude is caused by muscle fatigue.

5.4.3 Focused Frequency Testing

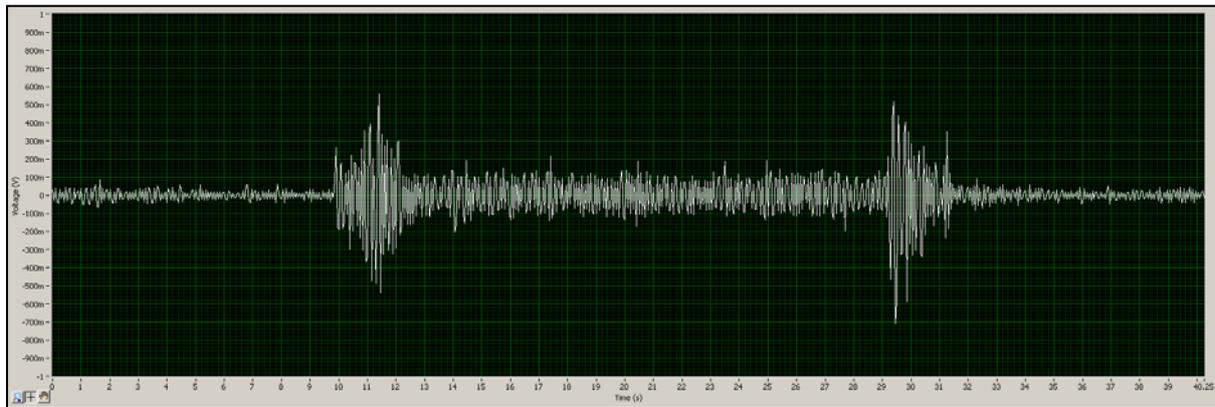


Figure 5.4.20 – Focused Frequency Test (5 – 100Hz)

Figure 5.4.20 shows a default filtered signal. The main goal of the focused frequency test is to decrease the ambient signal, 0:00 – 0:10 and 0:30 – 0:40; furthermore, to decrease the amplitudes of the muscle deformation artefacts (~0:10 – 0:12 and ~0:30 – 0:32). The ambient and artefact noise occur over a wide frequency range; for this reason fully removing the interference will be difficult and unlikely, the only course of action is to minimize its effects as much as possible.

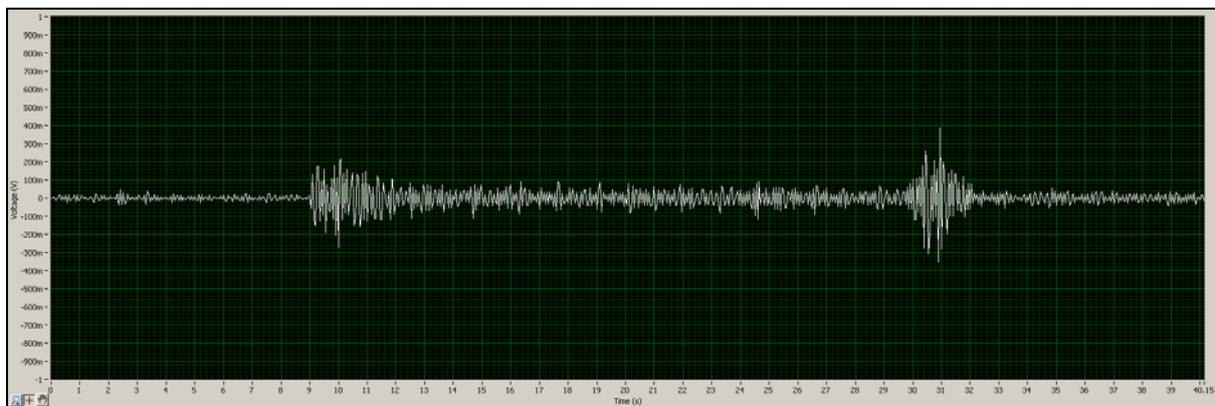


Figure 5.4.21 – Focused Frequency Test (5 – 20Hz)

Figure 5.4.21 shows a decrease in both the ambient and artefact amplitude; however there is also a significant drop in the contraction signal. This means a considerable portion of the ambient and artefact is being filtered out. However the contraction vibrations are also being filtered out.

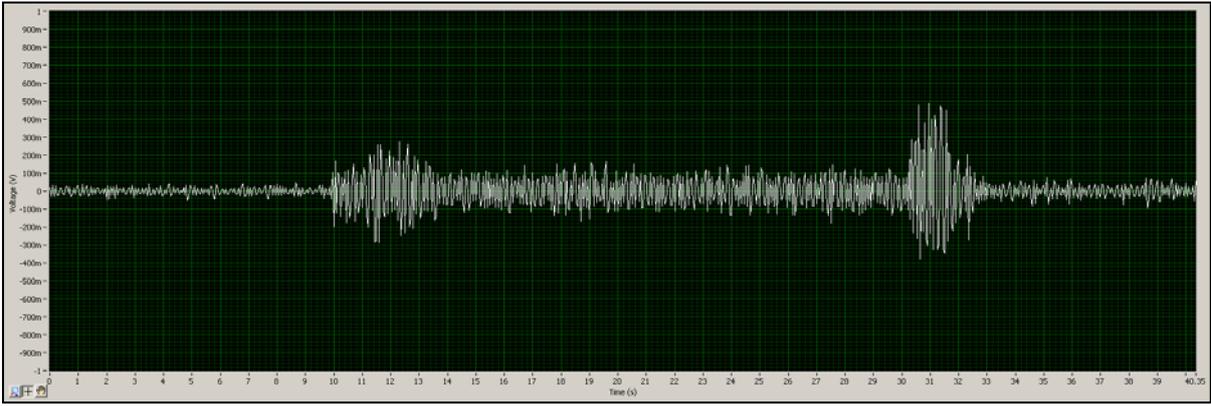


Figure 5.4.22 – Focused Frequency Test (5 – 50Hz)

As Figure 5.4.22 shows, the muscle vibration is becoming more prominent, meaning the filters pass band region is including more of the specific muscle vibration frequencies of this particular user.

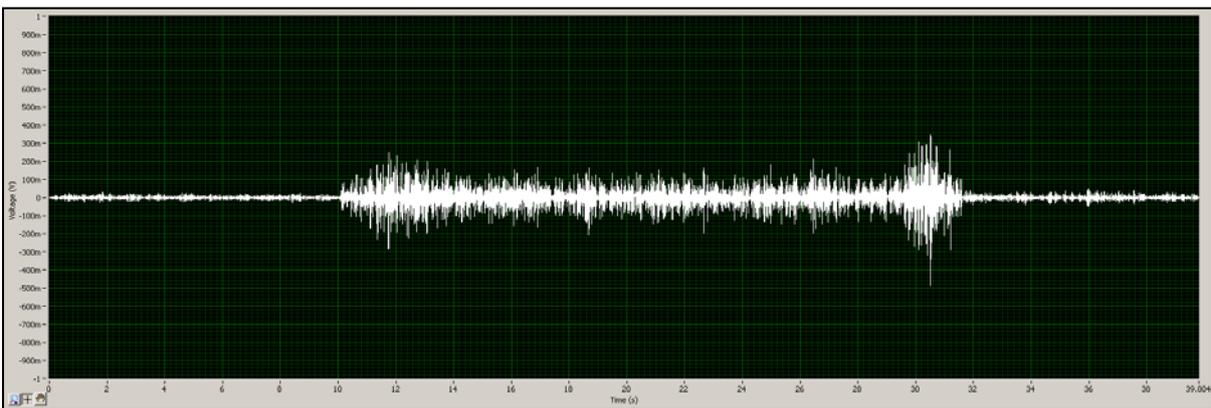


Figure 5.4.23 – Focused Frequency Test (20 – 50Hz)

Figure 5.4.23 has a similar form to that of Figure 5.4.22; there is a slight reduction on both the artefact and the ambient zones. Since Figure 5.4.23 has had its lower frequency component removed, the majority of the graph will appear to be “denser” this is meaningless.

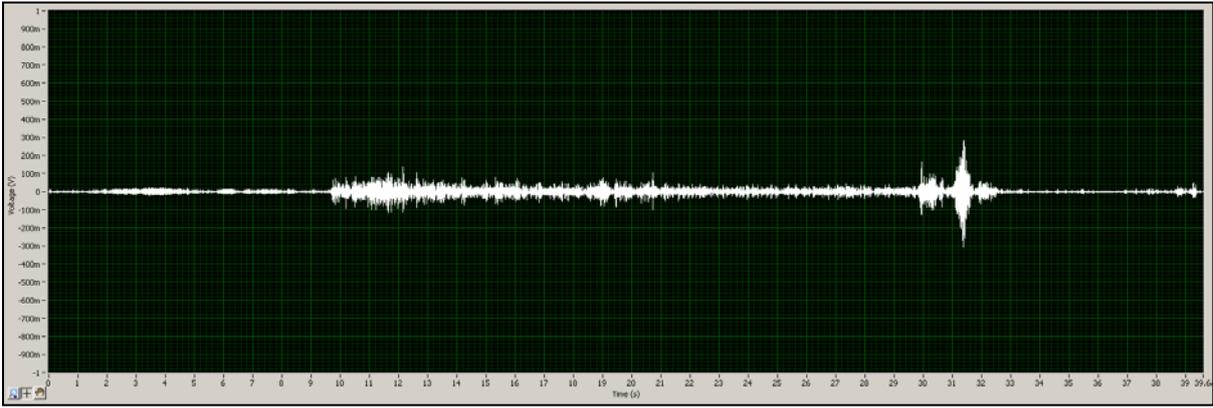


Figure 5.4.24 – Focused Frequency Test (50 – 100Hz).

Figure 5.4.24 shows clearly that between 50 and 100Hz there is a significant decrease in signal amplitudes; a clear indication that the significant muscle activity is occurring outside of the range of the filter's frequency response range.

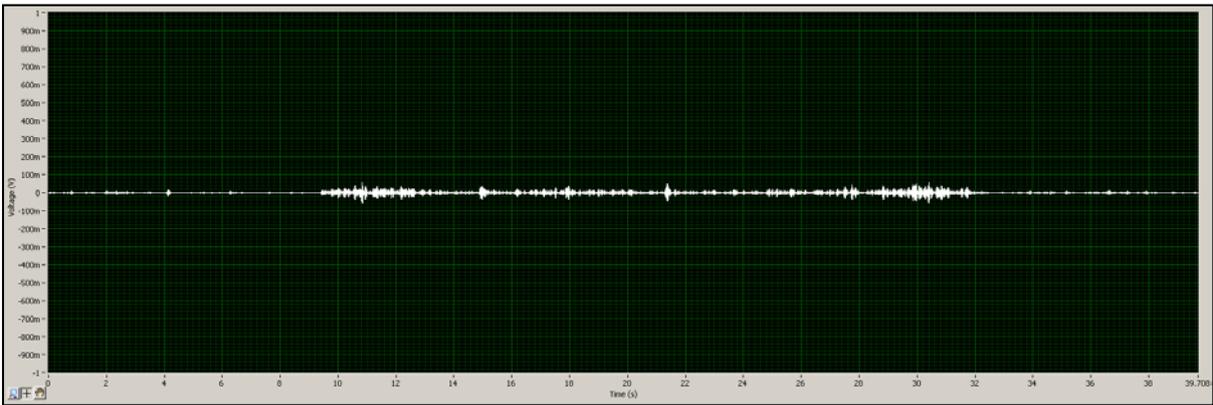


Figure 5.4.25 – Focused Frequency Test (80 – 100Hz)

Figure 5.4.25 is a clear indication that muscle vibration little to no vibration occur higher than 80Hz.

5.4.4 Additional Tests

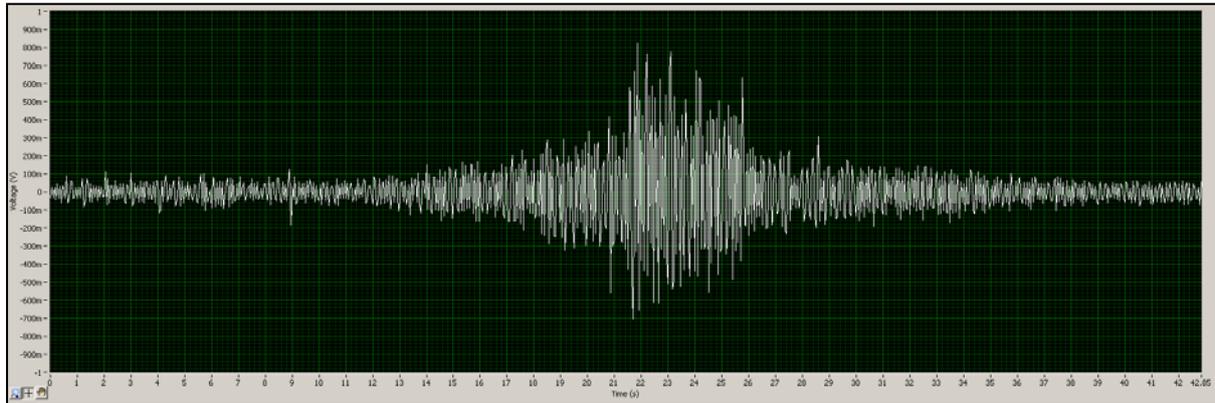


Figure 5.4.26 – Increasing Mass Test

Perhaps the best demonstration of the relationship between muscle force output and the muscle vibration output is shown by Figure 5.4.26.

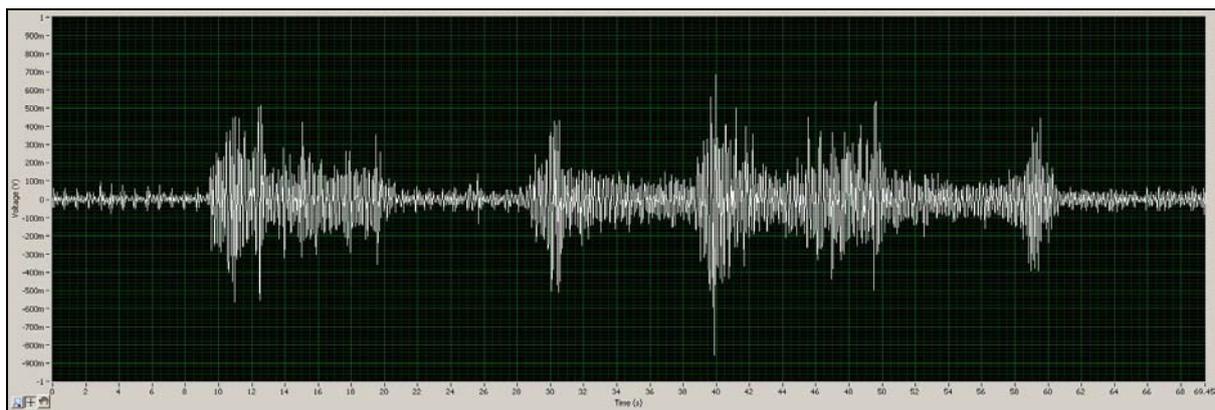


Figure 5.4.27 – Muscle Interaction Test

The muscle interaction test is a test to determine the affect a muscle contraction has on the sensor while in use on a separate muscle. In Figure 5.4.27, it is clear where each stage was started, however, this is only made possible by the unwanted deformation artefacts. The ultimate goal would be the measurement of muscle activity with minimal inclusion of the muscle deformation artefacts. Currently, without the muscle deformation artefacts it would be very difficult to determine which contraction was which. An interesting point of interest is during the first and second squeeze test, the artefact is much lower in relation to the main body of the contraction. The reason for this is because the sensor is not directly attached to the

muscle there is no direct muscle deformation caused by the Flexor Digitorum but because the Flexor Digitorum's contraction is partially dependent on the Bicep Brachii there is still a slight deformation artefact. Compare the ratio between artefact main body, the two Bicep contraction are far larger in relation to the main body and as shown during the lift and squeeze tests the opposite is true.

Unusual Signal Artefacts

One of the major opposing arguments to the validity of MMG as a method of muscle activity detection is that "muscle sound" is simply caused by the interaction between the surface of the skin and the sensor itself. This has been disproven by isolating the muscle and by adding an interfacing medium, water. Unfortunately, for any MMG sensor that isn't using a water medium between the skin and sensor, the skin/sensor problem remains; even though it is possible to minimize it, it cannot be fully eliminated. The large fluctuation seen at the beginning of the contraction maybe attributed to the deformation of the muscle causing large amounts of friction between the sensor and skin. Once the muscle dimension settles, during the sustained portion of the contraction the measurements react similar to that seen in previous research.

There are 2 clearly different templates in regards to the large muscle deformation artefacts, perhaps more. In the lift test, there is both an artefact in the beginning and the end of the contraction whereas the squeeze test only has one in the beginning. This is related to the nature of the Biceps Brachia's contraction. During lift or flexion stage of the contraction, in addition to the standard muscle deformation, the Bicep will shift its position. During the lowering or extension stage the Bicep must shift back into its original position causing the second artefact. In a situation such as the squeeze test where the targeted muscle is the Flexor Digitorum, the second artefact is not seen since the Flexor Digitorum only expands during contraction and doesn't change position.

In Gordon & Hobourn's research, in 1948, similar muscle deformation artefacts are seen. However, the objectives of their research allowed them to simply disregard the artefacts and focus on the main body of the contraction. Similarly, in a lot of other research designed specifically in the study of MMG signals, many researchers will

simply disregard the beginning and end sections of the signal because of the possibility of interference.

One of the major benefits that were considered when deciding on what technology to utilise for this sensor was the fact that microphone MMG sensor would not need to filter out regular muscle actions associated with the activation of a muscle such as muscle deformation and muscle movement. However, testing has shown that without an appropriate fastening method, all the regular muscle actions still need to be considered during signal processing.

Vascular Pulse

One major undesired component of the sensor is its ability to detect vascular pulse. Similar to muscle vibrations, vascular pulses transmit equally well through muscle tissue and will interfere with the performance of the sensor. In particular, smaller muscle or forces which produce smaller signal amplitude can be severely affected; larger signals will simply over power the interference. However, as shown in the testing, between 0 – 19.62N (2kg), still only produces amplitudes that are about 50 – 100mV above that of the baseline pulse amplitudes. 0 – 19.62N is what would be considered light to moderate levels of work; there would rarely be any regular and prolong work requiring muscle to output more than 19.62N.

There are 2 practical methods of removing the vascular pulse data from the signal. The first is to use 2 sensors, one placed on the muscle for regular detection and another placed on vascular pulse point independent of the muscle. By using the differential inputs on the DAQ, it should be possible to remove the pulse signal from the muscle signal. It is important that the pulse sensor is not near the muscle lest it also removes important muscle vibrations.

The second method is to simply restrict blood flow. Smith et al. (1998) and Gordon & Holbourn (1948) discussed issues relating to vascular pulse, Gordon & Holbourn performed experimentation with the objective of determining the relationship between muscle vibration and vascular sounds. During my preliminary testing, while testing methods of fastening, the results showed that there was considerably less vascular noise detected by the sensor when the sensor was strapped to the arm

tightly. A possible reason for this is reduction in noise is due to the straps of the sensor restricting the amount of blood flow through that section of the muscle. Furthermore, the edges of the sensor press down with considerable force to restrict blood flow even more which produces a “dead zone” where the sensor membrane is contacting the surface of the skin which minimizes the interference caused by vascular sounds.

Muscle Fatigue

In a lot of literature reviewed over the course of this research, many MMG articles have highlighted the unique ability for the sensor to detect muscle fatigue. Unfortunately, throughout the course of the testing, no evidence of muscle fatigue was detected. Any promising indication of muscle fatigue cannot be confirmed when compared with other tests from the same test set.

In the majority of previous research done in regards to muscle fatigue, significant signal degradation can be seen within 10 seconds (Barry, Geiringer & Ball, 1985; Gordon & Holbourn, 1948); no such event occurred, even with much longer contraction times of 30 seconds and more.

Despite the observation of previously documented evidence of muscle fatigue, throughout the squeeze tests, the observed diminishing amplitudes could indicate muscle fatigue. Because the variability of the signal between tests is quite high, caused by variables such as wait times between tests, times a test was repeated, etc; it is difficult to definitively say that it is muscle fatigue, the general trend does suggest the possibility of muscle fatigue.

Filtering

In order to produce a much sharper and more precise cut-off, trials were carried out using varying orders of filters. Even though a super high order filter would have been possible with LabView™, the decision was made to keep the filter a 5th order to replicate the performance hardware filter. Additionally, the super high order filters have very long settling times (refer to Appendix 8.3); the longer settling times would show a false representation of muscle fatigue during testing. Not only did the higher

order over tax the computers processing capabilities but it also made the entire system prone to crashing, coupled with the high sampling rate and averaging of the various data display blocks rendered SignalExpress and the computer almost unusable. A higher order filter also adds a significant time delay to the real-time output of the filter. The test results are all recorded data thus is not affected by the filters time delay; however in real-time the filter response is significantly lowered. This is caused by the high complexity of the filter transfer function; the computer is unable to process and output the data quickly enough, the delay is quite significant at orders past 20.

Even though a hardware filter is implemented, the hardware filter is only able to act as a general filter. Even though the signal was relatively clear and muscle activation could be clearly detected, the implemented hardware filter seemed to have been inadequate. Signal clarity improved significantly whilst using the software filter. The software filter provided minimal phase shift, minimal signal degradation and allowed immediate frequency range adjustments as needed.

For more precise and accurate measurements, I would recommend still implementing the software filter to allow the operator to more accurately bracket the desired muscle activity frequency range. This would only be possible if the sensor was used in conjunction with a computer. In an embedded system, where a variable software filter may not be practical and higher precision is needed, a hardware filter may need to be custom built for a particular user. Additionally, as suggested by various other sources, a high pass filter can be added to further clean up the signal.

Further Tests

More testing can be done to provide a greater understanding of the capabilities of the sensor. Some of these tests were not possible or simply wasn't sufficient time or equipment to complete.

- *EMG comparison test*

In addition to the timing tests, the initial plan was to do a comparative test with an EMG running concurrently. This will allow the both time delay and force levels to be compared.

- *Fine Motor Control*

This test would examine the muscle activity of smaller muscle such as the finger or facial muscles.

- *Sensor Placement Test*

As discussed previously, MMGs are not as significantly affected by position and location as EMGs. The proposed test would evaluate the actual affect position has on the MMG sensor. This test would need to be completed using multiple sensors, set up into an array each at different points of the muscle measuring simultaneously. This test was planned but later abandoned because there was only one sensor available and its sensitivity to noise and interference would not enable any objective comparisons to be completed.

5.5 Limitations

The size and weight of the entire sensor posed significant problems. As discussed previously, the most effective method of fastening the sensor to the muscle is to use adhesive to attach the sensor directly on to the surface of the skin. However, the size and weight of the sensor made this impractical. By using straps it introduced a considerable influence on the measurements and jeopardizes the validity of the results.

The sensor was definitely susceptible to external interference, in particular movement. Since the sensor is not completely secured to the surface of the skin, the shifting momentum of the sensor will cause undesired vibration. Because of this problem, the user of the sensor has to remain as still as possible. Furthermore, the sealed air chamber added to increase its sensitivity also increases its sensitivity of external noise.

The sensor testing lacks a comparative aspect, the initial intention was either to compare results to an EMG or a professional built MMG sensor. This would allow a more effective measure of the sensors performance. Unfortunately, budget and time constraints prevented effective comparative tests to be conducted.

Chapter 6 Conclusion

The objective of this thesis is to produce a working prototype of a MMG based muscle activity detection sensor. The sensor consisted of a low frequency electret microphone, microphone preamplifier and conical sealed air volume to provide a better mechanical response for the microphone. Outputs from the sensor were captured using a National instruments data acquisition card and National Instruments LabView™ software was used in conjunction.

The testing of the MMG sensor showed limited success. On first impressions, the results yielded showed appropriate measurements taken by the microphone in relation to previous research done. Frequencies for muscle vibrations during testing did occur within 5 – 100Hz range stated by Courteville et al., (1998), Oster and Jaffe, (1980) and Orizio et al., (1989). There was an exponential relationship between the force and signal as discussed by Orizio et al., (1989). However, results did not showed decrease in signal amplitude due to muscle fatigue, in agreement of research done by Courtville et al., (1998) and Barry et al., (1985), although there was evidence that indicated it as a possibility the evidence was not conclusive. Furthermore, even though the sensor was able to yield good results, the sensor underperformed a combination of both the skin and sensor interference issue and the method of fastening were both major factors in its underperformance.

In spite of the issues surrounding testing results, the key objectives of building a basic low cost sensor for muscle activity detection has been fulfilled. Major myographic methods and techniques were explored and a wide range of sensor designs from previous research were explored. A working prototype of a muscle activity detection sensor was built using knowledge gained from the exploration. Through a series of tests, some, though not all, unique muscle events were observed.

Although MMGs may never be as accurate an EMG, this study has demonstrated the cost effectiveness and potential of utilising a cheap audio transducer for the purposes of muscle activity detection. Additionally, as a consequence of this study has shown the need for further research and development. The potential applications where MMGs can be implemented in would perhaps surpass that of EMGs. With

more time and more research, we believe it can be fine tuned to provide an effective sensor for the detection of muscle activity.

6.1 Recommendations

In my opinion the MK250 should still provide the greatest performance despite the cost. The major focus in the future should be focussed on the development of the MK250 version of the sensor since microphones are only going to get cheaper as new technologies and manufacturing methods improve.

The next stage of development would be to implement a system to convert the raw sensor output into a control signal; this could be done either using hardware in the form of a rectifier or using software by developing a program to calculate RMS values.

Future research could also look into applying this sensor to an application.

Applications

- *Actuator or device control*

The obvious next step for developing a sensor would be to apply it to an application. Actuation controls, whereby the signal from the sensor can be used as a control signal for an actuator. This could open the door to further development of assistive devices or remote control applications.

- *Assistive devices*

A lot of research has focussed on the development of assistive devices; MMG sensor's size and robustness are all attributes well suited for this application. However, as discussed in section 3.3, MMG sensors do have limitations on the assistive applications it is able to be implemented in, for example MMGs are ill suited for situations involving amputees. Since MMGs rely on the physical response of the muscle to determine its activation, without the actual muscle MMGs are useless.

- *Medical myography*

MMGs are rarely used in medical situations since its most advantageous attributes (robustness, cost and size) are made pointless in lab environments.

Based on accuracy and precision, MMGs do not even come close to comparing to an EMG. Firstly, improvement would be to improve the sensors accuracy and clarity to make it a more viable alternative to using an EMG.

Additionally, improvements can be made to current sensor.

Improvements

- *Sensor arrays*

The sensor array would be the next step in developing an assistive device. In order to mimic human actions, multiple sensors would be needed to detect the multiple muscle activations associated with basic tasks such as opening or closing your fingers to pick something up.

- *PC heavy signal processing*

As discussed previously, a more PC orientated signal processing would allow for a lot better signal quality but far less portability. This could include things such as signal converter to convert the less useful AC signal into a more useful DC control signal. Additionally, applying more complex filtering, to remove things such as muscle deformation artefacts, vascular interference and any other non muscle related components.

- *Different transducers*

Section 3.3 discussed alternatives to using different transducers. Even though microphones are easy to implement and the raw signal sensor signal is easier to interpret without signal processing, small signals like finger muscle are difficult to detect because the smaller muscle vibrations generate a much lower SPL which the microphone sensor is not able to detect. Other transducers such as the accelerometer and piezoelectric contact sensors may provide better accuracy and sensitivity. A more complex signal processing routine would be needed.

- *More effective mounting/fastening system*

Due to the overall size of the sensor, straps are needed to secure the device to the skin surface; this is not ideal. Straps or other external fastenings have the potential to interfere with the operation of the sensor. Ideally, the sensor should be small and light enough to be attached via adhesive. Being small

also provides the benefit of not interfering with the users movements. The largest obstacle is removing/replacing the microphone cone, either replacing it with a smaller but similar device or removing it completely and somehow finding a microphone able to detect a lower SPL.

- *Water Coupling*

It has been shown that the direct contact between the sensor and the surface of the skin creates problems. Even though you can decrease the influence by improving the method of fastening, it cannot be fully eliminated. Other research has shown that by using water as an intermediary between the skin and sensor you can completely eliminate the problem.

Chapter 7 References

- Ballou, G. (2008). *Handbook for Sound Engineers*. Burlington: Elsevier Inc.
- Barry, D. T., & Cole, N. M. (1988). Fluid mechanics of muscle vibrations. *Biophysical Journal* , 899-905.
- Barry, D. T., & Cole, N. M. (1988). Fluid Mechanics of Muscle Vibrations. *Biophysical Journal* , 899-905.
- Barry, D. T., Geiringer, S. R., & Ball, R. D. (1985). Acoustic myography: A noninvasive monitor of motor unit fatigue. *Muscle & Nerve* , 189-194.
- Borwick, J. (1990). *Microphones: Technology and Technique*. London: Focal Press.
- Brozovich, F. V., & Pollack, G. H. (1983). Muscle contraction generates discrete sound bursts. *Biophysical Journal* , 35-40.
- Bu, N., Tsukamoto, J., Ueno, N., Shima, K., & Tsuji, T. (2008). Measuring muscle movements for human interface using a flexible piezoelectric thin film sensor. *30th Annual International IEEE EMBS Conference* (pp. 112-116). Vancouver: IEEE.
- Courteville, A., Gharbi, T., & Cornu, J.-Y. (1998). MMG Measurement: A high-sensitivity microphone based sensor for clinical use. *IEEE Transactions on Biomedical Engineering* , 145-150.
- Gordon, G., & Holbourn, H. S. (1948). The sounds from single motor units in a contracting muscle. *J. Physiol* , 456-464.
- Huber, D. M. (1988). *Microphone Manual: Design and Application*. Stoneham: Focal Press.
- Jaskolska, A., Madeleine, P., Jaskolski, A., Kisiel-Sajewicz, K., & Arendt-Nielsen, L. (2007). A comparison between mechanomyography condenser microphone and

accelerometer measurements during submaximal isometric, concentric and eccentric contractions. *Journal of Electromyography and Kinesiology* , 336-347.

Mercer, J. A., Bezodis, N., DeLion, D., Zachry, T., & Rubley, M. D. (2006). EMG sensor location: Does it influence the ability to detect differences in muscle contraction conditions? *Journal of Electromyography and Kinesiology* , 198-204.

Merletti, R., & Parker, P. (2004). *Electromyography: Physiology, Engineering and Non-Invasive Applications*. Hoboken: Wiley-IEEE Press.

Mita, K., Watakabe, M., Akataki, K., & Itoh, Y. (2001). Mechanical behavior of condenser microphone in mechanomyography. *Medical & Biological Engineering & Computing* , 195-201.

Morgan, D. (2007). *Surface Acoustics Wave Filters*. London: Elsevier Inc.

Murphy, C., Campbell, N., Caulfield, B., Ward, T., & Deegan, C. (2008). Micro electro mechanical systems based sensor for mechanomyography. *Biosignal* , 1-10.

Nigg, B. M., & Herzog, W. (2007). *Biomechanics of the Musculo-skeletal System*. West Sussex: John Wiley & Sons Ltd.

Norman, E., Cubitt, J., Urry, S., & Whittaker, M. (2000). *Advanced Design and Technology*. Essex: Pearson Education Ltd.

Oster, G., & Jaffe, J. S. (1980). Low frequency sounds from sustained contraction of human skeletal muscle. *Biophysical Journal* , 119-128.

Pina, E., Burgos, E., Prados, C., Gonzalez, J. M., Hernando, A., Iglesias, M. C., et al. (2001). Magnetoleastic sensor as a probe for muscular activity. *Sensors and Actuators* , 99-102.

Self, D., Bruce, R., Duncan, B., Hood, J. L., Sinclair, I., Singmin, A., et al. (2009). *Audio Engineering: Know It All*. Burlington: Elsevier Ltd.

Shonhara, M., & Sogaard, K. (2006). Mechanomyography for studying force fluctuations and muscle fatigue. *Exercise and Sports Science Review* , 299-305.

Smith, D. B., Housh, T. J., Johnson, G. O., Evetovich, T. K., Ebersole, K. T., & Perry, S. R. (1998, December 7). Muscle and Nerve. *Mechanomyographic and electromyographic responses to eccentric and concentric isokinetic muscle actions of the biceps brachii* , pp. 1438-1444.

Tanaka, T., Hori, S., & Yamaguchi, R. (2003). Ultrasonic sensor disk for detecting muscular force. *IEEE International workshop on Robot and Human Interactive Communication* (pp. 291-295). Millbrae: IEEE.

Tarata, M. Y. (2003). Mechanomyography vs. electromyography in monitoring the muscular fatigue. *Biomedical Engineering Online* .

Trager, G., Michaud, G., Deschamp, S., & Hemmerling, T. M. (2006). Comparison of phonomyography, kinemyography and mechanomyography for neuromuscular monitoring. *Canadian Journal of Anesthesia* , 130-135.

Tsutsui, Y., Tanaka, T., & Shun'ichi, K. (2005). Duplex Ultrasonic Muscle Activity Sensor. *IEEE Xplore* , 310-313.

Watakabe, M., Itoh, Y., Mita, K., & Akataki, K. (1998). Technical Aspects of mechanomyography recording with piezoelectric contract sensor. *Medical & Biological Engineering & Computing* , 557-561.

Wong, G. S., & Bembleton, T. F. (1995). *AIP Handbook of Condenser Microphones*. London: American Institute of Physics.

Chapter 8 Appendices

8.1 Enlarged Results Graphs

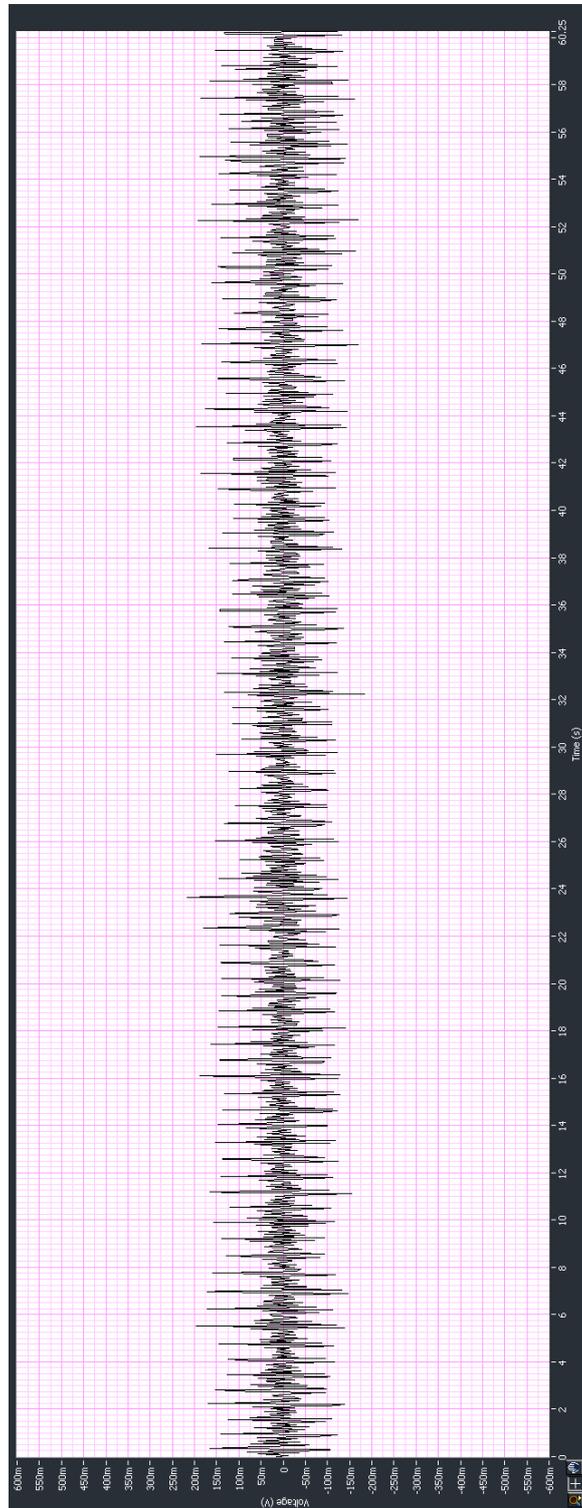


Figure 8.1.1 – Test 2a

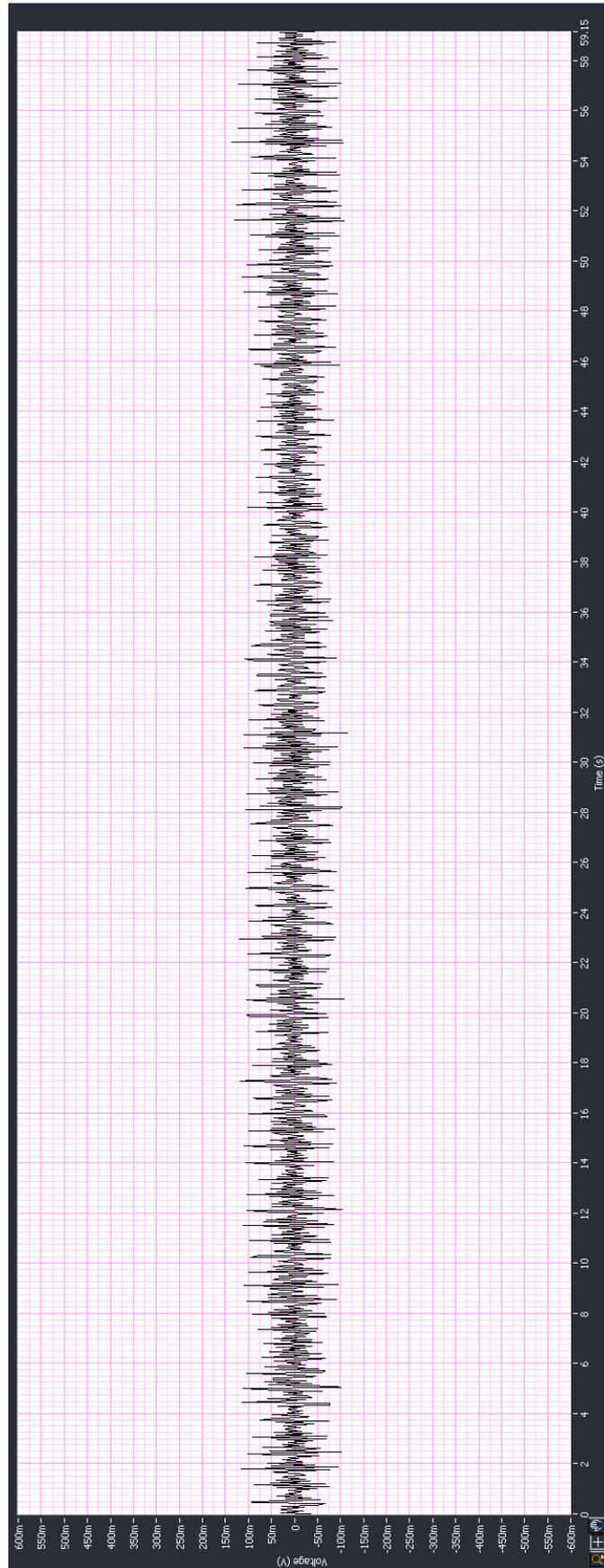


Figure 8.1.2 – Test 2b

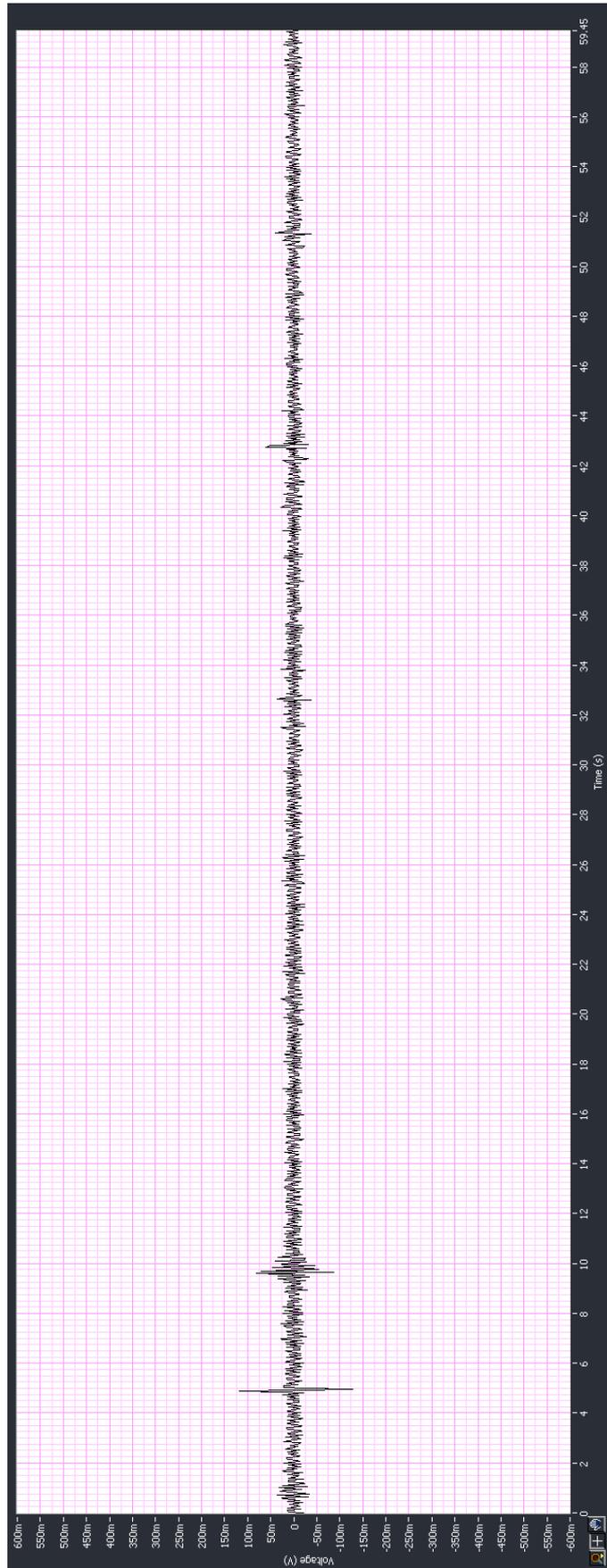


Figure 8.1.3 – Test 2c

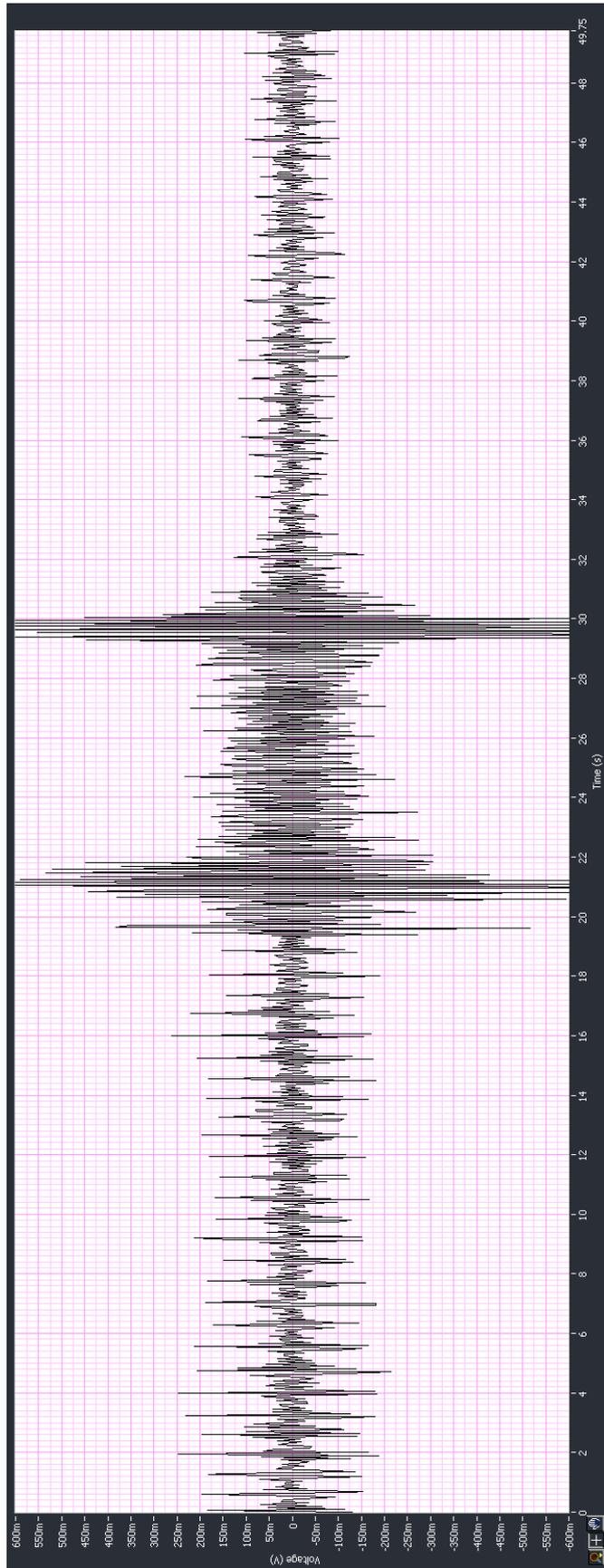


Figure 8.1.4 – Test 3a

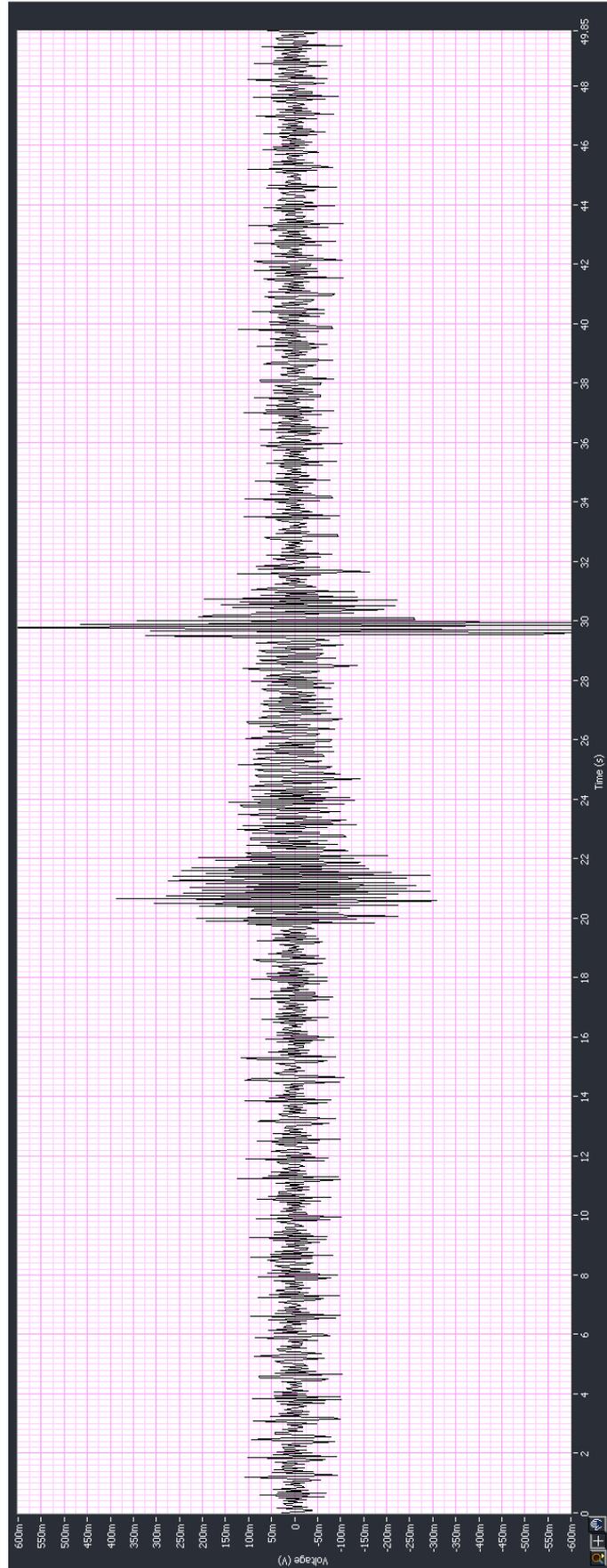


Figure 8.1.5 – Test 3b

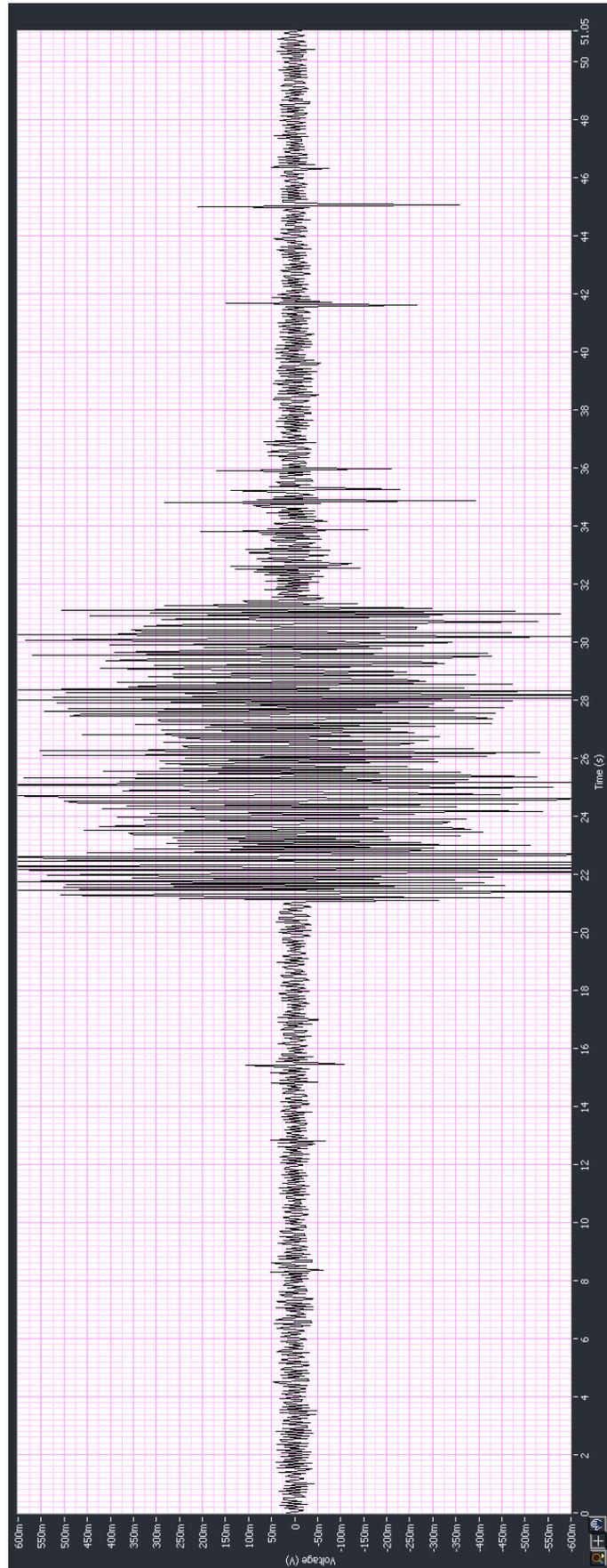


Figure 8.1.6 – Test 3c

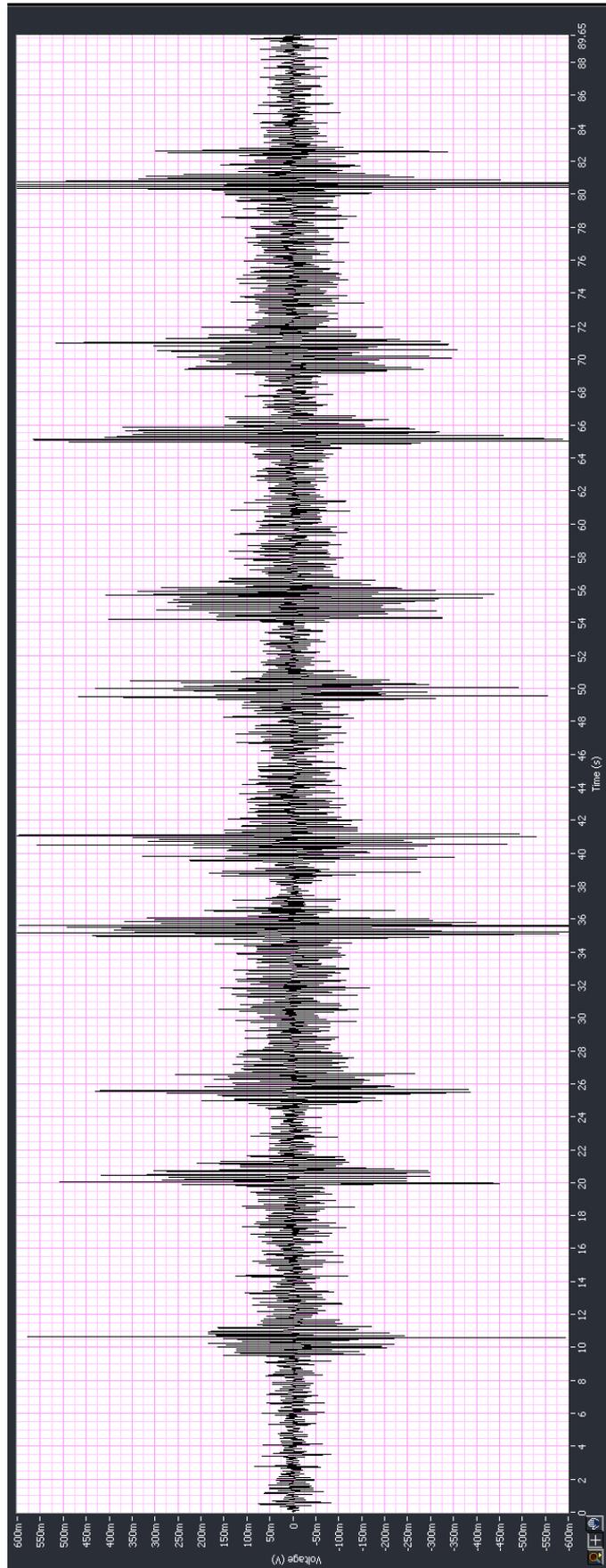


Figure 8.1.7 – Test 4a

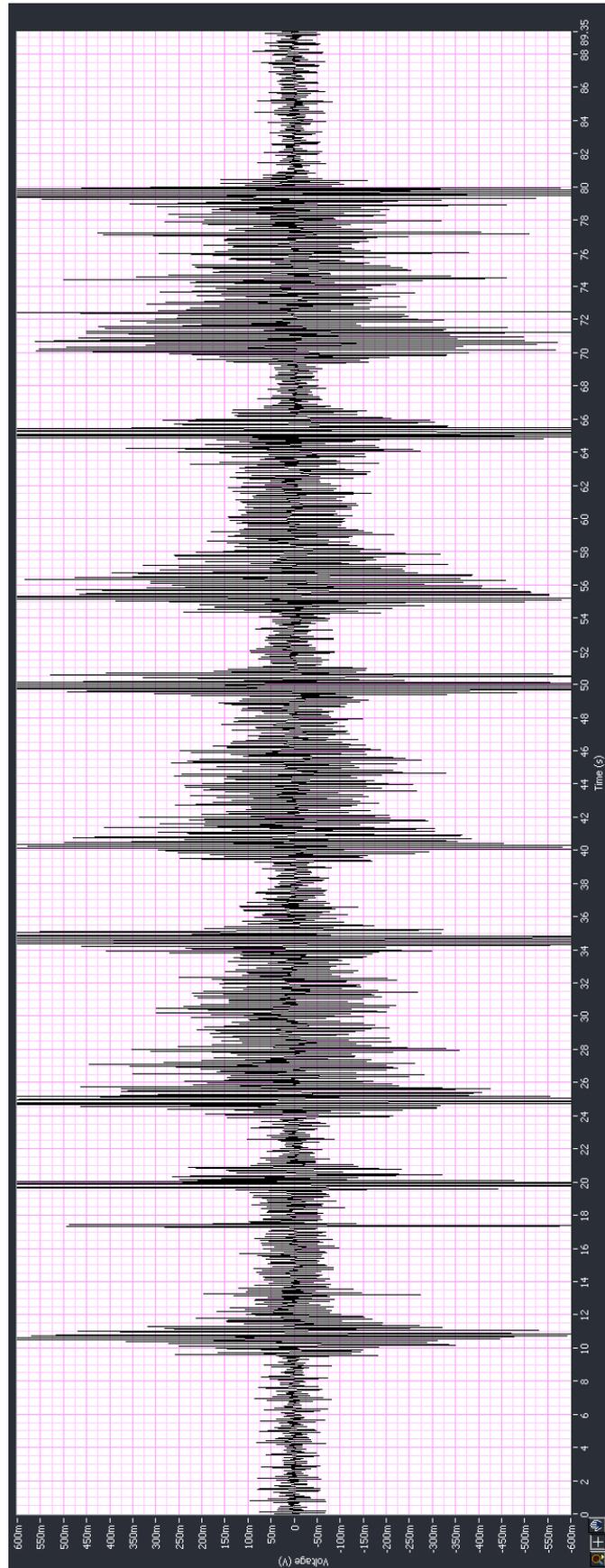


Figure 8.1.8 – Test 4b

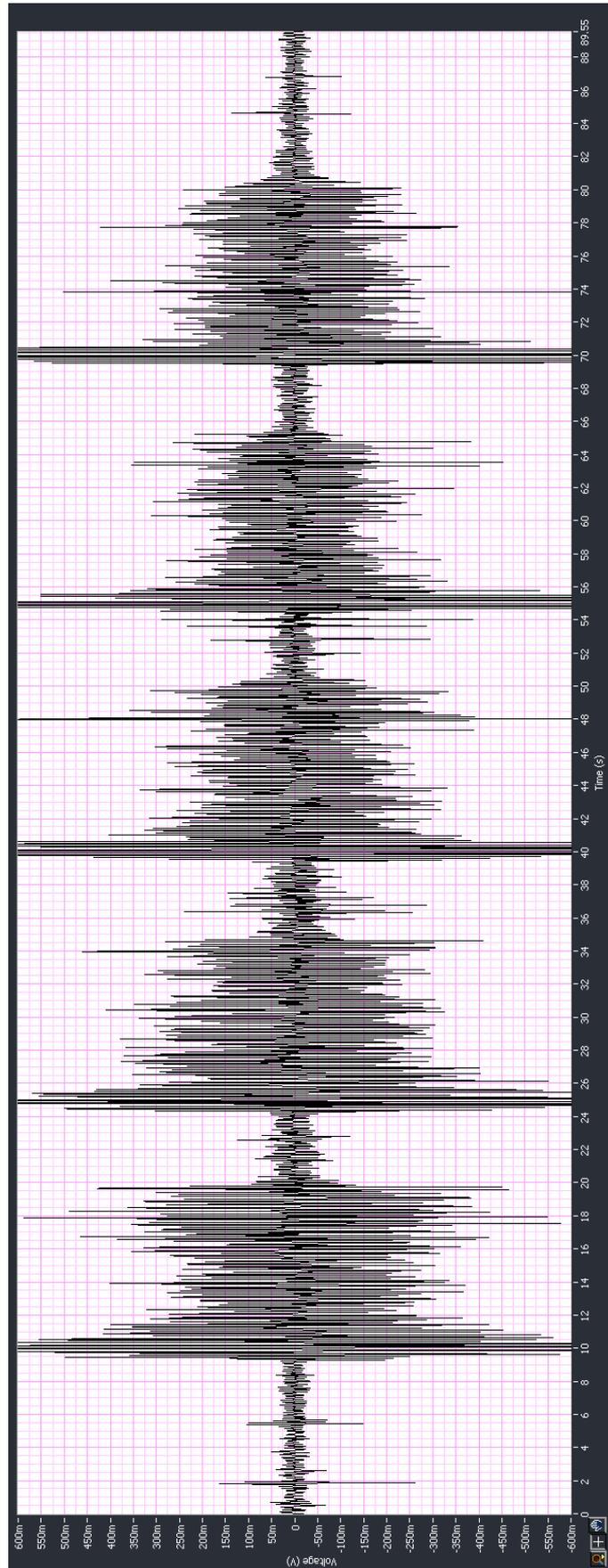


Figure 8.1.9 – Test 4c

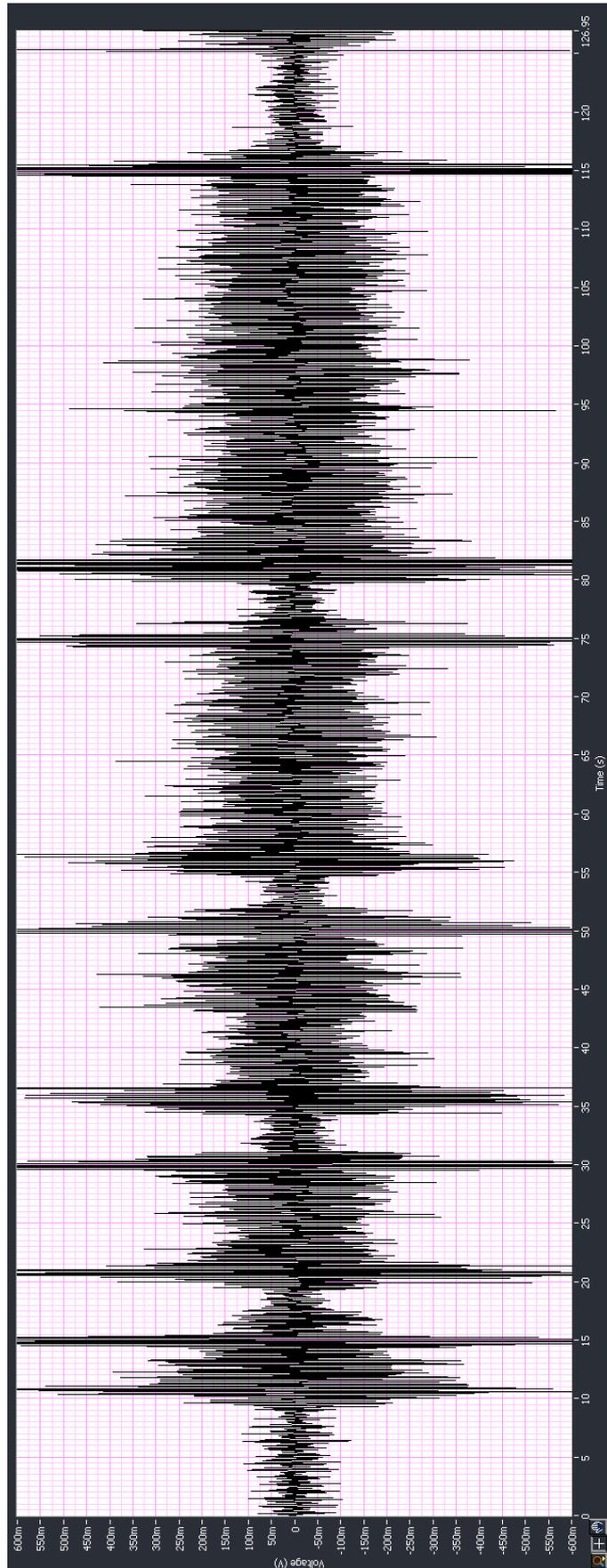


Figure 8.1.10 – Test 5a

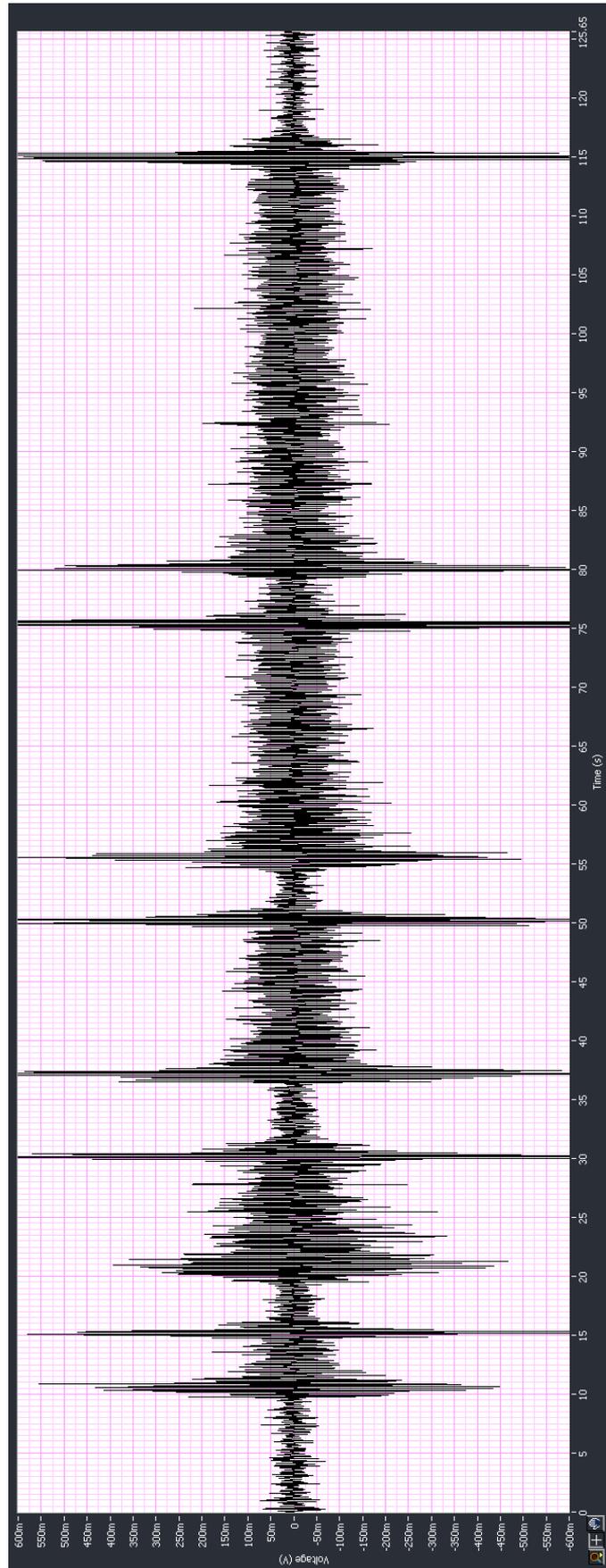


Figure 8.1.11 – Test 5b

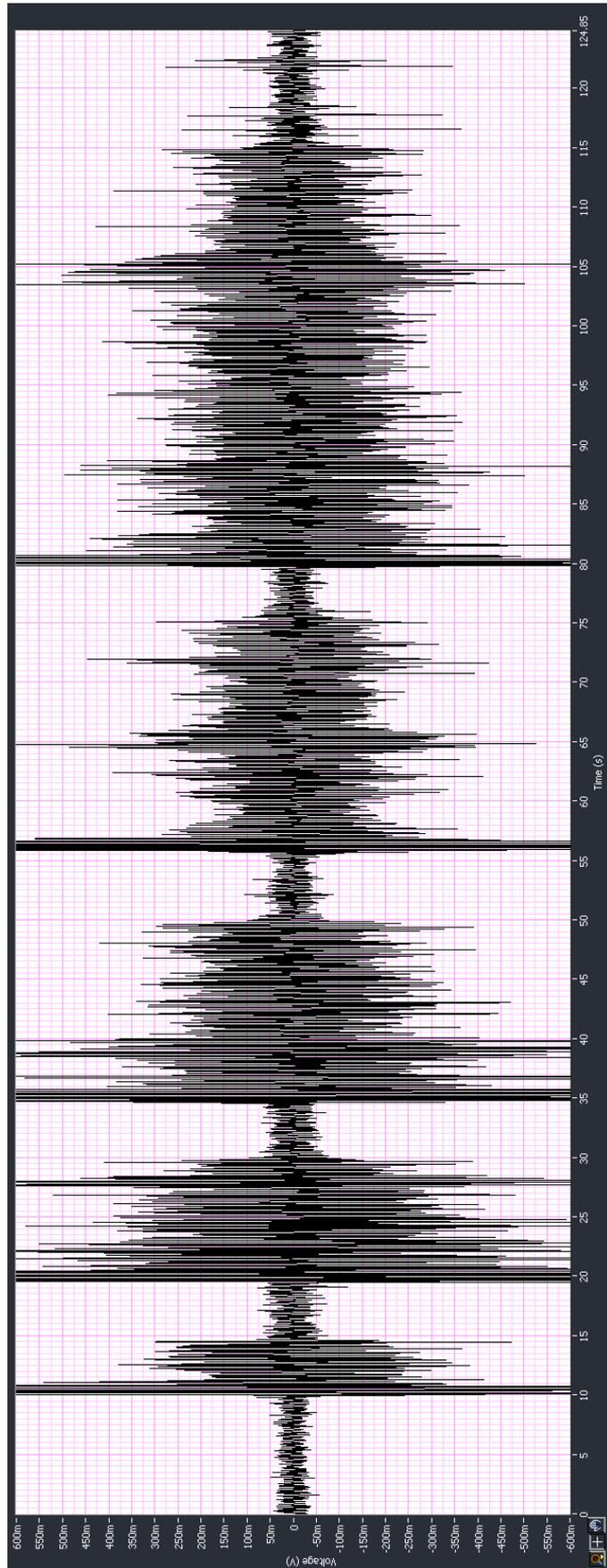


Figure 8.1.12 – Test 5c

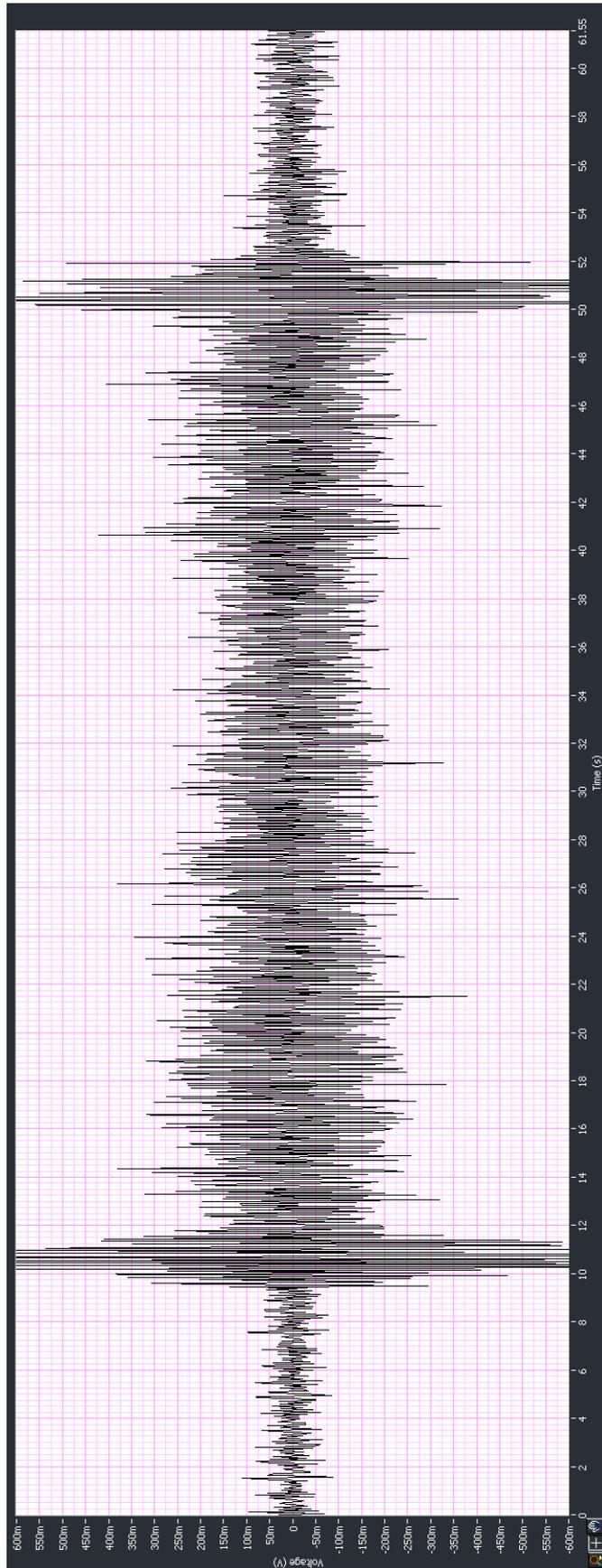


Figure 8.1.13 – Test 6a

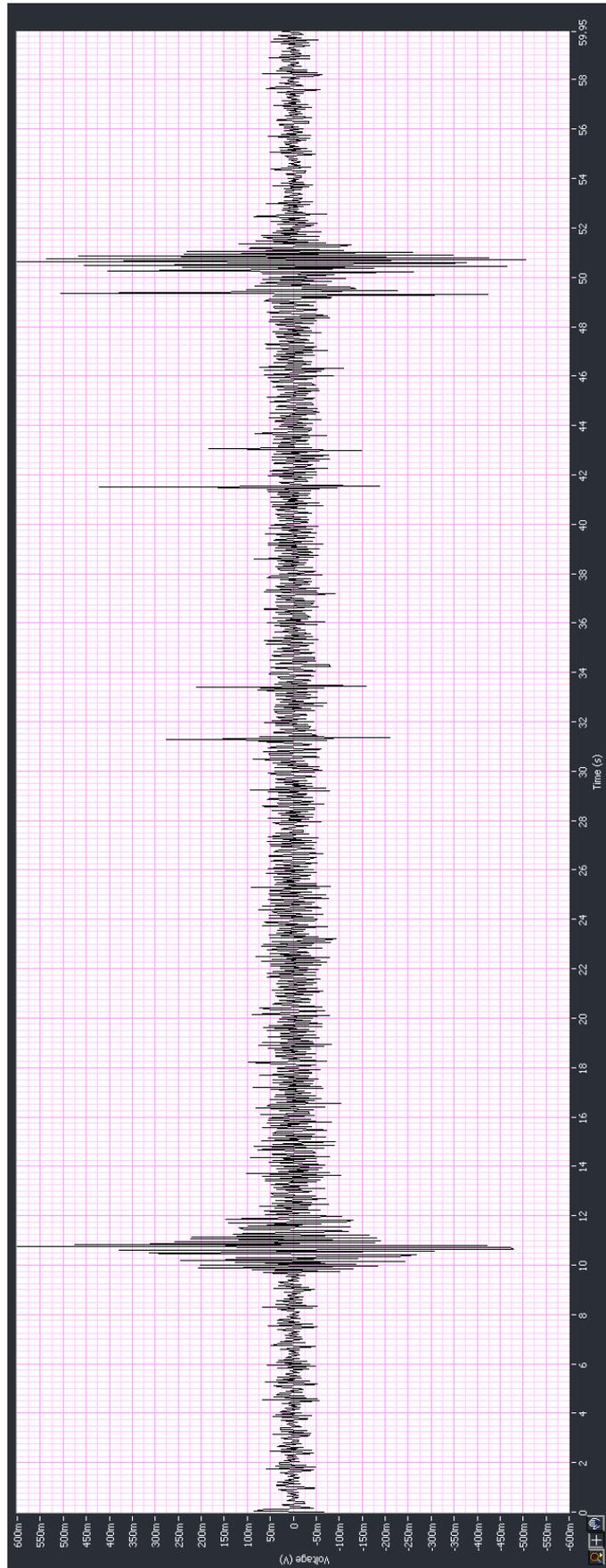


Figure 8.1.14 – Test 6b

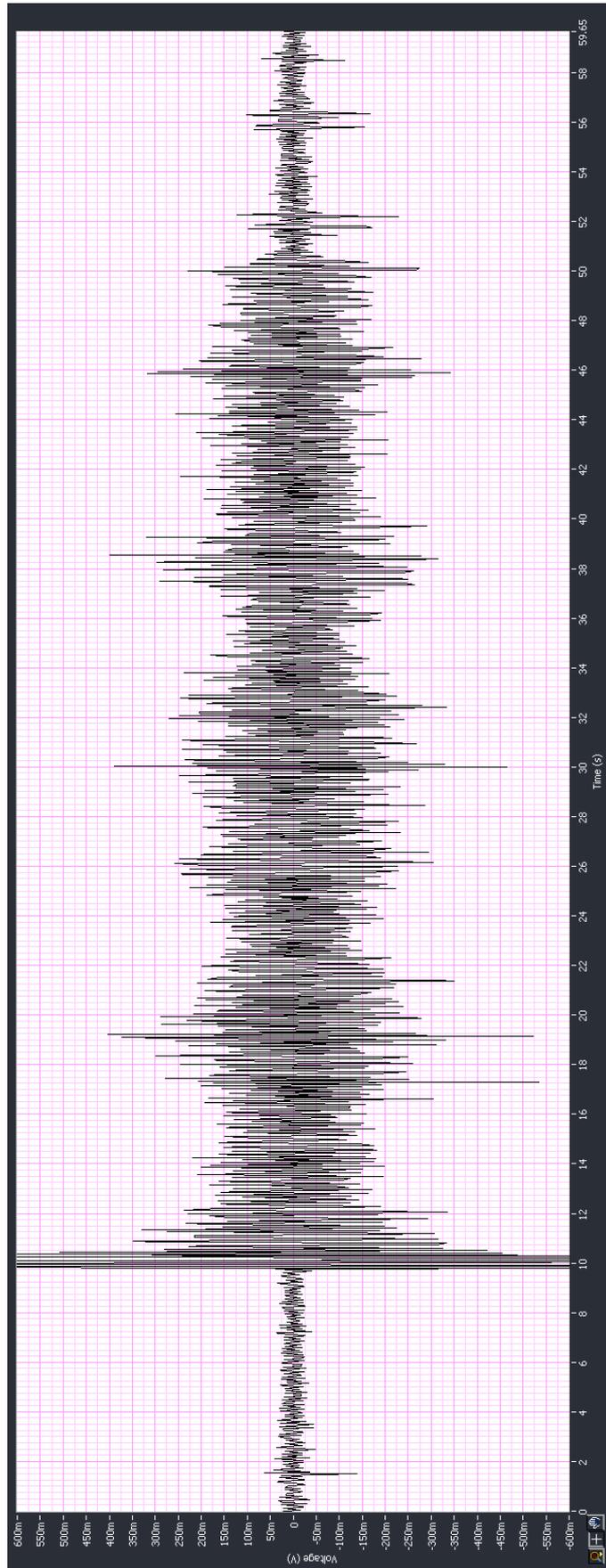


Figure 8.1.15 – Test 6c

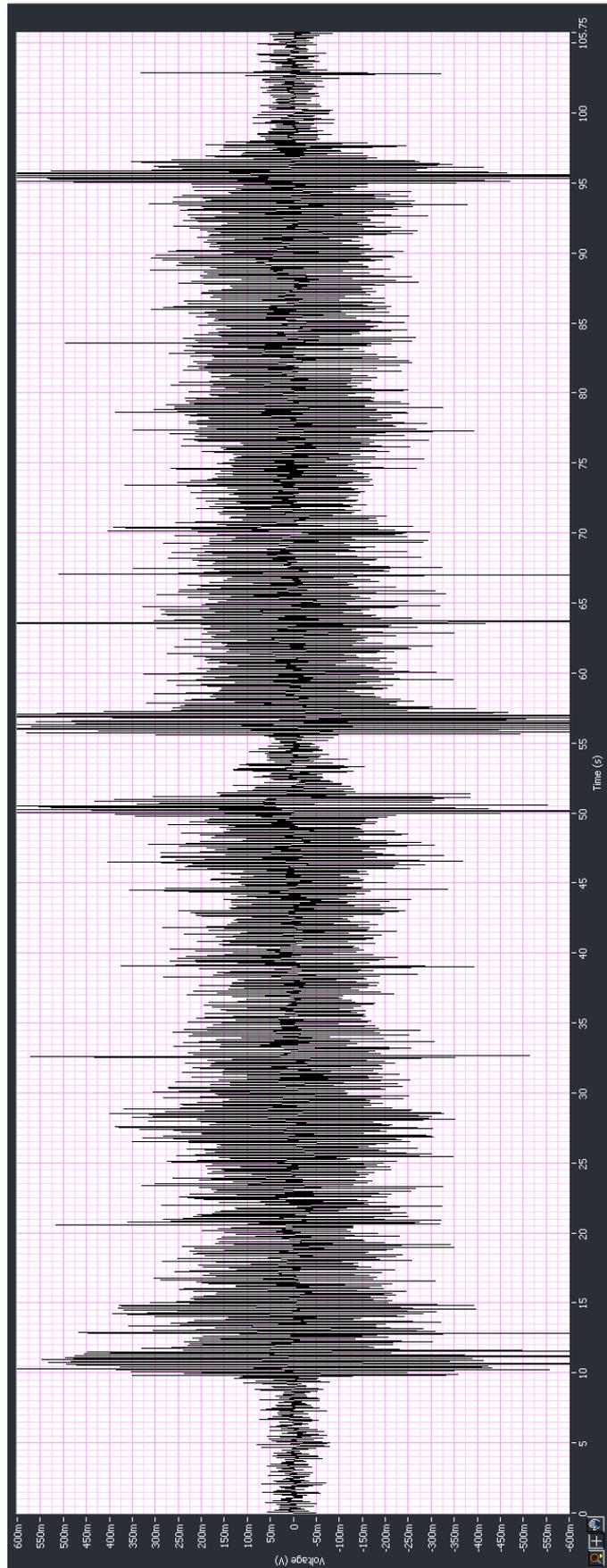


Figure 8.1.16 – Test 7a

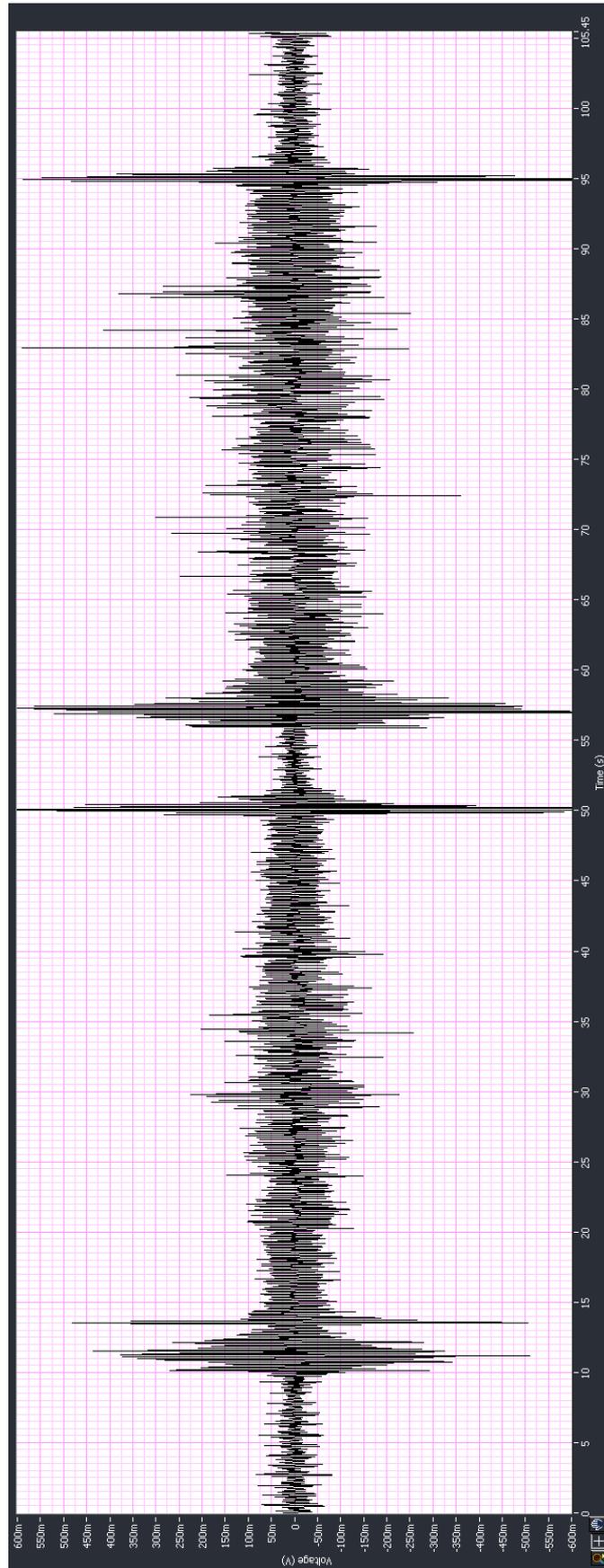


Figure 8.1.17 – Test 7b

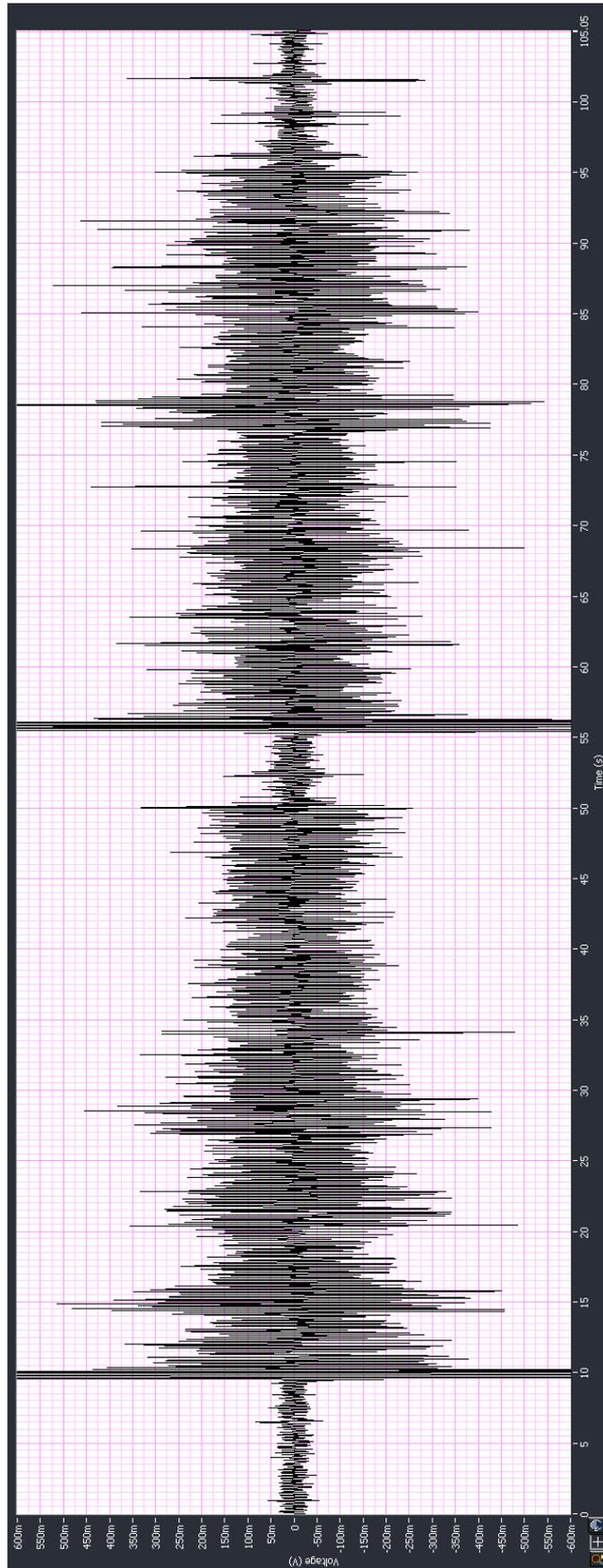


Figure 8.1.18 – Test 7c

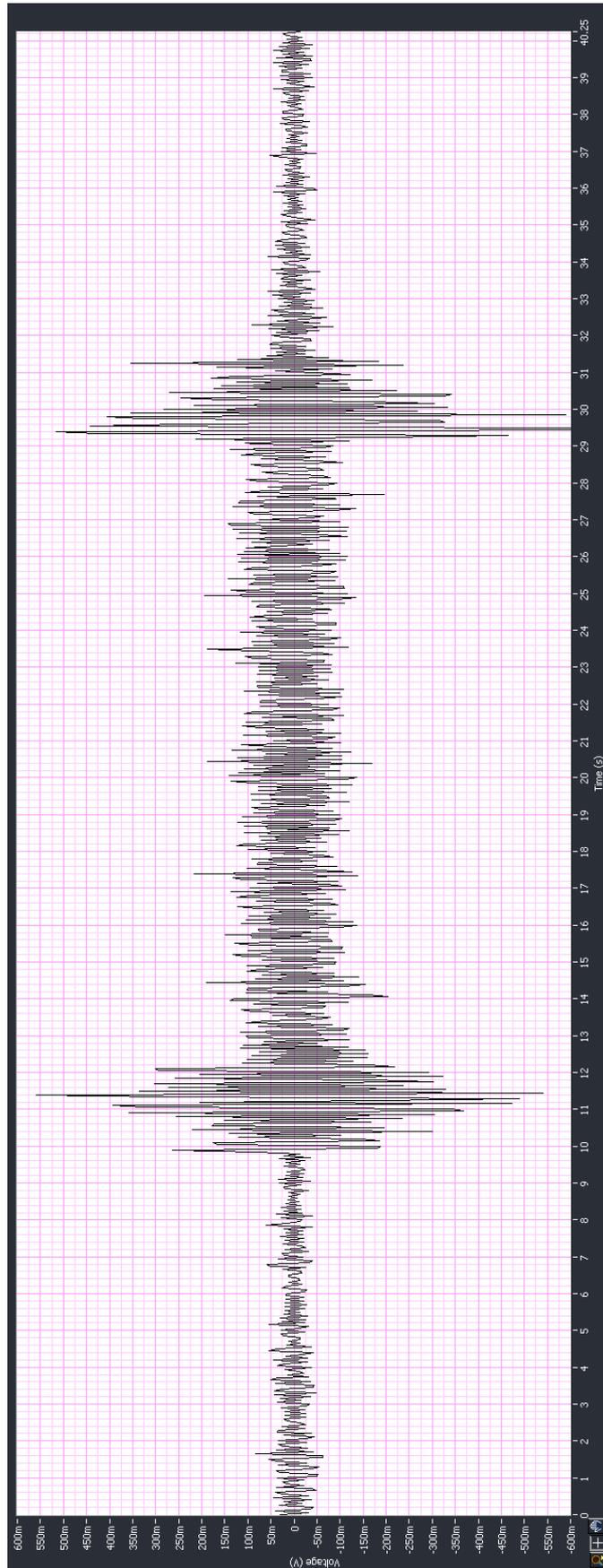


Figure 8.1.19 – 5-100Hz

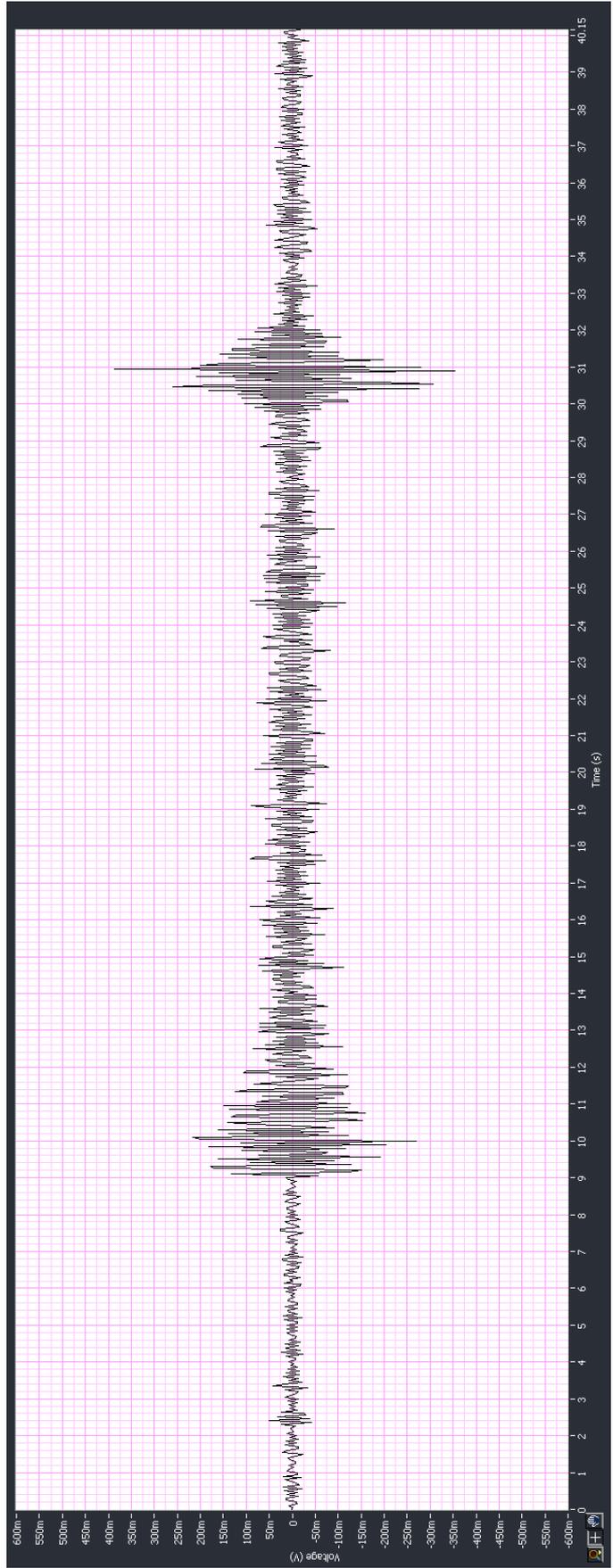


Figure 8.1.20 – 5-20Hz

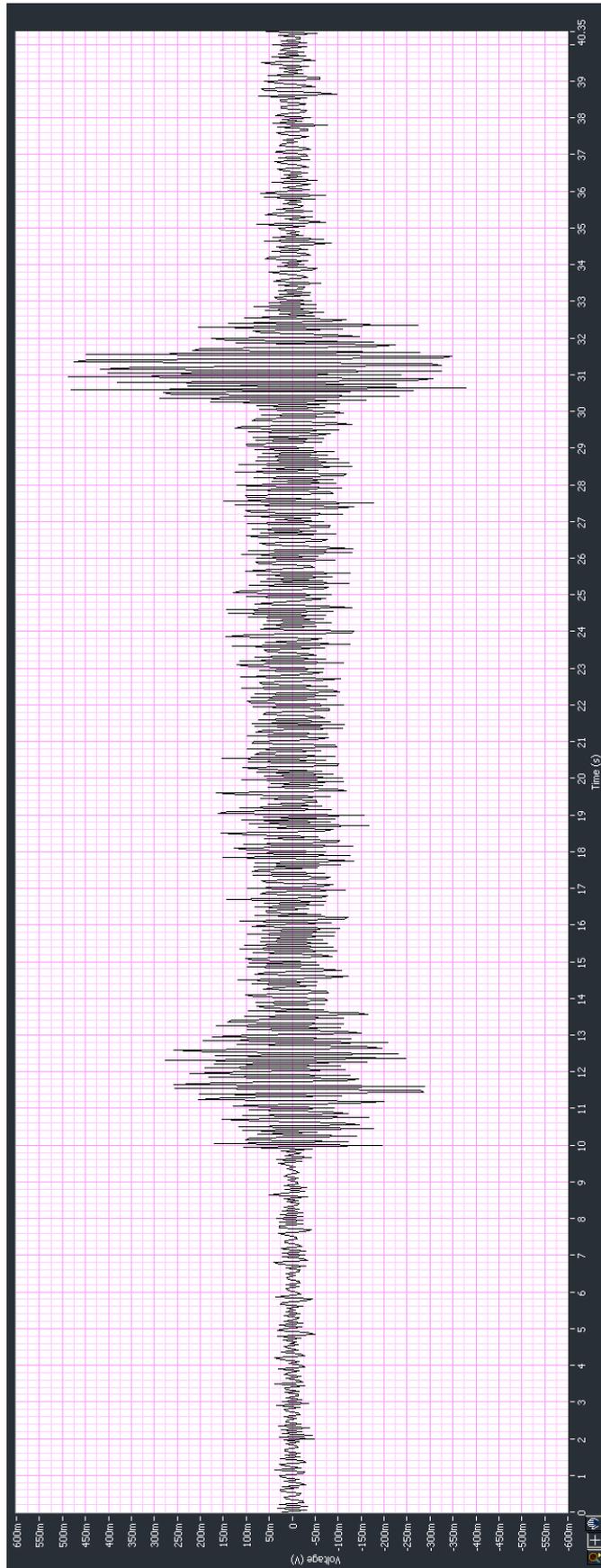


Figure 8.1.21 – 5-50Hz

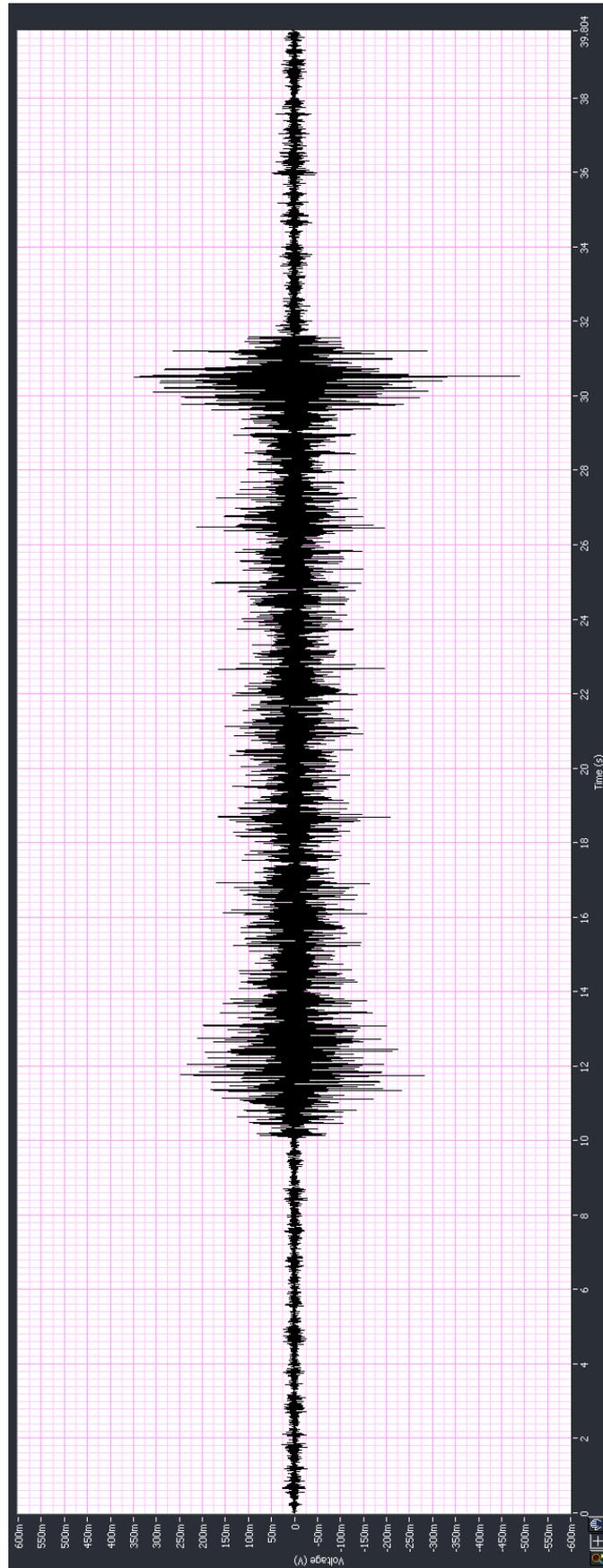


Figure 8.1.22 – 20-50Hz

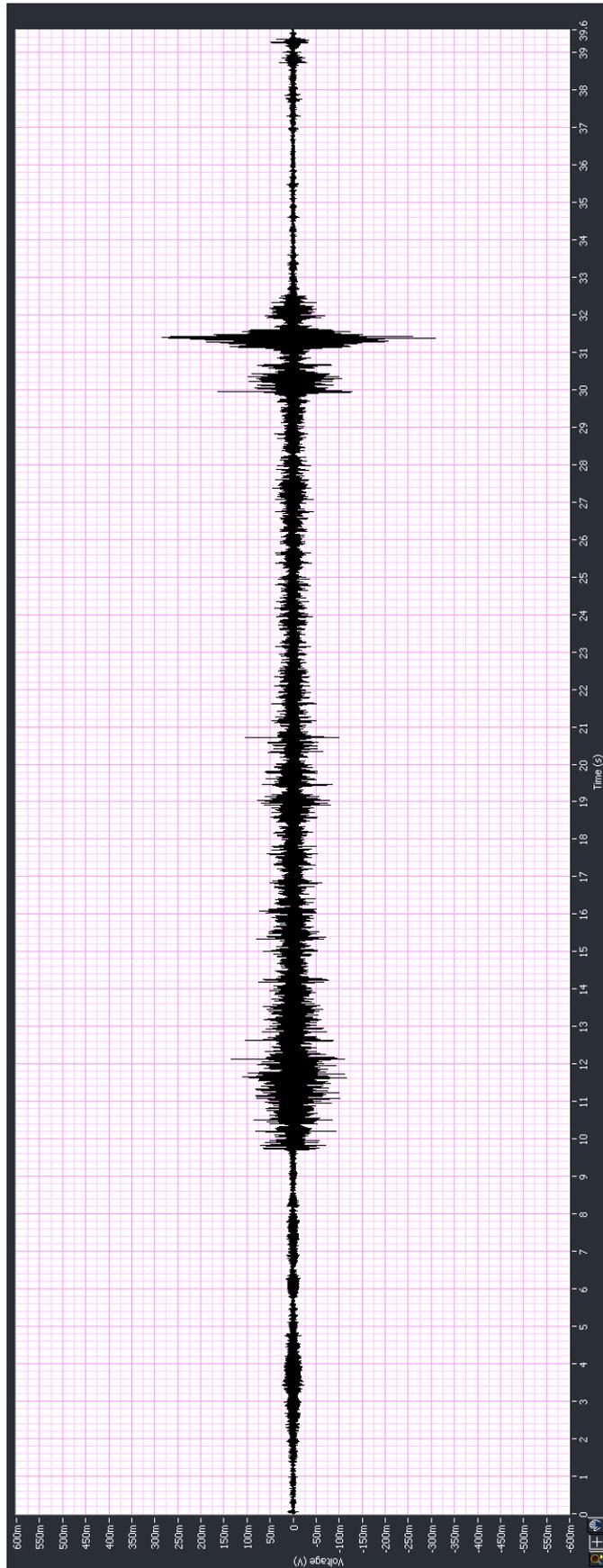


Figure 8.1.23 – 50-100Hz

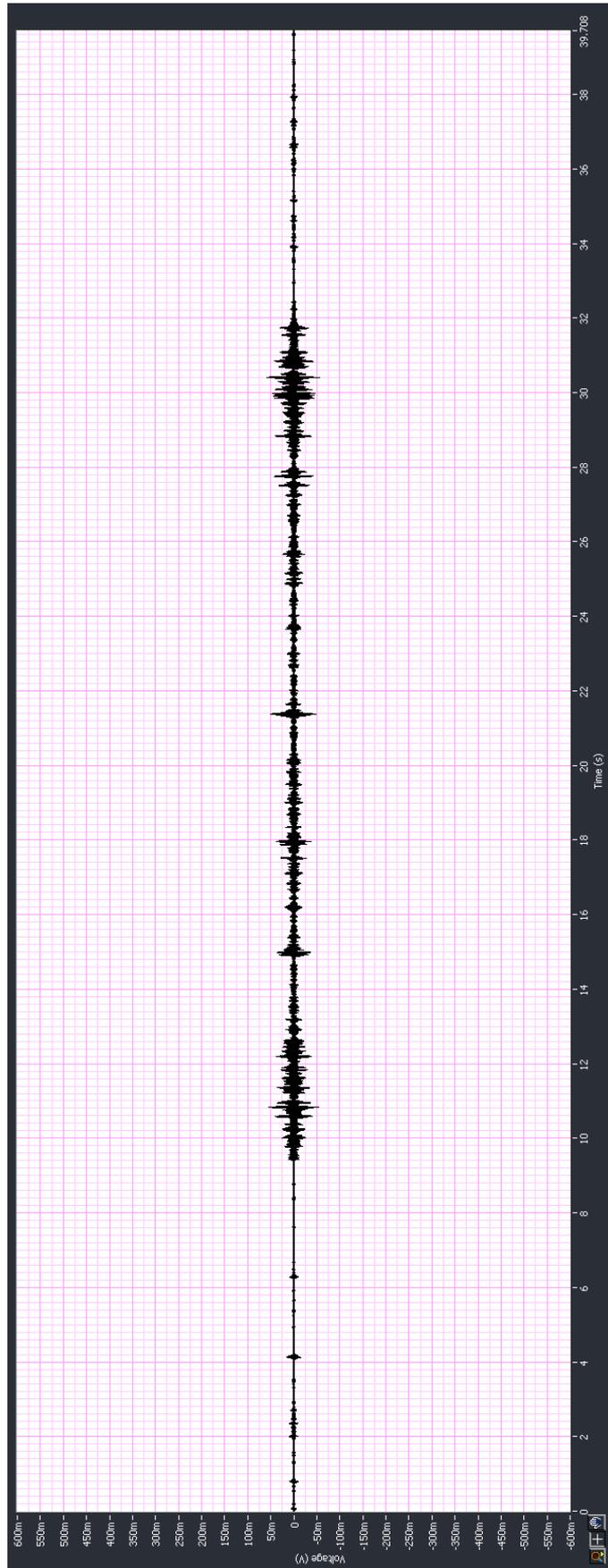


Figure 8.1.24 – 80-100Hz

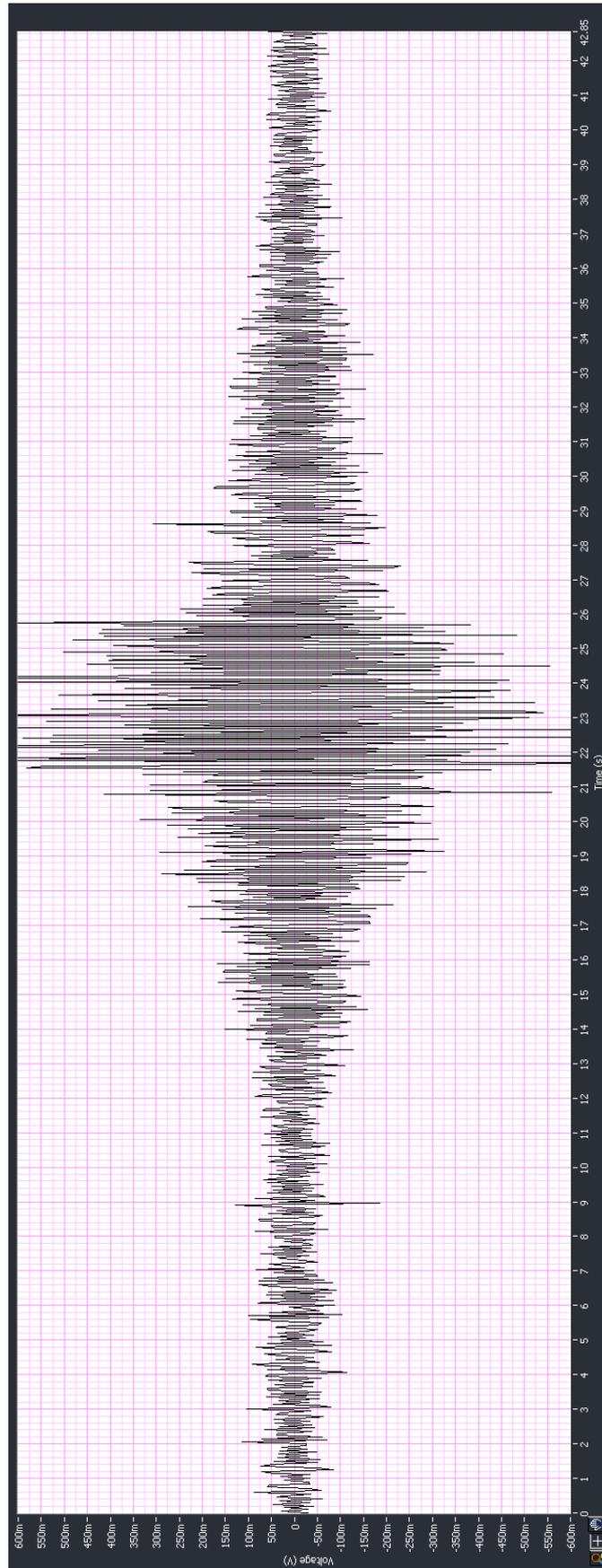


Figure 8.1.25 – Test Increasing Mass

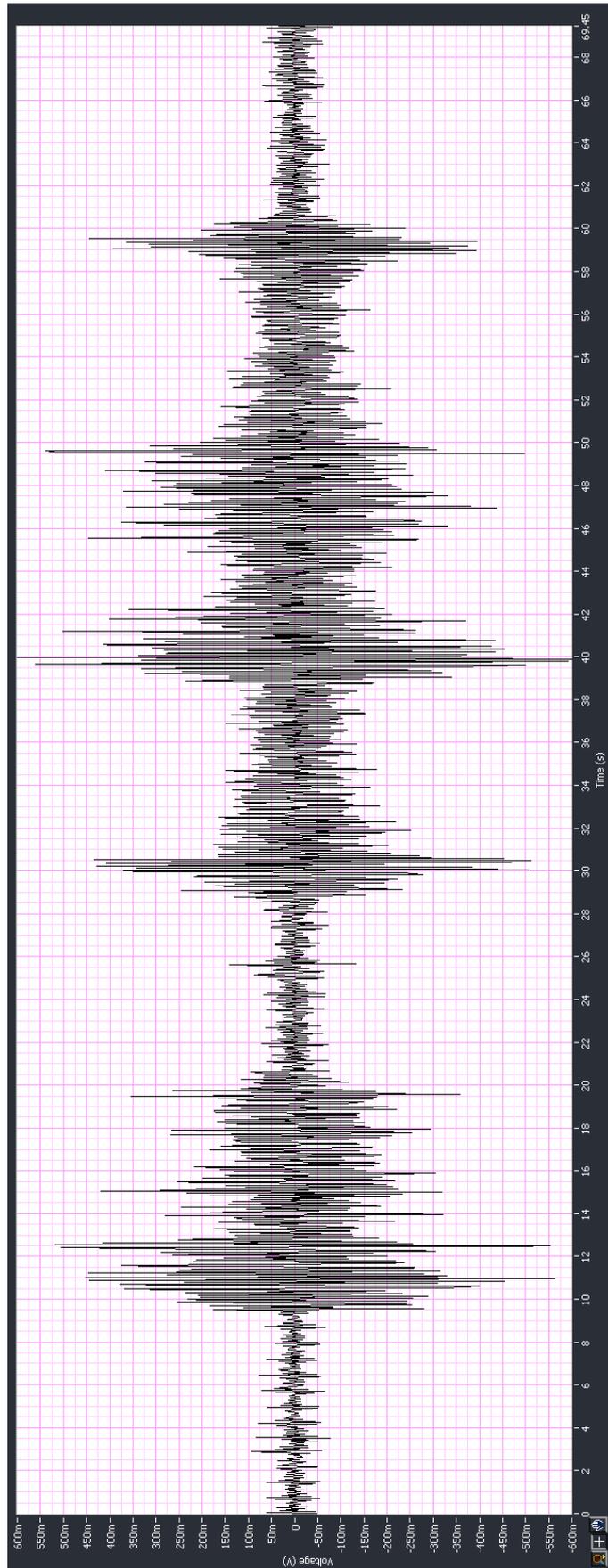


Figure 8.1.26 – Test Muscle Interaction

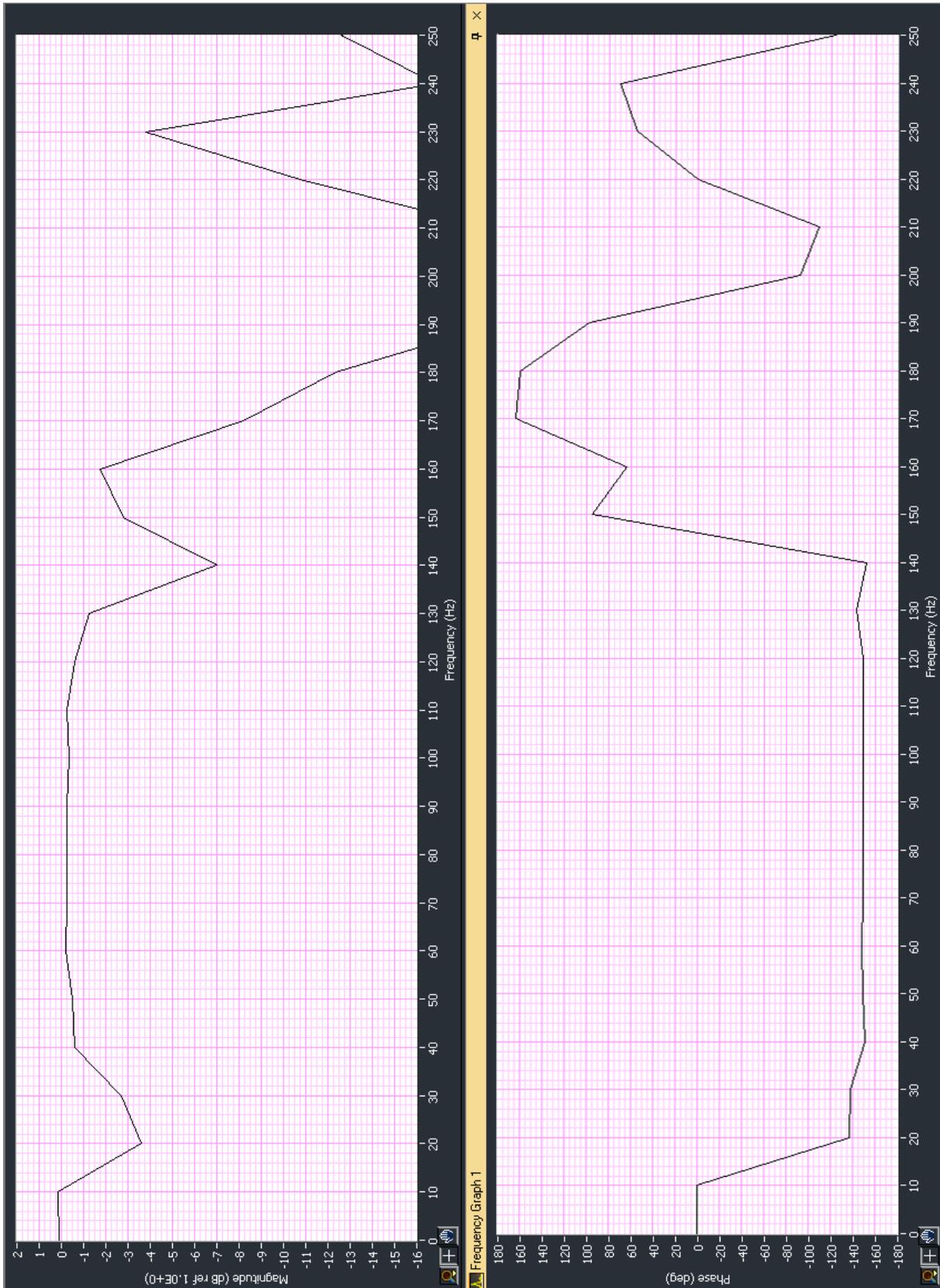


Figure 8.1.27 – Hardware Filter Frequency and Phase response

8.2 Eagle PCB Layouts

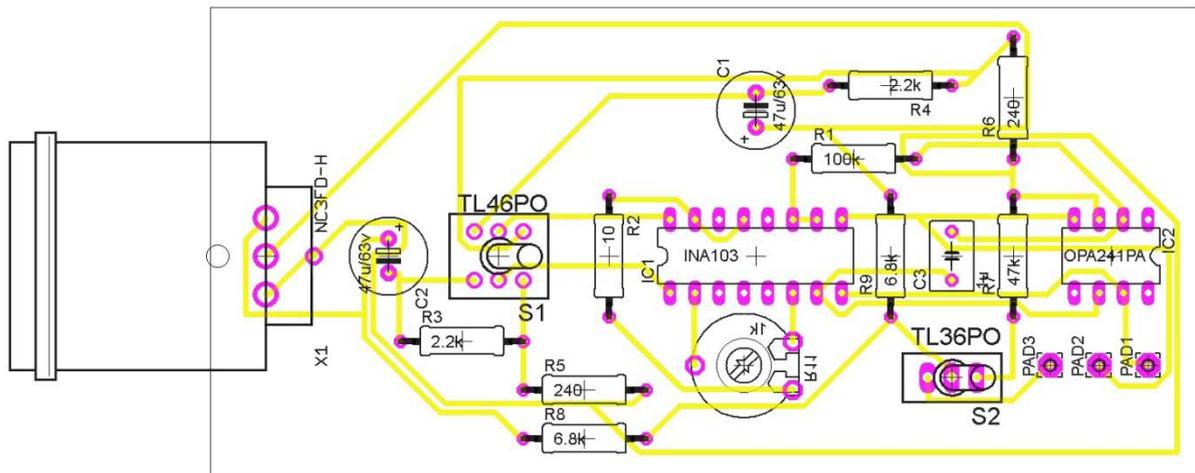


Figure 8.2.1 – Microphone Preamp Layout

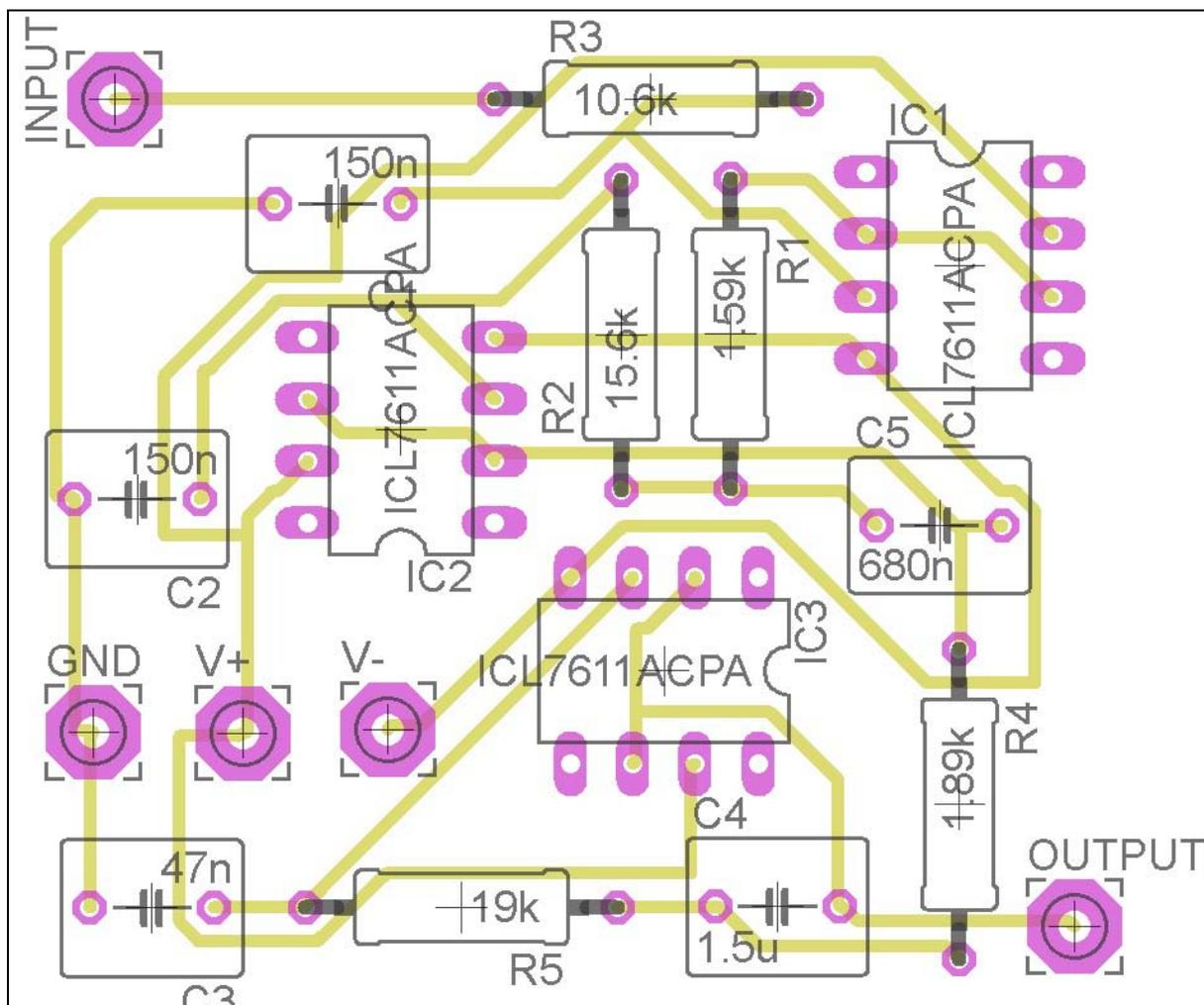


Figure 8.2.2 – Filter Layout

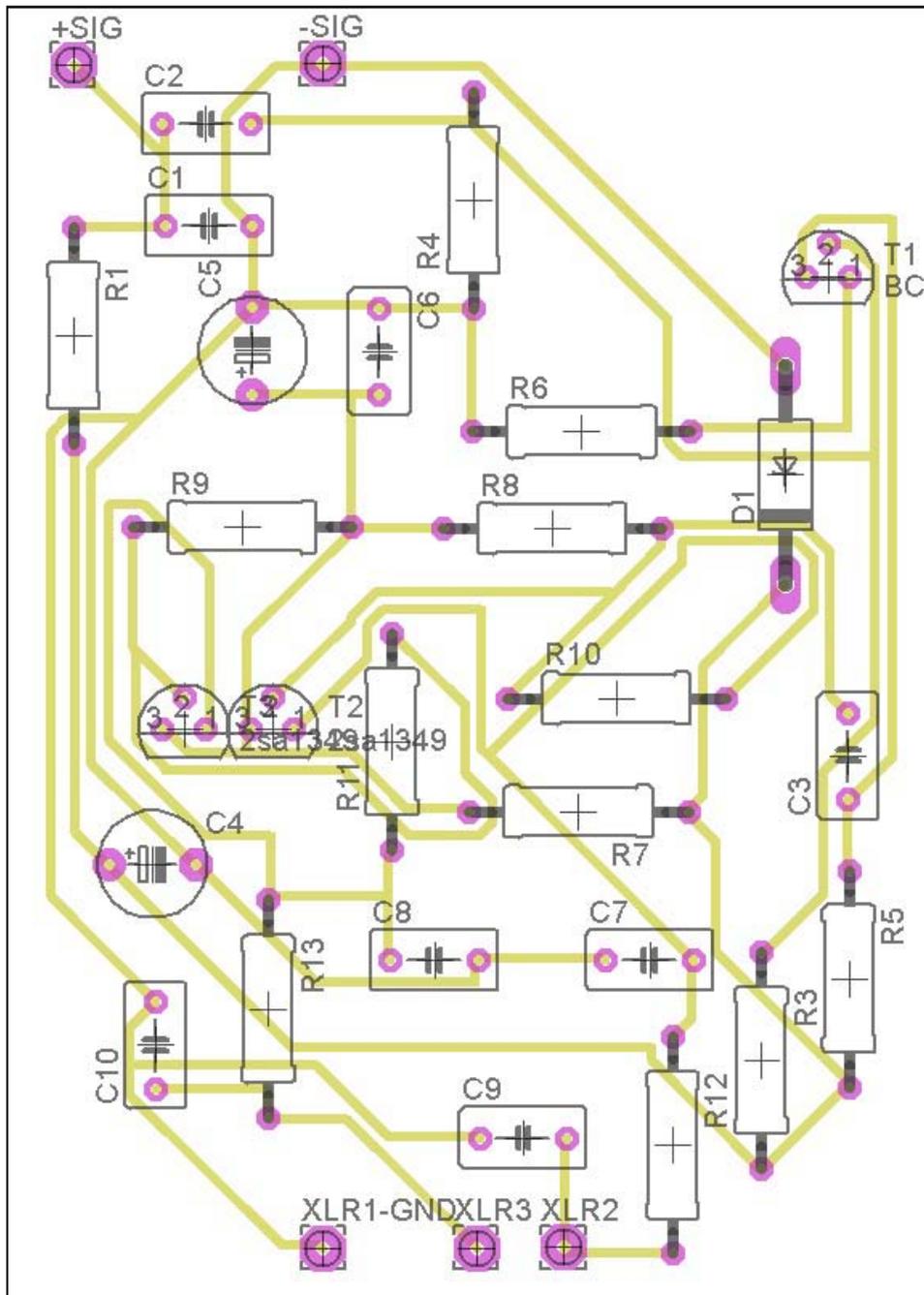


Figure 8.2.3 – Line Preamp Layout

Specifications:

- Track width: 0.5mm
- Pad Width: 0.5mm
- Track – track clearance: 0.3mm

- Track – pad clearance: 0.3mm
- Pad – pad clearance: 0.3mm
- Hole dimensions: 1mm
- Grid tolerance: 0.1mm
- Layer: top

The thinnest cutter head available is 0.2mm so any clearances must be larger than 0.2mm. Eagle has the ability to perform double layer boards however, for simplicity a single layer design was made.

8.3 Filter Order Test Results

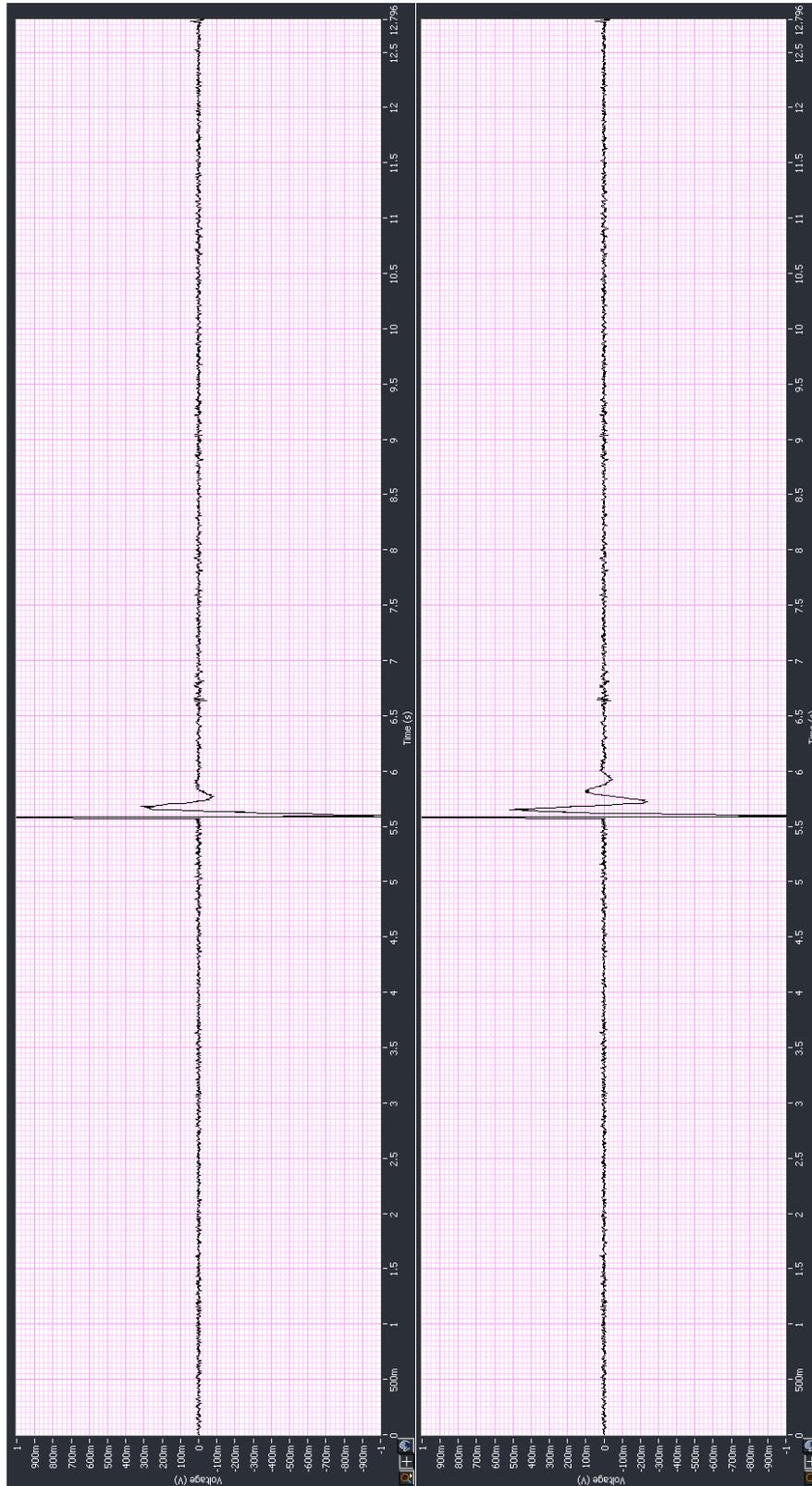


Figure 8.3.1 – 3rd Order (Left), 5th Order (Right)

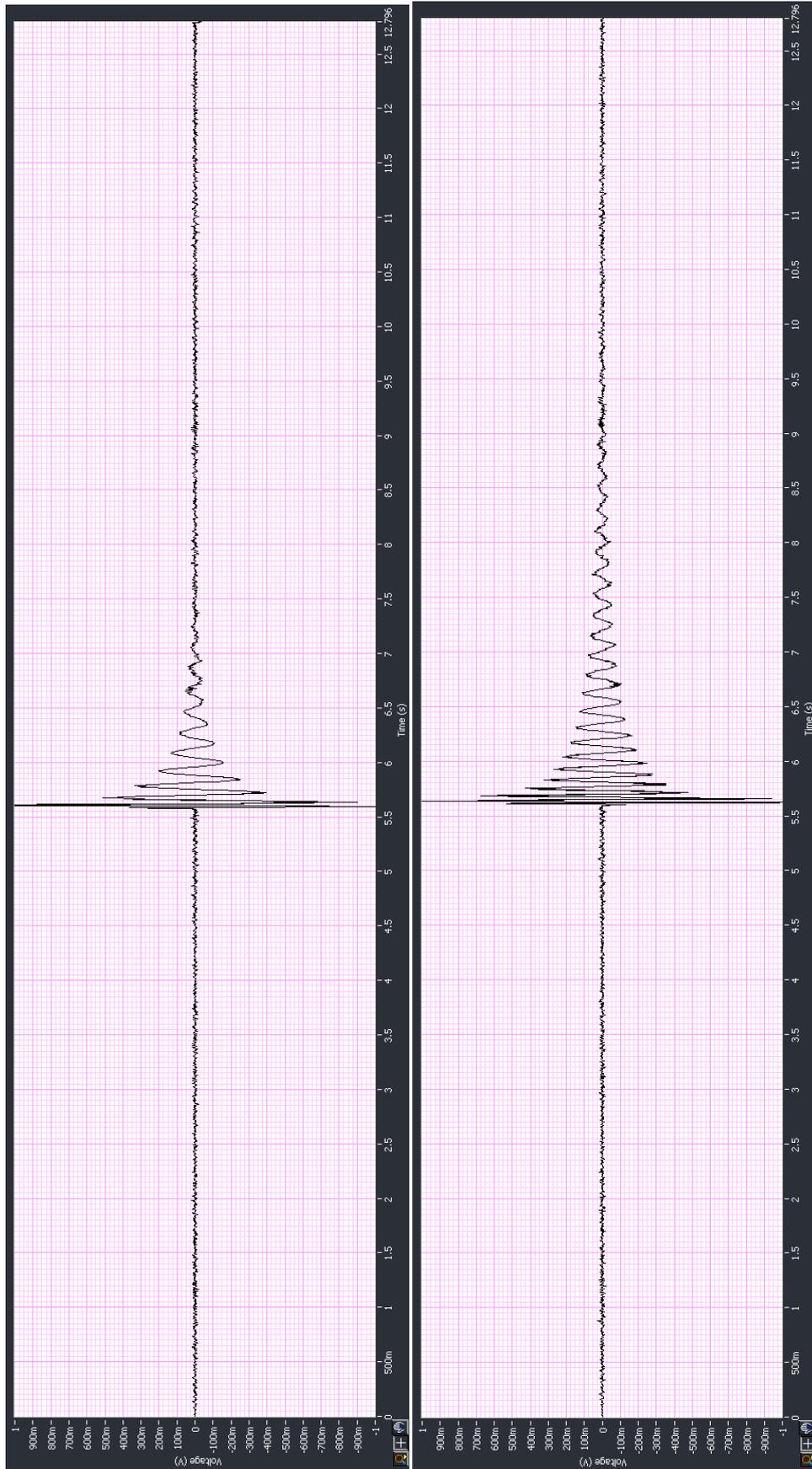


Figure 8.3.2 – 20th Order (Left), 50th Order (Right)

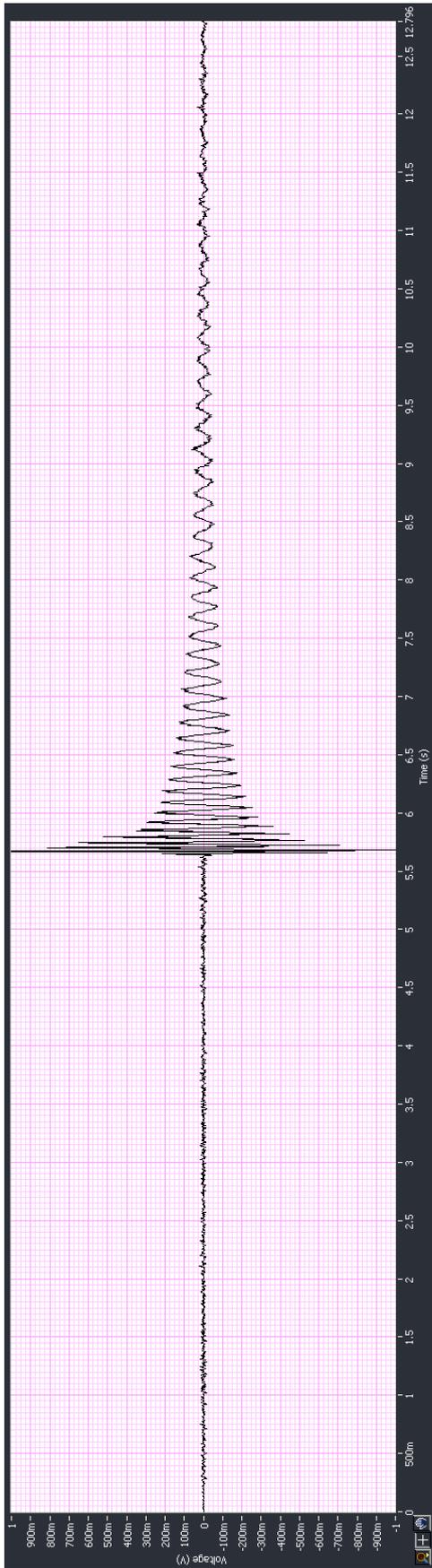


Figure 8.3.3 – 100th Order