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MASTER’S THESIS

Topic: Разработка биотехнической системы для управления протезом с повышенной надёжностью (Development of bioengineering system for the control of prosthetic device with enhanced reliability)

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TASK FOR THE MASTER’S THESIS

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Group 1500

Topic: Development of bioengineering system for the control of prosthetic device with enhanced reliability.

Institution: Saint Petersburg Electrotechnical University (ETU)

Initial data (technical requirements):
Design of the signal acquisition unit (SGU), control signal creation unit and control unit (CU) for a prosthetic device with enhanced reliability using alternative methods.

Contents of the thesis:
Problems found in the design of SGU and CU for prosthetic devices, solution to the problems found in these topics and a Safety chapter on the developed system.

List of reporting materials: the text of the GQW, illustrations, tables, presentation.

Additional sections: Safety

The task was given

«___»______________20___.

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CALENDER PLAN FOR THE GRADUATE QUALIFICATION WORK

Student: Салинас Бустильо Н. Э. (Salinas Bustillo N. E.)
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<table>
<thead>
<tr>
<th>№</th>
<th>Stages</th>
<th>Deadline</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Bibliography review</td>
<td>13.02 – 28.02</td>
</tr>
<tr>
<td>2</td>
<td>Patent search</td>
<td>01.03 – 14.03</td>
</tr>
<tr>
<td>3</td>
<td>Signal gathering unit design</td>
<td>15.03 – 10.03</td>
</tr>
<tr>
<td>4</td>
<td>Control signal creation</td>
<td>11.04 – 29.04</td>
</tr>
<tr>
<td>5</td>
<td>Control unit design</td>
<td>30.04 – 24.05</td>
</tr>
<tr>
<td>6</td>
<td>GQW submission</td>
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Scientific advisor: PhD, Professor (Academic degree, title) Yuldashev Z. M.
SUMMARY

Explanatory note 88 pages, 51 figures, 1 chart; 4 tables, 30 sources.

Key words: Electromyography, Eddy Currents, Lorentz Vibrations, Myotonic Signals, Prosthesis, Electromagnetic Crosstalk, Piezoelectric Effect, Piezo Element, Tensiomyography.

The subject of the research is: Design of a bioengineering system for the control of a prosthetic device with enhanced reliability.

The target of the GQW – To design the signal gathering unit, control unit and create control signals for a prosthetic upper limb device.

This thesis covers the development process of a bioengineering system for the control of a prosthetic device with enhanced reliability. Additionally, the device must be a low-cost solution for the control of the prosthesis and provide a reliable method that is more or as accurate as existing control methods of the same price range.

The project is divided in three main development stages: The design of a signal gathering unit using piezo elements, the conversion of the signals from the piezo elements into signals that can be used to control the actuation unit and the design of the control unit to interact with the actuators.

The final prototype is a sensor using two piezo elements with a rubber O-ring between them to increase the deformation. The sensor is followed by a comparing stage with the TL061 OPAMP to obtain a square signal. This signal is used to switch the outputs of a multiplexor or to change the output of a digital potentiometer to control the width of the pulse signal created by the IC555 used in the control signal. The final device has a 0,91 average accuracy, 0,95 average sensitivity and 0,87 average specificity when used in the brachioradialis muscle and 0,81, 1 and 0,67 average accuracy, sensitivity and specificity respectively when used in the triceps brachii muscle.
РЕФЕРАТ

В работе представлен процесс разработки биотехнической системы для контроля протеза руки с повышенной надёжностью. Предложенный вариант контроля должен быть недорогим в реализации и в то же время обеспечивать надёжный метод регистрации управляющих сигналов, не хуже, чем у существующих методов того же ценового диапазона.

Проект включает три основные стадии разработки: проектирование блока регистрации сигналов с использованием пьезоэлектрических датчиков, преобразование сигналов с датчиков в сигналы управления исполнительным узлом, и конструкция блока управления для взаимодействия с исполняющим приводом.

Конечным прототипом является датчик, в котором используются два пьезоэлектрических датчика с резиновым уплотнительным кольцом между ними для увеличения деформации. За датчиком следует сравнительный каскад на операционном усилителе TL061 для получения управляющего сигнала прямоугольной формы. Этот сигнал используется для переключения выходов мультиплексора или изменения выходного сигнала цифрового потенциометра для управления шириной импульсного сигнала с аналогового таймера, используемого в качестве управляющего сигнала для привода.
# TABLE OF CONTENTS

DEFINITIONS, DESIGNATIONS AND ABBREVIATIONS ........................................... 8

INTRODUCTION ..................................................................................................... 9

CHAPTER 1 KNOWLEDGE BASE ........................................................................... 10

1.1 Practical and Scientific Application ................................................................... 10
   1.1.1 Practical Application of the BES ............................................................... 10
   1.1.2 Scientific Application of the BES .............................................................. 10

1.2 Theoretical Analysis and Patent Search ............................................................ 11
   1.2.1 Myocontrolled Prostheses ....................................................................... 11
   1.2.2 Electromyography ................................................................................... 11
   1.2.3 Tensiomyography .................................................................................... 22
   1.2.4 Actuation devices and control methods used with prosthetic devices .... 26
   1.2.6 Feedback measurement for upper limb prosthetic devices ..................... 36
   1.2.7 Patent Search ........................................................................................... 39

1.3 Conclusions for the problems found at the knowledge base ......................... 42

CHAPTER 2 PROJECT DESCRIPTION ..................................................................... 44

2.1 Model Creation .................................................................................................. 45
   2.1.1 Piezoelectric effect ................................................................................... 45
   2.1.2 Piezoelectric force sensor ......................................................................... 46
   2.1.3 Electronic model of a piezoelectric force sensor ........................................ 48
   2.1.4 Sensor mathematical model description ................................................... 55
   2.1.5 Signal treatment board ............................................................................. 58

2.2 Control Unit ...................................................................................................... 58

2.3 Servomotor control ........................................................................................... 61

2.4 Mechanical mechanism for the control of the open and close motions .......... 64

2.5 Final system’s block diagram for the solving of the problems presented in
   chapter 1 ............................................................................................................. 65

2.6 Experimental results obtained with designed control device ......................... 67
2.6.1 Experiment Description and practical results .................................................. 68
2.6.2 Control Group .................................................................................................. 71
2.7 Conclusions for the application part of the project .......................................... 71

CHAPTER 3 SPECIAL ASPECTS OF SAFETY ...................................................... 73
3.1 Description of environmental conditions ............................................................ 73
   A. Temperature, humidity and atmospheric pressure ranges .............................. 74
   B. Indoor or outdoor operation ........................................................................... 75
3.2 List of harmful and dangerous factors and risk analysis .................................. 75
   A. Technical means to prevent harmful impacts .................................................... 75
   B. Electrical safety according to IEC 60601-1-2012 «General requirements for basic safety and essential performance» ......................................................... 76
   C. Electromagnetic compatibility according to IEC 60601-1-2-2014 «Electromagnetic disturbances – Requirements and tests» ........................................... 77
3.3 Conclusions for the special safety requirements ............................................... 82

CONCLUSIONS ........................................................................................................ 84

REFERENCES AND ONLINE BIBLIOGRAPHY ................................................ 86
APPENDIX A ............................................................................................................. 89
   Modes of vibration for common piezoelectric ceramic shapes ............................ 89
APPENDIX B ............................................................................................................. 90
   Pinout for the components used in the project ..................................................... 90
APPENDIX C ............................................................................................................. 91
   Voltage Divider Diagram and SPI interface Write Sequences and timing diagram and potentiometer measurements for TI’s TPL051 ............................................. 91
APPENDIX D ............................................................................................................. 93
   Example page to register the SGU response under muscle contraction ............ 93
DEFINITIONS, DESIGNATIONS AND ABBREVIATIONS

The present master dissertation work uses the following abbreviations and designations:

- AP – Action Potential
- BES – Bioengineering System
- EMG – Electromyography
- EMI – Electromagnetic Interference
- GQW – Graduate Qualification Work
- ME equipment – Medical Equipment
- PFS – Piezo Force Sensor
- PWM – Pulse Width Modulation
- TMG – Tensiomyography
INTRODUCTION

Nowadays, prosthetic devices have evolved in at a surprising speed. 3D printing has given designers, engineers and freelances the opportunity to test and try new and different architectures for these devices. The control for prosthetic upper limb devices has also develop in a fast manner, from myocontrolled prosthetic devices that use bioelectrical signals from the patient’s remaining arm muscles, prosthetic hands controlled with artificial synapses with signal taken directly from the brain, to more modest developments like speech recognition technology and eye movement measurements for the control of special prosthetic hands. All current devices have their strengths as well as their limitations, specially, the fact that most of them are very expensive and not affordable by the average consumer. High-end devices have a very large reliability, as they can be controlled with almost no mistakes and with high functionality, but this control methods usually round tenth of thousands of dollars in price. More affordable methods, suffer from one common enemy, their lack of reliability, they are hard to train, hard to control, prone to mistakes devices.

For the reasons presented above, a good reliability and affordable method for the control of prosthetic devices is still a necessity, and so the development of such a device is proposed in this project using piezo elements and low-end processing methods, but new in concept to tackle this situation.

The main target of the work is to design a signal gathering unit that does not relays in electrical signals, this, to reduce the noise produced during sensing and reduce the need for filtration, pre-processing and such stages. Additionally, the creation of a unit able to manage these signals and transform them into control signals is in need. And finally, the control signal to interact with the actuation unit and the prosthetic device must be design. The process of creation of all this units is presented in this work, as well as an overview of another low-end control method for comparison.
CHAPTER 1 KNOWLEDGE BASE

1.1 Practical and Scientific Application

1.1.1 Practical Application of the BES

The control method presented in this work intends to provide enhanced reliability to existing solutions of prosthetic devices. Its practical application is found in the field of myocontrolled prosthetics, specifically in upper limb prosthetics control.

The BES has two main applicability concepts. The first one, is to be added to a mechanical prosthetic device, such as a shoulder mounted mechanical prosthetic arm, the purpose is to add additional movements to device by measuring muscle activity in two muscle groups and provide three additional movements to the original movement set. The second application, is in the field of myoelectric controlled prosthetics, by adding this device to EMG boards, the degree of reliability in the measurements and the response time for simple tasks can be improved by delegating the control of this simple movements such as non-complex shape solids holds or one finger based gestures (pointing, thumbs up, etc.) to the BES and leaving the EMG measurements the task of controlling more complex actions and recognizing more sophisticated patterns coming from the patient.

1.1.2 Scientific Application of the BES

Although qualitative measurements are the focus of the BES at this stage of development, more reliable transducers and in a larger number can be aggregated to increase the possibilities of the BES overall.

These additions, can make it so the BES can provide quantitative measurements of the resulting forces and the changes in tension (muscle tone) of the muscle groups involved in a specific movement.

Having a particular interest for those in the field of preventing muscle injuries and the assessment of muscle strengthening and focused training.
1.2 Theoretical Analysis and Patent Search

1.2.1 Myocontrolled Prostheses

The term myocontrolled prostheses refers to any prosthetic device which is controlled using signal coming from a muscle [1]. These signals can come from different sources: the electrical activity of the muscles during contraction or electromyography (EMG), the muscle change in stiffness due to work or tensiomyography (TMG), the displacement of transducer’s tip due to the muscle belly upward inflexion during muscle activity or myotonic signal measurements. Regardless of the source, all signals measure muscle activity (or lack of any) to obtain and recognize a pattern that represents an order from the user that he wants to transform into an action for the prosthetic device.

The degree to which these devices are capable of correctly transforming these orders into actions and the time of this response are the two main factors that should be the focus when referring to the reliability of the prosthetic device. Recommendations given by Castellini (C., 2015) in their research regarding prosthetic devices reliability suggest using incremental learning as a mean to improve patient-device interactions and additionally provide assessment protocols to achieve the previously mentioned.

In the following sections of this chapter an explanation on EMG and TMG is provided as well as two common schemes for these measurements used in today’s practice as well as a novel method of myotonic measure presented in 2011 by Dordevic et. al.

1.2.2 Electromyography

Once appropriate algorithms and methods for EMG signal analysis are readily available, the proper method of acquisition has been chosen and the nature and characteristics of the signal can be properly understood the hardware implementations can be made for various EMG signal related applications.

Due to availability surface EMG (sEMG) signals will be used for the control of the prosthetic arm. Recent advances in technologies of signal processing and mathematical models have made it practical to develop advanced EMG detection
and analysis techniques. Various mathematical techniques and Artificial Intelligence (AI) have received extensive attraction. Mathematical models include wavelet transform, time-frequency approaches, Fourier transform, Wigner-Ville Distribution (WVD), statistical measures, and higher-order statistics. AI approaches towards signal recognition include Artificial Neural Networks (ANN), dynamic recurrent neural networks (DRNN), and fuzzy logic system. Genetic Algorithm (GA) has also been applied in evolvable hardware chip for the mapping of EMG inputs to desired hand actions.

Wavelet transform is well suited to non-stationary signals like EMG. Time-frequency approach using WVD in hardware could allow for a real-time instrument that can be used for specific motor unit training in biofeedback situations. Higher-order statistical (HOS) methods may be used for analyzing the EMG signal due to the unique properties of HOS applied to random time series. The bi-spectrum or third-order spectrum has the advantage of suppressing Gaussian noise [2].

Figure 1.1 shows a representation of the analysis methods mentioned above.

There are other techniques like the short-time Fourier transformation or neural networks, these techniques are very accurate, a lot of research has been made around them, but it is not relevant to this investigation since they are not used neither in the main project or for the control group for comparison of the results obtained.
The EMG signal is the electrical manifestation of the neuromuscular activation associated with a contracting muscle. It is an exceedingly complicated signal which is affected by the anatomical and physiological properties of muscles, the control scheme of the peripheral nervous system, as well as the characteristics of the instrumentation that is used to detect and observe it [3, 4].

**The motor unit action potential**

Under normal conditions, an action potential propagating down a motoneuron activates all the branches of the motoneuron; these in turn activate all the muscle fibers of a motor unit. When the postsynaptic membrane of a muscle fiber is depolarized, the depolarization propagates in both directions along the fiber. The membrane depolarization, accompanied by a movement of ions, generates an electromagnetic field near the muscle fibers (Figure 1.2).

A detection electrode is used at this point to measure the potential of the field. For technical reasons, the detection electrode is typically bipolar, and the signal is
amplified differentially. The waveform of the observed action potential will depend on the orientation of the detection electrode contacts with respect to the active fibers. [5]

Figure 1.2 – Mechanism of directly gated synaptic transmission at a neuromuscular junction

**Synaptic Transmission at the Neuromuscular Junction**

As shown in figure 1.2 the synaptic transmission happens in a series of steps:

1. An Action Potential (AP) is conducted down the somatic motoneuron down to the synaptic knob.

2. The reversal in electrical polarity at the synaptic knob causes an opening of “voltage-gated” Ca+2 channels (Voltage gated calcium ion channels open and allow calcium ions to flow inside due to the voltage change). Calcium ions are very important for the release of neurotransmitters and secretion of hormones by endocrine cells.

3. Calcium flows into the synaptic knob and it’s this influx that causes these vesicles to form and release/secrete neurotransmitters. In other words, the entry of
Ca+ into the synaptic knob causes the exocytosis (secretion) of the neurotransmitter Acetylcholine (ACh).

4. The ACh diffuses across the synaptic cleft and binds to “nicotinic” ACh Receptor Site Proteins on the membrane of the Skeletal Muscle Cell (Fiber). The first chemical they discovered that affects this receptor site, happened to be nicotine so that’s why it got the name, nicotinic cholinergic receptor.

5. Activation of the ACh Receptor Sites causes an opening of “ligand-gated” Sodium Ion Channels.

6. As sodium ions flow into the Skeletal Muscle Cell, it depolarizes to the threshold potential, triggering an Action Potential.

7. As the action potential spreads along the cell, it causes the muscle cell to contract.

8. The ACh which is attached to the receptor site, is split into acetate and choline by acetylcholinesterase (ACHase), an enzyme of the skeletal muscle cell membrane.

9. The “ligand-gated” sodium ion channels close, permitting the skeletal muscle cell to relax.

10. The acetate & choline are actively transported back-up into the synaptic knob (“Active Reuptake”) to be re-synthesized.

**Signal Characteristics**

The excitability of muscle fibers through neural control represents a major factor in muscle physiology. This phenomenon can be explained by a model of a semi-permeable membrane describing the electrical properties of the sarcolemma. An ionic difference between the inner and outer spaces of a muscle cell forms a resting potential at the muscle fiber membrane (approximately -80 to -90 mV when not contracted). This difference in potential which is maintained by physiological processes (ion pump) results in a negative intracellular charge (figure 1.3) compared to the external surface.
Figure 1.3 – Schematic illustration of depolarization / repolarization cycle within excitable membranes [4]

If a certain threshold level is exceeded within the Na+ influx, the depolarization of the membrane causes an Action potential to quickly change from –80 mV up to +30 mV (figure 1.4). It is a monopolar electrical burst that is immediately restored by the repolarization phase and followed by an after-hyperpolarization period of the membrane. Starting from the motor end plates, the action potential spreads along the muscle fiber in both directions and inside the muscle fiber through a tubular system. The EMG signal is based upon action potentials at the muscle fiber membrane resulting from depolarization and repolarization processes as described above.
The Raw EMG Signal

In kinesiological studies the motor unit action potentials of all active motor units detectable under the electrode site are electrically superposed and observed as a bipolar signal with symmetric distribution of positive and negative amplitudes (mean value equals to zero). It is called an Interference pattern. The two most important mechanisms influencing the magnitude and density of the observed signal are the recruitment of MUAPs and their Firing Frequency.

These are the main control strategies to adjust the contraction process and modulate the force output of the involved muscle. Because the human connective tissue and skin layers have a low pass filter effect on the original signal, the analyzed firing frequency e.g. of a surface EMG does not measure the original firing and amplitude characteristics. For simplicity, one can say that the EMG signal directly reflects the recruitment and firing characteristics of the detected motor units within the measured muscle (figure 1.5):
Figure 1.5 – Recruitment and firing frequency of motor units modulates force output and is reflected in the superposed EMG signal [2]

When the muscle is relaxed, a more or less noise-free EMG Baseline can be seen. The raw EMG baseline noise depends on many factors, especially the quality of the EMG amplifier, the environment noise and the quality of the given detection condition. Assuming a state-of-the-art amplifier performance and proper skin preparation, the averaged baseline noise should not be higher than 3–5 microvolts, 1–2 should be the target. The investigation of the EMG baseline quality is a very important checkpoint of every EMG measurement.

The healthy relaxed muscle shows no significant EMG activity due to lack of depolarization and action potentials. By its nature, raw EMG spikes are of random shape, which means one raw recording burst cannot be precisely reproduced in exact shape. This is since the actual set of recruited motor units constantly changes within the matrix/diameter of available motor units: If occasionally two or more motor units, fire at the same time and they are located near the electrodes, they produce a strong superposition spike. By applying a smoothing algorithm (e.g. moving average) or selecting a proper amplitude parameter (e.g. area under the rectified
curve), the non-reproducible contents of the signal is eliminated or at least minimized.

Raw sEMG can range between +/- 5000 microvolts depending on the patients’ capacity and typically the frequency contents range between 6 and 500 Hz, showing most frequency power between ~ 20 and 150 Hz [6].

**Signal Gathering**

The signal gathering is accomplished by using surface electrodes as mentioned previously. This type of approach has advantages, disadvantages and considerations to be included in all the stages of the design and implementation phase.

**Factors influencing the EMG signal**

On its way from the muscle membrane up to the electrodes, the EMG signal can be influenced by several external factors altering its shape and characteristics. They can basically be grouped in: [3, 4]

a) Tissue characteristics:

The human body is a good electrical conductor, but unfortunately the electrical conductivity varies with tissue type, thickness, physiological changes and temperature. These conditions can greatly vary from subject to subject (and even within subject) and prohibit a direct quantitative comparison of EMG amplitude parameters calculated on the unprocessed EMG signal. Inherent instability of signal comes as a factor of the tissue characteristically found noise. The amplitude of EMG is random in nature as described above. EMG signal is affected by the firing rate of the motor units, which, in most conditions, fire in the frequency region of 0–20 Hz. This kind of noise is considered as unwanted and the removal of the noise is important.

b) Physiological cross talk:

Neighboring muscles may produce a significant amount of EMG that is detected by the local electrode site. Typically, this “Cross Talk” does not exceed 10 %–15 % of the overall signal contents or is not available at all. However, care must be taken for narrow arrangements within muscle groups. ECG spikes can interfere
with the EMG recording, especially when performed on the upper trunk and shoulder muscles. They are easy to see and new algorithms are developed to eliminate them.

c) Changes in the geometry between muscle belly and electrode site:
Any change of distance between signal origin and detection site will alter the EMG reading. It is an inherent problem of all dynamic movement studies and can also be caused by external pressure.

d) External noise:
Special care must be taken in very noisy electrical environments. The most demanding is the direct interference of power hum, typically produced by incorrect grounding of other external devices.

- Ambient noise:
Electromagnetic radiation is the source of this kind of noise. The surfaces of our bodies are constantly inundated with electric-magnetic radiation and it is virtually impossible to avoid exposure to it on the surface of earth. The ambient noise may have amplitude that is one to three orders of magnitude greater than the EMG signal.

- Motion artifact:
When motion artifact is introduced to the system, the information is skewed. Motion artifact causes irregularities in the data. There are two main sources for motion artifact: 1) electrode interface and 2) electrode cable. Motion artifact can be reduced by proper design of the electronic circuitry and set-up.

- Electrode and amplifiers:
The selection/quality of electrodes and internal amplifier noise may add signal contents to the EMG baseline. Internal amplifier noise should not exceed 5 Vrms (ISEK Standards). Most of these factors can be minimized or controlled by accurate preparation and checking the given room/laboratory conditions.

**Signal Filtering**

Decomposition of EMG signal has been done by wavelet spectrum matching and principle component analysis of wavelet coefficients. But for these a clean and
correctly filtrated signal (excessive filtration is not advised, neither is the use of notch filters for EMG signal processing).

EMG-amplifiers act as differential amplifiers and their main purpose is the ability to reject or eliminate artifacts. The differential amplification detects the potential differences between the electrodes and cancels external interferences out. Typically, external noise signals reach both electrodes with no phase shift. These “common mode” signals are signals equal in phase and amplitude. The term "common mode gain" refers to the input-output relationship of common mode signals. The "Common Mode Rejection Ratio" (CMRR) represents the relationship between differential and common mode gain and is therefore a criterion for the quality of the chosen amplification technique. The CMRR should be as high as possible because the elimination of interfering signals plays a major role in quality. A value >95 dB is regarded as acceptable (SENIAM, ISEK).

State of the art concepts prefer the use of EMG pre-amplifiers. These miniaturized amplifiers are typically built in the cables or positioned on top of the electrodes (Active electrodes). The latter pre-amplifier type can have the disadvantage of a bulky electrode detection side with increased risk of pressure artifacts (e.g. when sitting on them) and they typically do not allow free selection of electrode types [8]. The main idea of using small EMG pre-amplifiers located near the detection site is early pick up of the signal, amplification, (e.g. 500 gain) and transmission on a low Ohm level that is less sensitive to (cable) movement artifacts.

An EMG signal that has not been amplified has typical charges between a few microvolts and 2–3 millivolt when reading on the skin. The signal is generally amplified by a factor of at least 500 (e.g. when using preamplifiers) to 1000 (passive cable units). The Input impedance of the amplifier should have a value of at least 10x the given impedance of the electrode.

The frequency ranges of an EMG amplifier (bandpass settings) should start from 10 Hz highpass and go up to 500 Hz lowpass. Any Notch filtering (to cancel e.g. power hum) needs to be avoided because it destroys too much signal information.
Figure 1.6 present a simple scheme of the circuit for EMG signal acquisition.

![EMG Circuit Diagram](image)

Figure 1.6 – EMG circuit used as the base for the design of the portable EMG board used in the control group of this project. (components and values may differ from the final circuit)

### 1.2.3 Tensiomyography

The anatomy-physiological bases for TMG are the same as for EMG and for such reasons this section begins immediately with a description of TMG technology. “Tensiomyography (TMG) has been developed in the late 1980s to evaluate deficient muscle initially, and it was introduced into sports medicine and athletic training. It is a simple to use selective and non-invasive for measuring a skeletal muscle response. The method is based on the measurement of the radial displacement of muscle belly, which is caused by an electrical stimulator. The displacement is measured with an electric sensor which is connected to a computer system. It gives the information of maximal displacement of the belly (Dm) with
following time parameters: delay time, contraction time (Tc), sustain time, and relaxation time. TMG studies usually focus on two common parameters: Tc and Dm. An increase in Tc indicates a muscle with a predominance of slow-twitch fibers. A decrease in Dm indicates an increase in muscle stiffness or tone” [7].

Figure 1.7 shows the common set up of a TMG system. As it can be seen the implementation of such method is rather simple and requires little comprehension of the biological methods involved in the process of muscle contraction.

Figure 1.7 – Components of a basic TMG system including: 1 – Sensor; 2 – Electrodes; 3 – Electrical stimulator; 4 – Notebook or any other user interface.

TMG: Tensiomyography [7]

Bioelectrical signals are not measured at any point rather biomechanical signals are gathered and transformed into an electrical signal using a transducer. This signal is later graphed as a curve displaying the displacement throughout the period of contraction. The basic waveform for a TMG signal is showed in figure 1.8. This signal is comprised by two components: Time and Displacement. The time variable has different moments of interest during the measurement and are all present in the curve; Td, Tc, Ts and Tr which represent the signal’s delay time, contraction time,
sustain time and relaxation time respectively and additionally Dm which represents the maximum displacement value during the contraction.

Contrary to EMG measurements, the TMG measurements are usually and almost exclusively done with a patient in a stationary position and active measurement is used instead, this is due to the size of the sensor and lead. A set of electrodes sends an electrical signal thru the muscle and the sensor is placed in between to measure the corresponding contraction. The sensor measures directly the displacement of the sensor’s lead and gives a signal per such displacement in time. The signal changes depending in the different displacement values reached during a contraction and these can vary from patient to patient, although calibration for the sensitivity of the device can be performed to normalize measurements within trials.

According with the findings in the research carried on the use of TMG as a neuromuscular assessment tool (Rusu et. al, 8) TMG Signals can be successfully used as a measurement tool for ongoing training of muscle hence having dynamic capabilities for gathering of biomedical signals on patients in supine position. These
devices offer high precision and repeatability in the measurements although this is true for the method described before in figure 1.7.

In 2011, a novel method for TMG measurement was developed [9] this sensor allows to measure the pressure generated during contraction over the tip of a sensor placed on top of the muscle. The correlation between the isometric force and MC amplitude showed high individual correlation ($0.97 \leq r \leq 1$) giving an indication of the possibility of this device to be used for muscle mechanic diagnostics and/or control of body activated prosthetics. A simple representation of the MC sensor functioning and structure are presented in figure 1.9. The sensor is constructed in such a way that its pressure on the subject’s skin causes the sensor tip to compress the skin surface and the intermediate layer, ultimately placing pressure on the measured skeletal muscle. Any suitable force meter or pressure meter can be used for measuring the force detected on the sensor tip.

Figure 1.9 – a) MC sensor for determining the mechanical and physiological properties of skeletal muscles: 1 – sensor tip; 2 – force meter; (3) – supporting part; (4) – skin surface; (5) – intermediate layer; (6) – skeletal muscle; b). A simplified representation of the MC measuring principle for the determination of the mechanical and physiological properties of skeletal muscles (1) – sensor tip; (2) – skin and intermediate layer; (3) – measured muscle [9]
1.2.4 Actuation devices and control methods used with prosthetic devices

There are many types of actuation devices used in the field of prosthetics. The two more commonly used are: engines or motors and pulley systems. The choosing of either one depends on the application of the device and the control method but it mainly depends on the socio-economic factors and characteristics of the target market.

**DC motors, stepper motors and servo-motors [10, Ch. 3, 4 and 5]**

*DC motors* are simple to use straightforward devices. They are ideal when the device will be powered by batteries, solar power or a USB interface. The main classification for DC motors is made into brushed and brushless motors, a list of their similitudes can be found in table 1.1. This type of motors works accordingly to Ampere’s Force Law that in simple terms states that “As the current entering the armature of a motor increases, the motor’s torque increases.” As it happens with current and torque, when voltage increases the RPM or rotation per minute (rotational speed) of the motor increases an approximate curve for both behaviors is shown in figure 1.10a and figure 1.10b.

Table 1.1 – Characteristics comparison of brushed and brushless DC motors.

<table>
<thead>
<tr>
<th>Motor</th>
<th>Characteristic</th>
</tr>
</thead>
<tbody>
<tr>
<td>Brushed</td>
<td>Brushless</td>
</tr>
<tr>
<td>✓</td>
<td>✓                              Torque is approximately proportional to current.</td>
</tr>
<tr>
<td>✓</td>
<td>✓                              Speed is approximately proportional to voltage.</td>
</tr>
<tr>
<td>✓</td>
<td>✓                              Control circuitry employs electrical switches to deliver power to the motor.</td>
</tr>
<tr>
<td>✓</td>
<td>✓                              A controller can govern the motor’s operation using PWM (pulse width modulation) signals.</td>
</tr>
</tbody>
</table>

Although they share a lot of similitudes there are some significant changes in their properties and applications that must be considered when choosing between one or the other.
Figure 1.10 – a) Current versus Torque relationship. b) Voltage versus rotational speed (w) relationship for DC motors

As it can be seen in figures 10a and 10b the relationship between a DC motor parameters of speed and torque are almost linearly proportional to the voltage and current supplied and are govern by a constant value $K_V$ and $K_T$ respectively. If a motor’s purpose is to run quickly, select a motor with high $K_V$ and low $K_T$. If its purpose is to provide torque, select a motor with high $K_T$ and low $K_V$.

One of the characteristics of brushed motors is that they have a metal contact to the armature called mechanical commutator which allows the motor to assure a
full revolution, from these motors three types exist permanent magnet DC (PMDC) motors which have a constant magnetic field (and a constant $K_V$) but they tend to lose their magnetization over time. Series-wound DC (SWSC) motors, in a SWDC motor, the field winding is connected in series with the rotor winding, which means the current entering the field winding is the same as that entering the armature, they produce an additional torque increase when the current increases and thus have a higher torque than PMDC motors. And lastly shunt-wound DC (SHWDC) motors in which the field winding is placed in parallel with the armature. This means the voltage across the field winding equals the voltage across the armature. This doesn’t produce as much torque as a series-wound motor, but the torque-speed curve is generally level. That is, the motor can maintain its speed for different amounts of load. The main disadvantages of brushed motors are the brush itself in that contact between it and the rotor at high speed erodes the brush over time. And additionally, this type of motor requires the commutator to rotate along increasing the load over the rotor. On the other hand, they are cheap to make and easy to control.

Brushless (BLDC) motors are more complex and more expensive than brushed motors, but because there’s no mechanical contact between the rotor and stator, they’re more reliable and efficient. BLDCs can be divided into two groups depending on the relative positions of the rotor and stator. If the rotor turns inside the stator, the motor is an in-runner. If the rotor runs outside the stator, it’s an out-runner as illustrated in figure 1.11.

![Figure 1.11 – Inrunner BLDC (low torque, high RMP and high](image-url)
efficiency – left). Outrunner BLDC (lower RPM, higher torque – right)

Servomotors in addition to setting the rotation angle, the controller can configure the servo’s rotational speed and acceleration. This added control is the main advantage of servos over steppers. The main disadvantage is that designing a servo controller is a difficult process. Servo motors can be of any kind, DC or AC, brushed or brushless, and mainly with or without feedback. The most common servo motor is illustrated in Figure 1.12.

Figure 1.12 – Brushed DC Servo motor

1.2.5 Creation of control signals for the actuation device

As seen in previously there are many kinds of actuation devices or final interfaces between the desired action of a patient (biomedical signal) and the response of the device (movement of the prosthesis), for this reason there are also different types of signals that can be used to control each actuation device.

DC motors of whichever kind require not only a power supply to function but a control signal so it can perform in a predefined manner so they allow a “real control”. DC and Servomotors are the best motors for prosthetic applications and so only the control for this two will be explained in this section.

Electrical switches

Simple DC motors can be control with the use of switches. Electrical switches are commonly used for this purpose. Transistors can also be used as electrical switch by using special configurations (Figure 1.13). Ideal electrical switches don’t exist
but we can approximate such a switch with the use of transistors. To be specific, most modern motor circuits rely on metal-oxide-semiconductor field-effect transistors (MOSFETs) or insulated-gate bipolar transistors (IGBTs) to serve as switches.

![Diagram of a motor control circuit](image)

Figure 1.13 – Control of a motor with the use of an electrical switch (left). MOSFET and IGBT real world electrical switches [10]

In general, MOSFETs are better suited to control small to medium size motors and IGBTs are used with bigger motors that demand a higher current [13]. This type of control serves the purpose of turning a motor either fully on or fully off. If the application requires the motor to run to a specific speed or to increase its speed gradually or just in general to have a more “controlled” functioning, pulse width modulation or PMW is a better option to open and close the electrical switch of a motor for precise amounts of time.

**Pulse width modulation**

As mentioned before PWM are pulses sent to the controller of a motor so its switch opens and closes to a specific rate, the current that reaches the motor is
proportional to the period of the pulses, and so control of the motor speed can be achieved; this is illustrated in figure 14. Choosing the right PWM frequency is critical. In making this decision, we have two important factors to consider:

1. If the frequency is too low, the rise/fall of the power reaching the motor will cause it to rotate in a rough, jerky fashion.

2. If the frequency is too high, the pulses will be too narrow to open and close the switches properly.

In addition, the electromagnets will generate heat, decreasing the motor’s efficiency.

There are no clear rules regarding PWM frequency, but many servomotor circuits directed at hobbyists expect a frequency of 50 Hz. This corresponds to a frame of 20 ms. The best place to look for information is the datasheet, but if the frequency isn’t given, it’s safe to assume that the PWM frequency is 50 Hz.

Frequencies between 30 Hz and 20 kHz produce noise within the human range of hearing. If this is a concern, you may want to set the PWM frequency higher than 20 kHz.

A simple scheme of control for a brushed motor is presented in figure 1.14 for both single-direction control and dual-direction control.
Figure 1.14 – a) Pulse width modulation (PWM). b) PWM modulation for a brushless motor with three phases

Brushless motors control is sustained in a similar fashion, with a small difference. Since BLDCs have three phases, each of these phases must be set at any time one to positive or High level (V+), one to negative or Low level (V-) and one floating (F) as illustrated in figure 1.15. The position of the BLDC can be known either with the built-in sensor in the motor (Sensored control) or by measuring the back-EMF signals of the motor (Sensorless control). Servomotors that provide feedback are high-end actuators and are usually much more expensive than their feedback-less or open-loop counterparts.
The scheme of control for a brushless motor is like the one used for a dual-direction control as shown in figure 1.15 with the difference of having to control three phases instead of two. Electronic speed control (ESC) systems are a good option when a prebuilt circuit is required to avoid the complications inherent to the controller’s design. This kind of controller are intent to be used with BLDC motors but there are some built to be used with brushed ones as well. ESC systems allow for the programming of some characteristics, (e.g. the cutoff, timing, break, etc.) usually through USB connection to a PC.

**Servo control [10, Ch. 5, subsections 5.2 and 5.3]**

Servo motors are intended for applications that require high-precision motor control. A system for servo-control can either have an Open-loop or a Close-loop configuration or in other words it can either have or not feedback as illustrated in Figure 1.16.

To solve the equations that arrive during the process of design of servo control the Laplace transform is used, the most common transformations can be seen found online. and a diagram of the closed-loop control in the “s” domain can be seen in Figure 1.17. with its corresponding transfer function for a two-block closed system:

$$\frac{Y(s)}{X(s)} = \frac{H(s)G(s)}{1 + H(s)G(s)}$$  \hspace{1cm} (1)
Where: \( Y(s) \) is the output of the system; \( X(s) \) is the input of the system and \( H(s) \) and \( G(s) \) are components of the black box that transforms signal \( X(s) \) into signal \( Y(s) \).

Figure 1.16 – Open loop and Close loop systems. Where: \( \theta(t) \) – The angle of the servomotor’s shaft; \( r(t) \) – The desired angle of the servomotor (called the reference or the setpoint); \( e(t) \) – The deviation (error) between the motor’s angle and the desired angle; \( v_a(t) \) – The control signal (voltage) provided by the controller

Substituting the terms for the four variables of interest in a servomotor and solving, the relationship within the armature voltage and the shaft angle:

\[
\frac{\theta(s)}{V_a(s)} = \frac{K_i}{JL_a s^3 + (JR_a + BL_a)s^2 + \left(\frac{K_t}{K_v} + R_a B\right)s}
\]

Where \( \theta(s) \) is the servomotor’s angle; \( V_a \) is the servomotor’s voltage; \( K_t \) and \( K_v \) are the servomotor constants and \( R_a \) and \( L_a \) are the passive components of the servomotor. This equation gives us a precise knowledge of how the servomotor responds to armature voltage. Meaning that, we can multiply this expression by a voltage function (transformed to the s-domain) to determine how the shaft angle will be affected. Figure 1.17 shows a closed-loop control system transformed into the frequency domain.
Figure 1.17 – Computing of the transfer function of a closed-loop control system using the Laplace transform

The overall transfer function for the system in figure 1.17 is:

$$\frac{\theta(s)}{R(s)} = \frac{C(s)M(s)}{1 + C(s)M(s)} \quad (3)$$

replacing the servomotor transfer function:

$$\frac{\theta(s)}{R(s)} = \frac{C(s)\left(JL_a s^3 + (JR_e + BL_a)s^2 + \left(\frac{K_t}{K_v} + R_a B\right)s\right)}{C(s)\left(JL_a s^3 + (JR_e + BL_a)s^2 + \left(\frac{K_t}{K_v} + R_a B\right)s\right) + 1} \quad (4)$$

and if PID control is used, $c(t)$ is given as follows:

$$c(t) = K_p e(t) + K_i \int_0^t e(\gamma) d\gamma + K_d \frac{d}{dt} e(t) \quad (5)$$

By controlling the three constant values of a PID control: $K_p$, the proportionality constant – Identifies how the controller should respond to the current value of the error; $K_i$, the integral constant – Identifies how the controller should respond to the sum of the error over time and $K_d$, the differential constant – Identifies how the controller should respond to the current slope of the error; many aspects of the servo’s behavior can be controlled. Figure 1.18 shows the final block scheme for the servomotor control using a PID controller.

Using Laplace transform on equation 4 and replacing it onto equation 3 we receive the following equation:
This final equation gives us the working of the servomotor and determination of its position during this working cycle.

\[
\frac{\theta(s)}{R(s)} = \frac{\left( K_p + \frac{K_i}{s} + sK_d \right) \left( JL_a s^3 + \left( JR_a + BL_a \right) s^2 + \left( \frac{K_t}{K_v} + R_o B \right) s \right)}{\left( K_p + \frac{K_i}{s} + sK_d \right) \left( JL_a s^3 + \left( JR_a + BL_a \right) s^2 + \left( \frac{K_t}{K_v} + R_o B \right) s \right) + 1}
\]

Figure 1.18 – Computing the transfer function of a closed-loop system for the PID control of a servomotor

**1.2.6 Feedback measurement for upper limb prosthetic devices**

Feedback can be defined as a measurement of the output signal of a system that gives information on the system behavior and when compared to a desired or control value, can be used to correct the system behavior in case of errors.

**Feedback measurement on the BES actuation unit [10, 11]**

When speaking about the actuation devices in this case the motors, from those discussed in this work only one of them provides such capability in a reliable way – Servomotors. Usually most servomotors for common daily use do not provide with feedback although other devices can be used to obtain these signals (which will be discussed later) and those servomotors that do offer the possibility to obtain feedback signals (Sensored feedback) are more expensive and difficult to find.

So, to identify the shaft’s angle of a servomotor (feedback) a rotary encoder must be attached to it, this can be found in two types (for use with servomotors):
Optical – A sensor detects light passing through a specially patterned disk and magnetic – A sensor detects the moving poles of a magnet. Both types of encoders are illustrated in figures 1.19 and figure 1.20; although only their main principle of working will be covered.

![Figure 1.19 – Optical Encoder disks used with servomotors](image)

Optical Encoder’s operation is made possible by a disk connected to the motor’s shaft. This disk is transparent in some areas and opaque in others. On one side of the disk, a light source directs light at one portion of the disk. On the other side, an optical sensor measures how much light passes through. The sensor delivers its results to a processor, which may assign a 1 to the presence of light and a 0 to the absence.

The disk on the left has alternating stripes of transparent and opaque regions. As the shaft turns, the optical sensor measures the time between successive flashes of light. The processor uses this to determine how quickly the motor’s shaft is turning. Because it provides speed but not position, this type of encoder is called an incremental encoder.

In contrast, the disk on the right of Figure 1.19 is used by absolute encoders because it identifies the shaft’s angle as well as its speed. In this case, the light shines along an axial stripe that has alternating transparent/opaque regions. The optical sensor detects this light and passes multiple readings to the processor. The microprocessor converts the pattern of light and darkness into a number and uses it to determine the shaft’s approximate angle.
Magnetic encoders are rarer than their optical counterpart, but they are more reliable and provide a better resolution. In these encoders, a circular magnet is attached to the shaft. A magnetic sensor is positioned close to the magnet to detect its north and south poles. As the shaft turns, the sensor measures the locations of the changing poles and determines the shaft’s angle and speed.

**Feedback measurement at the terminal devices**

One of the most common problems found when using a prosthetic device is the lack of reliable feedback. Although these devices are nowadays very reliable and accurate and provide a more trustworthy usage than traditional prostheses. One solution that has been around since 1969 but until the early 90’s computers and electronics cached up to the concept to provide a real application, a device introduced by Paterson et. al. [11] uses a polyvinylidene fluoride piezoelectric film that generates a signal (pressure variations) to then translate them into a corresponding voltage. The results of this investigation tended to confirm their initial hypothesis that the mode of feedback stimulation generated by a grasping prosthetic hand should attempt to replicate the stimulation that one receives from the grip of a natural hand.

Other studies have been carried to measure the applied pressure during grappling of the prosthetic hand but many problems arise from these applications.
New advances in the field of soft robotics look promising as a grasping method for upper limb prosthetics [13, 14].

1.2.7 Patent Search

The application presented in this work for piezo element devices and the comparisons between methods are relatively unknown. Therefore, not many patents concerning this research were found. Three patents related to construction of force piezo element sensors, the abstracts for all of them are presented above:

**US Patent US5571972A: Sensor using piezoelectric elements**

The periphery of a disk having flexibility is fixed to a sensor casing, and a force applied to the central portion is detected. A doughnut disk-shaped piezoelectric element is positioned on the upper surface of the disk, and upper electrode layers indicated by patterns of D1 to D6 are formed on the upper surface of the piezoelectric element. Further, lower electrode layers similarly having pattern of D1 to D6 are formed on the lower surface of the piezoelectric element, and the lower surface of the lower electrode layer is fixed on the upper surface of the disk. Six detection elements D1 to D6 are formed each of which is constituted by a pair of upper and lower electrode layers and a portion of piezoelectric element put there between. Thus, force components exerted at an origin defined in the central portion of the disk in respective axes directions of X, Y, Z can be detected based on charges produced in detection elements D1, D2, detection elements D3, D4, and detection elements D5, D6, respectively. See figure 1.21 for the device schematic.
Figure 1.21 – Apparatus for, or methods of, measuring force, e.g. due to impact, work, mechanical power, or torque, adapted for special purposes for measuring several components of force using piezo-electric means

**US Patent US5209126A: Force sensor**

A force sensor and related method for determining force components. The force sensor includes a deformable medium having a contact surface against which a force can be applied, a signal generator for generating signals that travel through the deformable medium to the contact surface, a signal receptor for receiving the signal reflected from the contact surface, a generation controller, a reception controller, and a force determination apparatus. The signal generator has one or more signal generation regions for generating the signals. The generation controller selects and activates the signal generation regions. The signal receptor has one or more signal reception regions for receiving signals and for generating detections signals in response thereto. The reception controller selects signal reception regions and detects the detection signals. The force determination apparatus measures signal transit time by timing activation and detection and, optionally, determines force components for selected cross-field intersections. The timer which times by activation and detection can be any means for measuring signal transit time. A cross-field intersection is defined by the overlap of a signal generation region and a signal reception region. See figure 1.22 for the device schematic.
Figure 1.22 – Apparatus for, or methods of, measuring force, e.g. due to impact, work, mechanical power, or torque, adapted for special purposes for measuring several components of force

**US Patent US3274828A: Force sensor**

A force sensor comprising, in combination: oscillator means including at least one solid-state element in the form of a resonator having an Eigen frequency which varies when a force is applied thereto; means connected to said oscillator means for detecting a change in frequency when a force is applied to the element, said oscillator means and said solid-state element being arranged so that the element oscillates in the shear mode, and means for transferring forces below a predetermined maximum to said element on a portion thereof where the propagation wave vector of oscillation for a given direction of propagation is at an insignificant value and in the direction of the neutral plane of shear vibration in which plane particle displacement reverses itself, whereby the frequency may be changed by the application of force without changing the q of the resonator, said force transferring means including a force applying element in contact with said element, a force receiving element movable relatively toward and away from said force applying element, and a spring disposed between said force elements. See figure 1.23 for the device schematic.

Figure 1.23 – Measuring force or stress in general using properties of piezo-electric devices using piezo-electric resonators
1.3 Conclusions for the problems found at the knowledge base

In the field of prosthetics, EMG is the main method used for the acquisition of signals coming from the muscles. As mentioned previously the use of EMG signals for the control of a prosthetic device implies many complications and must undergo a high amount of processing to be ready for use. These implications cause the devices that use this method (myocontrolled prostheses) to be expensive to construct and impossible to be made standalone maintenance free devices (understanding maintenance free as a patient can realize it at home with no special equipment).

The goal of this project is to find a more reliable control method that requires less processing and is cheaper to build so it can be made accessible to a larger group of the population. Servomotors are without a doubt the best option to use as actuation devices, they offer the best voltage/torque relationship to the lower price amongst the other options presented in this work. Feedback would be ideal since a closed loop system would indeed improve the systems precision and accuracy as well as improving the hysteresis cycle of the working of the prosthetic device. The methods for obtaining this feedback are not discussed in this work as it would increase the cost of the prototype and financial limitations have been established for the project.

Tensiomyography was considered probably the best test method to create a novel prosthetic device control unit but again the price of myotonic sensors is very high and interferes with fulfilling the fifth objective of this project. Instead, the use of piezo elements is discussed as the measurement system main unit and it is compared with an EMG control board in a similar price range.

To design a Bioengineering System (BES) for the control of a prosthetic upper limb device. Obtaining a control method with enhanced reliability when compared to traditional Electromyography (EMG) control methods and a comprehensive model that accurately predicts the behavior of the device with enhanced reliability. Keeping in mind that the final device must be a low-cost method for controlling prosthetic devices and has a higher or equal accuracy and precision as similar control devices in the same price range or slightly above.
The following tasks are suggested to solve the problems found at the knowledge phase:

1. Design a BES which is portable enough to be embed or attached easily to existing prosthetic devices.
2. The BES must provide a good on-time response and accurately emulate the desired movement the patient intends to perform.
3. The BES can successfully interpret between muscle activation combinations in 90% of the cases.
4. The model accuracy is higher than 85% based on the final waveform and wave characteristics.
5. The final device uses as few components as possible to reduce both size and cost.
CHAPTER 2 PROJECT DESCRIPTION

The project is based on the utilization of the piezoelectric effect to transform deformation into a voltage signal. The principle of construction follows a similar one to the one found in piezoelectric force columns as the distance between each piezo transducer varies to control the dynamic range of the voltage measurements. The main objective is to obtain a real-time trigger control signal for a servomotor or any other electromechanical device at the time a muscle from the body is contracted and the change in the muscle surface stiffness displaces the sensor due to the muscle belly upward motion. This concept will be further explained in the following sections of this chapter.

The Bioengineering system (BES) was developed combining two ideas: The use of piezoelectric sensors to analyze and study muscle tension characteristics [9] and a simple force sensor layout found online using two piezoelectric elements and a rubber O-ring [15]. The piezoelectric element is showed in figure 2.1.

![Piezo Element](image)

Figure 2.1 – Piezo Element 7BB-20-6L0 with dimensions D=20,0 mm, a=14 mm and b=12,8 mm, thickness of the plate 0,42 mm and plate thickness of 0,20 mm

The idea behind the sensor is to increase the inflexion range of the sensor with the rubber O-ring when there is a force applied at the piezo element’s piezo disk (the center white part of the sensor) thus increasing the achievable voltage response range. Additionally, a resistor is connected in parallel between the active part of the
sensor and the ground plate (peripheral metallic part of the sensor), the value of the resistor can be from 1kOhm or 2kOhm for knock sensors and up to 1MOhm to get the maximum voltage range sensitivity [15].

2.1 Model Creation

The object of study for this project is a patient’s upper limb muscle, but observation is made at the piezoelectric element since the force value obtained for a single contraction is not the focus as quantitative measurements are not the goal of the sensor, but to determine if whether it is a voluntary contraction or not.

2.1.1 Piezoelectric effect

In 1880, Jacques and Pierre Curie discovered an unusual characteristic of certain crystalline minerals: when subjected to a mechanical force, the crystals became electrically polarized. Tension and compression generated voltages of opposite polarity, and in proportion to the applied force. Subsequently, the converse of this relationship was confirmed: if one of these voltage-generating crystals was exposed to an electric field it lengthened or shortened according to the polarity of the field, and in proportion to the strength of the field. These behaviors were labeled the piezoelectric effect and the inverse piezoelectric effect, respectively, from the Greek word piezein, meaning to press or squeeze.

This effect varies in piezo elements depending on its structure, material, temperature, etc. including the direction of the force.

Since the voltage will be obtained when the sensor is deformed by the upward force coming from the muscle, the voltage output for disc piezo elements is defined by the following formula:

$$ V = \frac{g_{33} F_t h}{-r^2} $$

(7)

where $g$ is the piezoelectric voltage constant (Vm/N), $F_t$ is an applied force, $h$ is the thickness of the piezoelectric element and $r$ is its radius [16]. See appendix A for other relationships of voltage and the piezo element characteristics depending on its shape.
2.1.2 Piezoelectric force sensor

For the creation of the force sensor the idea was taken from an online do it yourself project [16], here two piezo elements were placed back to back with a rubber O-ring in between. For this project, all possible connection configurations were tried to find the one that provided the best signal possible and both piezo elements were connected at one point on contraire of the original project by bridell. Resistors R of 1MOhm and 820kOhms were connected in parallel between the rim and the center of the piezo element, both prove alright to use the piezo with a fairly large range of voltage values.

Additionally, the sensor was put inside a large shrink tube. The final construction can be seen on figure 2.2.

![Figure 2.2 – Final piezoelectric force sensor constructed with two back to back ceramic piezo elements and a rubber O-ring inside a shrink tube embebed in a velcrom band to fix to the body](image)

The final connection configuration for the sensor is presented in figure 2.3. As it can be seen the rim of one piezo element and the center of the other are connected, this is to get a better response time and to more accurately determine the inflexion moment and improve the system response, this is done instead of measuring both piezo elements separately as suggested on bridell’s website.
Figure 2.3 – Connection configuration of the final piezoelectric force sensor

The sensor is constructed with two leads, one positive to the center of the upper piezo element, and one negative connected to the ring of the lower piezo element, additionally the rim of the upper piezo element and the center of the lower piezo element are soldered together. This configuration was obtained after testing the other two possible connections, rims together and measure at the center (connection 2) or centers together and measure at the rims (connection 3). The signals obtained for each of these latter configurations had either a low amplitude but with a good response time (connection 2) or had a high amplitude but with ripple and artifacts from additional forces during each contraction as well as eco waves from the muscle contraction upwards and downwards moments for beginning and the end of the contraction (figure 2.4).

Figure 2.4 – Additional connection configurations tested for the initial setting of the force sensor and their respective output signals

As can be seen in Figure 2.5, these two optional configurations give either a low amplitude but stable signals or a high amplitude signal with a lot of noise.
Therefor connection 1 was chosen as the default measuring method when using these sensors for the present project.

![Oscilloscope graph with labeled connections]

Figure 2.5 – Output signals for all configurations tested for the piezo-force sensor. Connection 1, 2 and 3 respectively

### 2.1.3 Electronic model of a piezoelectric force sensor

To understand better how the sensor works, a model of the same is presented above in figure 2.3. When modelling a piezo element, it is important to have in mind in which portion of its frequency response curve we are working on [17], as well as where do we desire it to be. If the sensor works in the usable region of the frequency response curve (Figure 2.6) the piezo element can be modelled as a voltage source in series with the sensor’s capacitance $C_s$ (Figure 30b). If it works in the high-pass region it will show only high frequency noise from the sensor contact with the skin surface.
Figure 2.6 – Rough sketch of piezoelectric sensor frequency response (relationship between force input and voltage output vs frequency). Low frequencies are filtered out by the leakage resistance, and high frequencies resonate. By Omegatron – The source code of this SVG is valid. This vector image was created with Inkscape by user Omegatron., CC BY-SA 3.0 [16]

Otherwise we must add an inductance due to the seismic mass and inertia of the sensor itself, a capacitance $C_e$ inversely proportional to the elasticity of the sensor, also the static capacitance of the transducer $C_0$ and $R_i$ which represents the leakage resistance of the transducer element as shown in figure 2.7a. [16]

Figure 2.7 – a) Schematic symbol and electronic model for a piezoelectric sensor; b) Simplified schematic model for a piezoelectric device. Original image by Omegatron - The source code of this SVG is valid. This vector image was created with Inkscape by user Omegatron. This SVG electrical schematic was created with the Electrical Symbols Library, CC BY-SA 3.0 [16]
The signal obtained from the sensor showed in figure 2.2 can be seen in figure 2.8–right. The signal is a mixture of both the signal from the upper piezo element and the lower piezo element. To manage this signal a simple comparator using operational amplifiers can be used as shown in figure 2.9. This will transform the signal from the sensor into a pulse wave signal (Figure 2.8–left) with an amplitude equal to that of the voltage supply (according to the ideal working properties of operational amplifiers in comparator configuration) that can be used as an enable signal after some processing for a servomotor drive board. To improve the output signal, DC offset can be removed using a capacitor at the signal output lane together with a limiting resistor. Although DC offset will present depending in the coupling characteristics of the completed final board.

To test the circuit configuration the sensor was attached to the arm, specifically to the brachioradialis muscle (Figure 2.10) and the subject was asked to realize intermitting contractions to stablish a threshold value for the comparator reference voltage. The results for 50 measurements to a female and a male patient are shown in chart 2.1.
Figure 2.9 – Voltage comparator configuration with operational amplifiers. For this project was used a supply voltage of 5V and a reference voltage of 500mV to 1V is acceptable depending on the initial signal of the sensor and whether DC filtering was done before or after the comparing stage [18]

Chart 2.1 – Piezo element force sensor results for 50 repeated measurements with two patients in two locations
After evaluation in three muscles: the brachioradialis muscle, the triceps brachii and biceps brachii, the first two were found best suited to deliver good signals for this application and were chosen as the object of research, the biceps upward force lacks a high enough dynamical and explosive characteristic to affect or deform the piezo element in an enough manner. See figure 2.10.

Figure 2.10 – Muscles of the posterior arm and forearm (left). Muscles of the posterior forearm (right)

In figure 2.10, the muscles used to test the sensors (triceps brachii and brachioradialis) are highlighted. These muscles were chosen because of their size, contractile characteristics and location besides their relatively large distance and dissociation between them both. This also suggest that above elbow amputations are considered for the application of this project as the target population to receive prosthetic devices controlled with this method.

Because of the nature of the sensors (mechanical sensors) and their working principle (piezoelectric effect), for them to provide an electrical signal to measure, they must suffer from a large enough deformation, as explained before the purpose of the O-rings is to provide the space and subjection for this deformation to happen; nonetheless if the muscle mechanical strength during contraction does not create a
large displacement upward force in the contact point (muscle belly, see figure 2.11). In EMG, the biceps brachii is the muscle usually chosen to do these measurements, this is because of its large size and is also a rather easy muscle to control for the patient, but the muscle is very hard to contract by itself and usually involves the triceps brachii and/or other muscles contract along it; this creates a problem when trying to sense different muscles.

A second problem arrives when speaking about the upward force the muscle provides, although transversal contraction is large and generates a large current is created and measured with the EMG electrodes, this is not true for the mechanical upward motion generation at the muscle belly, for this reason insufficient mechanical stress is placed over the piezo sensor and the incidence of false negatives increases, if subjection is increased to overcome this problem, the amount of false positives also increases because of the additional noise from the initial distortion characteristics.

These problems are all solved by choosing the muscles mentioned above. The sensors must be correctly located in the center of the muscle belly were, the upward force peaks in value.

Figure 2.11 – Sensor placement in the muscle belly to access the point of maximum inflexion [19]
This is the main reason why the brachioradialis muscle and the triceps brachii were chosen. Given their relative mechanical independence from one another and the good mechanical upward motion they generate, overcoming the problems found with sensor placement at the biceps brachii.

The final circuit board for the sensor is showed below in figure 2.12.

![Diagram of the electronic model of the piezoelectric force sensor and the output comparator to create the pulse signal](image)

Figure 2.12 – Electronic model of the piezoelectric force sensor and the output comparator to create the pulse signal

The circuit shown in figure 2.12 is the complete electronic model for both piezo elements, as mentioned before they are connected from the negative lead of the upper piezo element and the positive lead of the lower piezo element. The output of the piezo element is connected to the non-inverting input of an operational amplifier set as a voltage comparator, the reference voltage is obtained from the voltage divider form by the resistors R1 and R2 to be set accordingly to the initial voltage level of the sensor and the threshold value for each subject is measured and determined to provide a good estimation of the reference voltage needed to decrease the incidence of false positives and movement artifacts. The DC filter form by the capacitor C and the resistor R can be placed either at the final stage of the circuit or after the sensor, this filter is only necessary if a high DC offset level is found during measurements, after some test and error it was found that incorrect placement of the
sensors may induce this kind of offset signal and can also produce a AC ripple background noise in the signal. As the result of this findings the strap must be placed and marked for the optimal contraction that is beneficial for the measurements and comfortable to the patient.

2.1.4 Sensor mathematical model description

As mentioned before, quantitative evaluation is of no interest at this stage of the project, but it does not mean that it will not be in a near future. Further development of the piezoelectric control unit contemplates the possibility of embedding several sensors into one sleeve to create a pressure map of the remaining limb for study, analysis and better control. Additionally, some sort of conversion from a voltage value to a force/pressure value is to be expected to be able to evaluate patient’s performance and energy demand from the control device.

To obtain such measurements, a mathematical model from the electronic model of the circuit is presented below in the case the sensor voltage cannot be calculated using the formula presented in the equation one.

Considering the electronic model of the sensor presented in figure 2.7a, the following equation regarding voltage can be obtain using simple circuit analysis from the RLC circuit (Figure 2.13) following the voltage supply in this equivalent circuit.

![Equivalent circuit of the RLC circuit obtained from the sensor electronic model](image)

Figure 2.13 – Equivalent circuit of the RLC circuit obtained from the sensor electronic model
The voltage between the lead of $Z_{eq}$ (the equivalent resistance of the load) represents the output voltage of the sensor, and can be calculated using the voltage relationship of resistors in series for $Z_{ce}$ (the capacitor’s impedance) and $Z_{lm}$ (the inductor’s impedance):

$$V_{out} = V_{Zeq} = V \left( \frac{Z_{eq}}{Z_{ce} + Z_{lm} + Z_{eq}} \right)$$  \hspace{1cm} (8)

Where $Z_{eq}$ is the impedance form by the parallel of $C_o$ and $R_i$:

$$Z_{eq} = \frac{1}{Z_{Ri}} + \frac{1}{Z_{Co}} = \frac{Z_{Co} + Z_{Ri}}{Z_{Ri}Z_{Co}}$$  \hspace{1cm} (9)

Substituting equation 8 in 9:

$$V_{out} = V \left( \frac{Z_{Co} + Z_{Ri}}{Z_{Ri}Z_{Co} \left( Z_{ce} + Z_{lm} + \frac{Z_{Co} + Z_{Ri}}{Z_{Ri}Z_{Co}} \right)} \right)$$

$$V_{out} = V \left( \frac{Z_{Co} + Z_{Ri}}{Z_{Ri}Z_{Co}Z_{ce} + Z_{Ri}Z_{Co}Z_{lm} + Z_{Co} + Z_{Ri}} \right)$$  \hspace{1cm} (10)

Finally, we substitute the impedances with their respective equations for capacitors, inductors and resistors and we solve for $V_{out}$.

$$V_{out} = V \left( \frac{1}{jwC_o + Ri} \right)$$

$$V_{out} = V \left( \frac{Ri}{w^2C_oCe + \frac{RiLm}{C_o} + 1} + \frac{1}{jwC_o + Ri} \right)$$

$$V_{out} = V \left( \frac{(jwC_o + Ri)(jw^2C_oCe)}{jwC_o(jRi + jw^2CeRiLm + wCe + jw^2CoCeRi)} \right)$$

$$V_{out} = V \left( \frac{(jwC_o + Ri)wCe}{wCe \left( \frac{jRi}{wCe} + jwRiLm + 1 + jwCoRi \right)} \right)$$

$$V_{out} = V \left( \frac{Ri + jwCo}{1 + \frac{jRi}{wCe} + jwRiLm + jwCoRi} \right)$$

$$V_{out} = V \left( \frac{\left( C_o + \frac{Ri}{jw} \right)}{\left( \frac{1}{jw} + \frac{Ri}{Ce} + RiLm + CoRi \right)} \right)$$  \hspace{1cm} (10)
If we assume that we are working in the usable part of the piezoelectric working frequency (Figure 2.6) then we can use the circuit presented in figure 2.7b resulting in the following equation assuming a load $R$ at the output:

$$V_{out} = V_{load} = V\left(\frac{Z_{Cs} + Z_{load}}{Z_{Cs}Z_{load}}\right) = V\left(\frac{1 + jwCs}{R_{load}}\right)$$

$$= \left(\frac{1 + jwCsR_{load}}{R_{load}}\right)V \quad (11)$$

in a system without load $V_{out}$ is simply a function of the capacitor $Cs$ as proved by making zero all parasite components in equation 2.5.

$$V_{out} = jwCsV \quad (12)$$

If we even equations 7 and 12 considering a load (parallel resistor connected between the piezo element and the metallic ring), we can obtain an equation for the force in terms of the inner capacitance of the piezo element and its design characteristics:

$$Ft = V\left(\frac{j_r^2(1 + jwCsR_{load})}{g_{33}R_{load}}\right)$$

$$Ft = -\frac{V}{g_{33}}\left(w_r^2Cs - \frac{j_r^2}{R_{load}}\right) \quad (13)$$

Equation 13 is useful when the voltage to force curves are not available from the piezo element manufacturer and describes the mathematical model of the force sensor. The output voltage is directly dependent to the force applied and will also depend on the piezo element construction material characteristics, as well as the frequency of the deformation signal and the dimensions of the piezo element. The capacitance $Cs$ can be easily calculated using the standard formula for capacitance for parallel plates [20, 21].

The two piezo elements are connected one to the other but measure different force moments, and have their own part in the signal during muscle activity, thus
been the calculations for equation seven valid for each of them because their effects are additive and not multiplying affects.

### 2.1.5 Signal treatment board

As seen in figure 2.8-left, the signal obtained from the sensors is as that one obtained in myotonic studies, a very defined bell-like shape with a rising part with a width equal to the time delay that the piezo element needs to detect the contraction or movement from the muscle (action to reaction time), a maximum value which is proportional to the maximum deformation that the piezo element suffers, followed by a downhill part with a width equal to the time it takes to the piezo to normalize or go back to its natural state (relaxation time).

The signal presents little to none parasitic elements (noise), for this reason filtration is not required unless a more detailed study like pressure profile estimation is require; and so, the output from the sensors can be directly used as the input for the operational amplifier (op amp) configured as a comparator, the signal obtained is a clean square signal with +5 V peak voltage and a 0 V voltage base as seen in figure 2.8-right. The 555IC circuit works at a +9 V level, for that reason a voltage converter will be used to supply the +5 V needed to obtain the digital high level.

To control the servomotors a circuit based on a monostable 555IC circuit is used in this project. The interaction within this circuit, the control signal from the comparator and the servomotors will be discussed further in the next section of this paper.

### 2.2 Control Unit

The actuation unit is the part of the circuit in charge of taking the control signal coming from the treatment board and translating it into a signal fit to control the motor or actuation unit. The simplest and cost/effective method of controlling a servomotor is using a monostable 555IC circuit configuration (Figure 2.14). This board generates a constant PMW signal which pulse width is controlled by changing the configuration of the series resistor formed by R1 and R2. Commonly R2 is changed for a potentiometer to manually change the pulse width and thus the
position of the servomotor. Pinout for the 555IC can be found in appendix B.1 (Circuit components pinouts).

For this project, the actuation unit oversees the control of the individual movements of the prosthetic hand. A mechanical mechanism as those found in hook prosthetics it thought for the control of the close and open motions, as it is considered a very precise method and provides the patient with a certain degree of feedback during grasp.

Figure 2.14 – Monostable configuration for the 555IC used to control servomotors. Potentiometer R2 is used to change the series resistor value and decrease or increase the pulse width of the output signal.

For this project R2 – a potentiometer which controls the series resistance value and hence the voltage divider form with R1, this value as mentioned before controls the width of the output of the 555IC circuit – is proposed to be replace by either a digital potentiometer or a multiplexed array of resistors, in both cases the resistance level will be modified using the pulse coming from the treatment board. The diodes between the leads of the potentiometer are used to regulate the reverse current flow and prevent unprovoked triggering, in other words it helps stabilize the circuit.
functioning. The results obtained from both these configurations are presented in chapter four. The circuits for the digital potentiometer and the multiplexor resistor array are presented in figures 2.15 and 2.16 respectively.

![Functional Block Diagram for a Digital Potentiometer](image)

**Figure 2.15 – Functional Block Diagram for a Digital Potentiometer from Texas Instruments used as R2 to control the pulse width of the 555IC output [22]**

Figure 2.15 shows the functional block of TI TPL051 100kΩ digital potentiometer (DP), which has 3 inputs for interface (SPI): SCLK, DIN and CS, and three pins for the output connections to the potentiometer, although, this figure can be used to generalize the functional diagram of DPs in general. For this application, the DP will be used in floating rheostat mode, as it fits the connection depicted for the potentiometer R2 in the 555IC control circuit presented in figure 2.14.

The TPL0501 uses a 3-wire SPI compatible serial data interface. This write-only interface has three inputs: chip-select (CS), data clock (SCLK), and data input (DIN). To enable the serial interface and clock data synchronously into the shift register on each SCLK rising edge CS must be driven low. After loading data into the shift register, CS must be driven high to latch the data into the appropriate potentiometer control register and disable the serial interface. During the entire serial data stream CS must be kept low to avoid corruption of the data [22]. See Appendix C for the rheostat mode configuration of the TPL1051 and the SPI interface write sequences, timing diagram and potentiometer measurements.
Figure 2.16 – Resistors array with multiplexor to substitute R2 for the control of
the pulse width of the 555IC output and truth table [23]

The circuit presented in figure 2.16 works in a simple manner. All resistors are
c connected together at one end, this end represents R2 connection to diode 1N914’s
erg negative terminal in the circuit presented in figure 37. The output of the

multiplexor represents the other pin of R2 that is connected to R1. By changing the
input select “a” or “b” (one remaining always constant) with the value from the
sensor, the value of R2 can be changed between two values (100kΩ or 1kΩ) to
modify the width of the output pulse from the 555IC circuit.

2.3 Servomotor control

As mention in chapter one subchapter D, there are many types of servomotors, for this
project I used the MG995 open loop DC servomotors (See appendix B.2 for
details). These servomotors are low cost, low end, easy to use devices which are
controlled with a single input signal and can be fed with voltages from +5 V.
To control the servo's, position the output of the actuation unit (555IC circuit) will be used as the input of the servo control pin to rotate the same and control the position of the shaft.

![Diagram of a state transition between servomotor positions](image)

Figure 2.17 – State diagram for the servomotor position depending on the sensor output for the control of the finger positions (except thumb).

Made with: draw.io online tool

The idea is to use the select inputs of the multiplexor resistor array or the digital potentiometer to do this and that the same signal will trigger the opposite action from the same part of the prosthetic device, depending on the current position and the sensor input combinations according to the circuit’s in figure 2.16 truth table. One servomotor will control a group of two fingers except for the thumb which will act independently from the others; and both servos (for the two finger groups) will be control with an input combination from two sensors. The thumb will be control using the remaining possible combination (a=1 and b=1). To improve the control of the fingers and avoid clumsy movements a set of stabilizers like the one on figure 41 is used to maintain a horizontal pull and avoid the formation of angles when the “tendons” are pulled by the servomotors. This idea was taking from a prosthetic device found in open bionics’ website [24]. See figure 2.18.
The purpose of this element is to have a 0° (180°) angle from the vertical to maximize the force value in the direction of the force according to the formula:

\[
F_{1,2,3} = T_{1,2,3} \times \cos \theta
\]  

(13)

Many prosthetic devices suffer from force losses because of the inappropriate distribution of the tension of the strings connecting the motor unit and the prosthetic fingers. This also provokes the strings to deform and lose both flexibility and rigidity, making them more prone to rolling in the shaft of the servomotor by coming out of the holding structures. To avoid this the stabilizers in figure 2.18 are used to translate this movement to a more efficiently distribute one as mentioned before.

The second variant for the control of the servomotor’s position using a digital potentiometer is presented in figure 2.19.
2.4 Mechanical mechanism for the control of the open and close motions

As it was mentioned in previous sections, the control method presented in this work is meant to be used in addition with a mechanical prosthetic to enhance reliability and increase functionality. All mechanical prosthetic devices have a similar working process, for that reason a body-powered prosthetic from Ottobock us used as an example; in this device, a single line is used to transmit the movement of the shoulder (it can be the shoulder in the same side of the device or the opposite shoulder) that is why the term “body-powered” is used. A picture of a body-powered prosthetic device from Ottobock is presented in Figure 2.20 with two attachments, a hook for precise movements and holding small and slim objects, and a hand for holding wider and bigger objects.
Additional control of each finger will be provided by the piezo element force sensor control unit. With two sensors, a total of three additional movements can be configured. As the number of sensors increases, the amount of possible movements increases as well by the power of two minus one:

$$\text{Number of movements} = (\text{number of sensors})^2 - 1$$ (14)

This equation is used for the determination of the number of possible commands available to the device depending on the number of placed sensors.

**2.5 Final system’s block diagram for the solving of the problems presented in chapter 1**

In this section, a summary of the chapter is presented, additionally the final block diagram for the system in shown in figure 2.21 with its respective explanation.
Figure 2.21 – Block diagram for the bioengineering system with 4to1 MUX unit used to control a prosthetic device with enhanced reliability.

Made with: draw.io online tool

The process begins with the object research, as established before, the brachioradialis and the triceps brachii muscles. The signal gathering unit is comprised by the two piezo element force sensors, as mentioned more sensors can be attached to increase either precision or functionality or both. The sensors are deformed when the patient realizes a contractile movement in either the brachioradialis (brm), the triceps brachii (tbm) or both muscles. The biceps brachii (bbm) can be used with a third sensor as a reference sensing point since the generated force generated from it is different during the contraction of the either one of the remaining two muscles (brm, tbm).

The contraction signal (deformation) is converted into an electrical signal due to the piezoelectric effect. This new electrical signal is then sent to a comparator (not
depicted in the block diagram as it is embedded onto the sensor) and a square signal is obtained.

These square signals are used as the select inputs for the 4to1 multiplexor, which will select one of the resistor inputs according to select input combination and the truth table of the multiplexor. This resistor value will be sent to substitute R2 in the 555IC control unit changing the parameters of the voltage divider and thus the width of the 555IC pulse wave output.

As mentioned in chapter one subsection D, servomotors are pulse wave modulated (they use PWM for control of the shaft position), the change in the pulse width of the signal will change the position of the servomotor and at the same time, the position (close/open) of the finger connected to it. Three servomotors are proposed to be used, one controlling a group of two fingers, another for a group of the remaining two fingers and one for the control of the thumb.

2.6 Experimental results obtained with designed control device

The analytic measurements for the experiments realized were taken from Cornell’s University performance measures for machine learning online curse [26]. The following formula is provided for accuracy calculations:

\[
Accuracy = \frac{a + d}{a + b + c + d}; \text{a, b correct values; c, d mistakes} \quad (2.10)
\]

Other useful measurements are sensitivity of the device, specificity and their confidence intervals. (95% confidence interval will be used in this project for all values.)

The following relationship is used to construct the accuracy threshold curve for the data:
Additionally, MedCalc online application [27] for diagnostically test result calculations was used to obtain the values for specificity and sensitivity.

**2.6.1 Experiment Description and practical results**

Two experiments were carried out utilizing the signal gathering unit, the processing unit (in charge of transforming the signal from the sensors into a control signal) and the control unit described in this project. The tests were carried with only healthy individuals since, time and resources were limited. Additional testing is advised before using with real patients. The experiments are described as follows:

**Accuracy calculation of Signal Gathering Unit (SGU) response under muscle movement**

Number of subjects: 2

Subject 1: Female, 26 years old, Caucasian, height: 159 cm, weight: 49 kg, healthy, normal muscle tone and in a good overall physical condition.

Subject 2: Male, 26 years old, Latin-American, height: 171 cm, weight: 77 kg, healthy, normal muscle tone and in a good overall physical condition.

Number of repetitions: 40 each

Experiment description:

The piezo force sensor was placed on the patients brachioradialis muscle and triceps brachii one at a time. He was asked to perform two Tasks: activate the sensor
willingly, realize movements without activating the sensor (e.g. move the arm, flex the arm and the wrist, contract other muscles of the arm), to measure the amount of true positive and true negative results achievable with a given time frame of 30 minutes. The results of this experiment in both locations – the brachioradialis and triceps brachii muscles – are presented in Table 2.1.

Table 2.1 – Results obtained from the SGU response experiment

<table>
<thead>
<tr>
<th>Patient 1. Location: Brachioradialis</th>
<th>Controlled contraction</th>
<th>Random movement</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sensor active</td>
<td>18</td>
<td>1</td>
</tr>
<tr>
<td>Sensor inactive</td>
<td>2</td>
<td>19</td>
</tr>
</tbody>
</table>

Accuracy: 0.925 (92.5 %); Sensitivity: 0.9 (90 % with 68.30 % to 98.77 %, 95 CI); Specificity: 0.95 (95 % with 75.13 % to 99.87 %, 95 % CI)

<table>
<thead>
<tr>
<th>Patient 1. Location: Triceps brachii</th>
<th>Controlled contraction</th>
<th>Random movement</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sensor active</td>
<td>16</td>
<td>0</td>
</tr>
<tr>
<td>Sensor inactive</td>
<td>4</td>
<td>20</td>
</tr>
</tbody>
</table>

Accuracy: 0.9 (90 %); Sensitivity: 0.80 (80 % with 56.34 % to 94.27 %, 95 CI); Specificity: 1 (100 % with 83.16 % to 100 %, 95 % CI)

<table>
<thead>
<tr>
<th>Patient 2. Location: Brachioradialis</th>
<th>Controlled contraction</th>
<th>Random movement</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sensor active</td>
<td>20</td>
<td>4</td>
</tr>
<tr>
<td>Sensor inactive</td>
<td>0</td>
<td>16</td>
</tr>
</tbody>
</table>

Accuracy: 0.9 (90 %); Sensitivity: 1 (100 % with 83.16 % to 100 %, 95 CI); Specificity: 0.8 (80 % with 56.34 % to 94.27 %, 95 % CI)

<table>
<thead>
<tr>
<th>Patient 2. Location: Triceps brachii</th>
<th>Controlled contraction</th>
<th>Random movement</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sensor active</td>
<td>20</td>
<td>7</td>
</tr>
<tr>
<td>Sensor inactive</td>
<td>0</td>
<td>13</td>
</tr>
</tbody>
</table>

Accuracy: 0.825 (82.5 %); Sensitivity: 1 (100% with 83.16% to 100%, 95 CI); Specificity: 0.65 (65% with 40.78% to 84.61%, 95% CI)

Control Unit (CU) response

Number of subjects: 2
Subject 1: Female, 26 years old, Caucasian, height: 159 cm, weight: 49 kg, healthy, normal muscle tone and in a good overall physical condition.

Subject 2: Male, 26 years old, Latin-American, height: 171 cm, weight: 77 kg, healthy, normal muscle tone and in a good overall physical condition.

Number of repetitions: 40 each

Experiment description:

A number was assigned to each of the prosthetic device finger groups from one to three. A randomly generated list of 40 numbers comprised with only ones, twos and threes was given to each participant. The participant then tried to realize the correct movement using a combination of contractions with the muscles been sensed. Table 2.2, show the list of combinations to achieve the movement of a finger group. The experiment was double blinded, the observant merely annotated the results seen at the prosthetic device (realized movement) and did not knew the list of combinations the patient received and the patient as well could not see the prosthetic device to know if he realized the correct movement for him to focus only in contracting the right muscles.

Table 2.2 – Combinations possible with the SGU and the control unit

<table>
<thead>
<tr>
<th>Number</th>
<th>Digital combination</th>
<th>Correspondent movement</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>01</td>
<td></td>
</tr>
<tr>
<td>2</td>
<td>10</td>
<td></td>
</tr>
<tr>
<td>3</td>
<td>11</td>
<td></td>
</tr>
</tbody>
</table>

The digital combination refers to sensors 1 and 2 active (1) or inactive (0) state.

An interval of about 30 seconds to 1 minute was left between measurements to allow the device to normalize between each contraction. The table of results is presented below. A complete example table with all results of what the randomized task can be, ca be found in appendix D.

Table 2.3 – Example table for the results obtained from CU response experiment
### Measurements

<table>
<thead>
<tr>
<th></th>
<th>Predicted Positive</th>
<th>Predicted Negative</th>
</tr>
</thead>
<tbody>
<tr>
<td>Positive</td>
<td>NC</td>
<td>NC</td>
</tr>
<tr>
<td>Negative</td>
<td>NC</td>
<td>NC</td>
</tr>
</tbody>
</table>

#### 2.6.2 Control Group

The results obtained using the EMG board for the control group, are presented in the table below:

<table>
<thead>
<tr>
<th></th>
<th>Controlled contraction</th>
<th>Random movement</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Patient 1. Location:</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Biceps brachii</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Sensor active</td>
<td>16</td>
<td>6</td>
</tr>
<tr>
<td>Sensor inactive</td>
<td>4</td>
<td>14</td>
</tr>
</tbody>
</table>

Accuracy: 0,75 (75 %); Sensitivity: 0,80 (80 % with 56,34 % to 94,27 %, 95 % CI); Specificity: 0,70 (70 % with 45,72 % to 88,11 %, 95 % CI)

<table>
<thead>
<tr>
<th></th>
<th>Controlled contraction</th>
<th>Random movement</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Patient 2. Location:</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Biceps brachii</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Sensor active</td>
<td>15</td>
<td>7</td>
</tr>
<tr>
<td>Sensor inactive</td>
<td>5</td>
<td>13</td>
</tr>
</tbody>
</table>

Accuracy: 0,70 (70 %); Sensitivity: 0,75 (75 % with 50,90 % to 91,34 %, 95 % CI); Specificity: 0,65 (65 % with 40,78 % to 84,61 %, 95 % CI)

These results are for the RAW signal of each sensor, the sensor designed for this project and the EMG sensor designed to be used as the control group.

#### 2.7 Conclusions for the application part of the project

The final device design is comprised of a piezoelectric sensor, the sensor sends the output signal to a comparing stage to provide a square signal. This signal is taken to drive a IC555 circuit that controls the actuation unit made with servomotors.

The results obtained after testing are proof that the method is a valid method for the control of prosthetic devices. The control signals obtained drive precisely the
servos and have a very small margin of error (10% ±1 error rate) which decreases when making additional measurements.

When compared with an EMG method in a similar price range the gathering of the RAW signal was found to be easier and more precise. The signal has less noise and so requires little to none filtering. The EMG signal provides more physiological information, but for controlling a third device where information about the muscle condition is not required or it is not important, the piezo force sensor provides an easier to process signal and at a quarter of the price.
CHAPTER 3 SPECIAL ASPECTS OF SAFETY

This device is classified as a Class III device with basic insulations from patient and a floating connection (F-type isolation) of the electrical parts of the device between the signal gathering unit and the control unit (by means of the electrical conversion from a raw signal to a pulsed signal), generating a very low voltage, below the minimum specified for patient protection as well as using a divided grounded circuit for both the analog and the digital parts of the device and is a component with high-integrity characteristics meant to be used under continuous operation in a unfixed manner (as the straps can be detached by the user at any time) internally powered by a battery pack with under the minimum leakage current values and mechanical isolations between any sharp edges of the sensors and the patients skin to prevent hazardous situations (e.g. skin rupture) under normal use. The device is portable and requires no programming, but only scheduled calibration from service personnel if malfunction or high error rate is present. As defined by IEC 60601-1-2012: “general requirements for basic safety and essential performance” standard.

3.1 Description of environmental conditions

The device is meant to be used during operation of a prosthetic upper limb device. This statement reassures the fact that different environmental conditions might be met during the lifespan of the device as an average person usually puts himself in very different situation and in various scenarios. All testing for the proper working of the device were carried in control environments indoors and estimations about the effect of external inputs is considered only within the frame of the piezo elements own characteristics as described by the manufacturer. The list of standards to be met according to the application of the device are presented in the following sections of this chapter.
3.1.1 Temperature, humidity and atmospheric pressure ranges

The device was tested in an indoor environment (+18 degrees °C and R.H of 55 %) and outdoors (spring, +5 degrees °C and R.H of 75 %) at a normal atmospheric pressure of approximately 1 atm.

According to the piezo element datasheet (component part No. 7BB-20-6L0), the device working and storage conditions are:

- Temperature: -10 to +40-degrees °C
- Humidity: 15 to 85 % R.H.

As for the atmospheric pressure conditions, the device operates as any other electronic component and works correctly at any altitude having little to none significant changes with slight (±0.11 atmospheres).

Considering that an average person doesn’t undergo any extreme conditions, the device is safe to use in a wide range of scenarios. Extreme conditions are not advice (eg. deserts, extremely cold places like snow mountains, very high-altitude peaks, etc.) as testing has not been carried out for such working conditions and device operation and behavior is unknown.

To test the outer isolation of the sensor the ball-pressure test was carried out, the test is described in Figure 3.1.

Figure 3.1 – Ball-pressure test under a 75ºC temperature to test temperature resistance of the insulation

The test consists in placing a 20N force steel ball over the insulated part of the device in a +75 ºC temperature environment during a certain amount of time
(1 hour). The diameter of the impression made by the ball was measured and found to be result according to subsection 8.8 of the same standard a value greater than 2 mm in diameter constitutes a failure.

The Piezo elements used in this project also has the characteristic to produce heat when under extreme amounts of deformation for long periods of time or when exposed for prolonged resonating states due to a large voltage application between its terminals. But it does not surpass or exceeds the allowable temperature values presented in table 22 of Section 11 subsection 11.1 of the IEC 60601-1-2012 standard.

3.1.2 Indoor or outdoor operation

The device is meant to be used in both indoor and outdoor operation. This last is true if outdoor operation is carried with “normal weather” (understating “normal weather” as the lack of any abnormal climatological conditions e.g. rain, snow, etc.).

As mentioned before outdoor limitations are advise as the device is not waterproof and water penetration is likely to happen if not careful, additionally, it should be taken into consideration that the case cannot withhold extreme mechanical strain-stress forces and use under extreme applications (e.g. hiking, skating, or other extreme sports, etc.) is not advised.

3.2 List of harmful and dangerous factors and risk analysis

3.2.1 Technical means to prevent harmful impacts

All parts of the device presented in this work have fixed parts and under normal circumstances should not have any moving parts. Section 9.3 indicates that any rough surfaces, sharp corners and edges of ME equipment that could cause injury or damage shall be avoided or cover, this description might to a certain extent fit to the piezo elements used for this project, for this reason the sensor was covered to avoid any damage to the patient’s skin as the patient will use the device during long periods of time and can cause bruising, cuts or skin rupture and leave body marks (Figure 3.2).
3.2.2 Electrical safety according to IEC 60601-1-2012 «General requirements for basic safety and essential performance»

According to classification for equipment meant to be used in medical applications, the device presented in this work is classified as a class III type B protection (Figure 3.3) device as the piezo element are insulated from the patient by the means of a shrinking non–conductive tube (Figure 2.2) and one lead directly connected to ground. In case of rupture of this protective layer, the device voltage never surpasses the 60 V limit established in subchapter 8.4 subsection 8.4.2c. Additionally, all capacitive devices comprising the device never have under any circumstances (of normal working, breaking, short circuit or open circuit without external interference or connection to external power sources) a residual voltage exciding the level depicted (60 V) in subsection 8.4.4 of the same standard.
The protective insulation of the sensor for the patient is suitable for use under any temperature condition below 100 degrees °C and is considered safe even under rupture as no direct contact with the sensor exists and both terminals of the sensor are separated on from the other (Figure 2.1) avoiding the effect of parasitic capacitive voltages.

3.2.3 Electromagnetic compatibility according to IEC 60601-1-2-2014 «Electromagnetic disturbances – Requirements and tests»

The system presented in this project is meant to be used with a ME Equipment as defined by the IEC 60601 standard. An artificial hand to be specific. Before testing the device, itself, the above mentioned artificial hand most be tested and connected as follows according to the «IEC 60601-1-2-2014: “electromagnetic disturbances – requirements and tests” »:

- For patient coupling points that do not have a conductive contact, the patient coupling point is terminated with the artificial hand and (series) RC element shown in Figure 3.4. The metal foil of the artificial hand is sized and placed to simulate the approximate area and location of patient coupling when the ME equipment or ME system is providing its intended use. The metal foil of the artificial hand is connected to terminal M of the RC element and the other terminal of the RC element is connected to the ground reference plane.
• For patient coupling points that have conductive contact to the patient (patient connection), terminal M of the RC element is connected directly to the patient coupling point, and the other terminal of the RC element is connected to the ground reference plane. If normal operation of the ME equipment or ME system cannot be verified with terminal M connected to the patient coupling point, an insulating material with a maximum thickness of 5 mm may be applied between the metal foil of the artificial hand and the patient coupling point. In this case, the metal foil of the artificial hand is to be sized and placed to simulate the approximate area and location of patient coupling when the ME equipment or ME system is providing its intended use, and terminal M of the RC element is to be connected to the metal foil but not to the patient coupling point. The other terminal of the RC element is connected to the ground reference plane in all cases.

• For ME equipment and ME systems that have multiple patient coupling points intended to be connected to a single patient, each patient coupling point and each patient-coupled part is to have an artificial hand applied as specified above. The artificial hands are connected to a single common connection and this common connection is connected to terminal M of the RC element (as specified in 8.3 of CISPR 16-1-2). For ME equipment and ME systems intended to be connected to multiple patients, artificial hands are to be applied as specified above and a separate common connection and RC element is to be used for each patient for which the capacitive coupling effect and RF impedance is to be simulated. The other terminal of each RC element is connected to the ground reference plane in all cases.
Once these tests are carried out, results must be presented in the provided format so that the patient and other personal involved with the ME equipment are aware of the limitations and cautions to have when using the ME equipment in unknown environments.

For the system provided for the control of the prosthetic device presented in this paper, according to IEC 60601-2-4 standard for electromagnetic shielding and protection of ME equipment, secondary devices like this one that interact directly in the functioning of the main ME equipment (a.k.a. the upper limb prosthetic device) it is necessary for it to comply with the same normatives and standards as the main ME equipment. For this reason, magnetic shielding must be provided for the control unit.

A valid method to do so, is to use the same approach as the one used to calculate the magnetic shielding of piezoelectric devices in ultrasound applications for medical use (with frequencies below 10MHz) as described by Wen et. al [30].

This method is utilized in medical ultrasound applications, as so some implications differ from the method used to control the prosthetic device explained in this report but, is a valid method that can be extrapolated to find a suitable solution for the sensor used with the control unit presented in figure 3.2.
A direct approach to reducing the crosstalk noise in figure 3.5 would be to put shielding layers in front the probe that are much thicker than the RF penetration depth, this noise increases as the initial force applied to the sensor during the attachment with the Velcro stripe is conducted and can be seen a high frequency AC signal. However, even in medical ultrasound applications it is difficult to find a material that does not disrupt the acoustic signal at those thicknesses. For this reason, an electromagnetic waveguide shown in figure 3.6 is commonly used in hall effect imaging (HEI). The waveguide is a thin copper cylinder coaxial to the probe, with one end connected to the metal jacket of the probe and the other end capped with a cellophane membrane. The diameter of the waveguide is 26 mm, and the distance between the probe surface and the cellophane membrane is 37 mm. The waveguide is filled with mineral oil as the acoustic coupling medium. The inner surface of the waveguide is lined with a foam layer for insulation against acoustic noise.

Crosstalk is not the only noise that affects piezo elements, Lorentz vibrations are also a source of electromagnetic interference (EMI) and must be dealt with to comply with the IEC and other international standards.
The schematic presented in figure 3.6 is valid for ultrasonic probes were the principle of the inverse piezoelectric effect is used, although a similar way of thinking can be used to design a similar shielding for the sensor presented in this report.

The Lorentz vibration noise is the part of the coherent noise that only occurs in the static magnetic field. The mechanism for this noise is as follows. The excitation pulse applied to the sensor during a muscle contraction also produces RF electric and magnetic fields in the vicinity of the sensor (through the sensor cables to the voltage comparator attachment). These RF fields induce eddy currents in the metallic components nearby to the sensor and if possible metallic parts should be avoided if possible in its construction. In medical ultrasound applications (US), to reduce the Lorentz vibration noise from the waveguide shield, a foam layer is used to acoustically isolate it from the ultrasonic probe (figure 3.6). The rest of the noise comes from the metallic parts inside the probe, especially the ones in direct contact with the PE element, including the nickel electrode plating and the tungsten backing.
In the application presented in this report, such noise can be presented as well in the metallic ring base of the piezo element.

In US Two methods are demonstrated to be effective in reducing the Lorentz vibration noise from inside the probe. It is based on the idea that if the metallic plating on the PE element are made normal to the magnetic field, then the Lorentz forces on the plating will be tangential to the PE element and therefore will not emit acoustic noise into the element or the sample. Using a prism and some other considerations the direction of the pulse is controlled. In the waveguide shielding method shown in figure 3.6, the waveguide cylinder is acoustically insulated from the probe and the saline; thus, the Lorentz vibration noise comes from the metallic components inside the probe. Neither of this methods has been tested for the application presented in this report, for that reason they cannot be recommended, but since the metallic object in the vicinity of the piezo element is part of the piezo element, little to none existing solutions have been found for this problem, in practice Lorentz vibrational noise for this application is small and the solution found was to use a comparator stage as mentioned before to isolate the measured signal from the control unit, but having it always control the timing and triggering of the control signal creation.

3.3 Conclusions for the special safety requirements

IEC standards provide a very clear normative as to how to protect the patient in almost every typical scenario and additional normatives and standards can be found from ISO to complete any missing information. The most important aspect to consider, is that the patient’s safety is a priority above all. To ensure the patient can use the device in a safe manner the following considerations must be considered: The device can be used indoors and outdoors but not in any extreme scenarios, as it has not been tested under extreme conditions both environmental or mechanical. For this reason, when in extreme cold or hot weathers the device should be covered to maintain its temperature under working range; extreme activities should be
performed with additional accessories, tools or attachments for the prosthetic device to ease the shear stress forces affecting the device.

Mechanically and electrically the device can be considered safe, as no leakage voltages or currents above the defined limits of the IEC 60601 normatives was found, and no mechanical stress is put over the patient’s limb due to the use of the device. To ensure this, coverage for the sensor is provided, only if the patient suffers from any allergies to nylon or polyolefin especial measures should be taken.

Electromagnetic interference (EMI) is a factor to take into consideration, as piezoelectric devices suffer from this type of noise from different sources and are prone to be affected by them. EMI sources that are inherent to piezo elements are, crosstalk noise and Lorentz vibrations and as mentioned before several methods exist to tackle this problem but none has been tested with the device design for this project as it was not tested in a real-life scenario and more exhaustive tests should be performed.

Overall the device is considered safe to use with real patients in a wide variety of scenarios, but always attention should be paid to any abnormal behavior in environments not discussed in the “especial safety considerations” chapter of this report.
CONCLUSIONS

Briefly (one to two pages) describe the main results of the work, analyze their compliance with the target set, bring recommendations on the practical use of the results and show the possible prospects for further development.

The work was completed as planned. All three components: the signal gathering unit, the control signal creation unit (processing) and the control unit; were design according to the objective established for the GQW. The signal gathering unit obtained is in principle a piezoelectric force sensor (PSF), although it works as a contraction detection sensor, with an average signal value between 1,5–3 Volts and a maximum peak value up to approximately 12,5 Volts when realizing a hard and quick muscle contraction. The processing of the signal is a signal thresholding circuit comprised by a signal comparator based in the patient’s response with the sensor to calibrate a value that equals a comfortable and easy-to-achieve level for the patient. And finally, the control unit is one of two: a digital potentiometer circuit or a multiplexor sensor; connected to a 555IC monostable circuit to produce a PWM signal and control the servomotors that serve as actuators for a prosthetic device.

The system was tested against an EMG board of the same technological level and a similar price range. Measurement of the RAW biological signal for both sensor and boards were performed; the PFS system had an average accuracy of 91,25 % when used at the brachioradialis muscle and 81,25 % when used at the triceps brachii muscle. The EMG board had a 72,5 % accuracy used at the biceps brachii muscle.

When using both PFS together the specificity of the sensor at the triceps brachii lowered by almost 10 % due to the little amount of time the subjects had to adapt to using the sensors. The EMG boards works the same regardless of the subjects’ time invested with the sensor, but, the device only worked correctly when flexing the arm to a certain degree level.

As it can be concluded by the results obtained, the PFS has a higher accuracy when compared with an EMG board of the same technological level, obtained as expected a board that provides a higher reliability for the patients, because it is easier
to use, it requires less time to master and has a flatter training curve (additional testing is recommended).

The control method proved to be efficient for the activation of servomotors and thus provides a reliable and affordable solution for the control of the device. Although servomotors themselves introduce a certain degree of uncertainty to the system as a whole, other solutions and devices are available to overcome these problems.

The device can be easily included to several of the current prosthetic solutions in the market and its practical application is in this field as a control method. The device is thought as only a quantitative method for the sensing of the contraction moment at the muscles’ belly. Qualitative and physiological information is not as complete and complex as the one received from EMG sensors and boards, for this reason the application of the device for estimation of muscle condition and assessment in medical applications like injury or effort under exercise assessment is not recommended and other methods should be used instead.

Although this is true, a possible improvement to the system is the addition of multiple sensors in a sleeve to create a pressure map of the amputated limb under working conditions is a viable and real possibility for the device further application.

Finally, the device fulfills completely the objective and tasks proposed and gives margin for future development.
REFERENCES AND ONLINE BIBLIOGRAPHY


5. Copyright © John Wiley & Sons, Inc. All rights reserved. Chapter 12 Nervous Tissue Lecture slides prepared by Curtis DeFriez, Weber State University. Published: 14.08.2014.


**APPENDIX A**

*Modes of vibration for common piezoelectric ceramic shapes*

<table>
<thead>
<tr>
<th>Axis</th>
<th>Polarization Direction</th>
<th>Applied Field</th>
<th>Mode of Vibration</th>
<th>Frequency Constant</th>
<th>Capacitance</th>
<th>Static Displacement</th>
<th>Static Voltage</th>
</tr>
</thead>
<tbody>
<tr>
<td>Disc</td>
<td>length or transverse (or c)</td>
<td>$N_1 = \beta$</td>
<td>$C = \frac{K_e a^2}{h}$</td>
<td>$\Delta = \frac{d}{2}$</td>
<td>$V = \frac{dF}{2}$</td>
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<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>$N_2 = \pi$</td>
<td></td>
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</tr>
<tr>
<td></td>
<td></td>
<td>$N_3 = \alpha$</td>
<td></td>
<td></td>
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<tr>
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<td>$\Delta = \frac{d}{2}$</td>
<td>$V = \frac{dF}{2}$</td>
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<td>$N_2 = \alpha$</td>
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<tr>
<td></td>
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<td>$N_3 = 2\alpha$</td>
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<td>$\Delta = \frac{d}{2}$</td>
<td>$V = \frac{dF}{2}$</td>
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<td></td>
<td>$N_3 = \alpha$</td>
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<td>$\Delta = \frac{d}{2}$</td>
<td>$V = \frac{dF}{2}$</td>
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<tr>
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APPENDIX B

Pinout for the components used in the project

B.1. 555IC pinout and internal configuration

![555IC pinout and internal configuration](https://commons.wikimedia.org/w/index.php?curid=2802571)

555IC Inner configuration and pinout. By BlanchardJ - Own work, Public Domain. URL: https://commons.wikimedia.org/w/index.php?curid=2802571

B.2. MG995 Servomotor

MG995 Tower Pro servo motor characteristics:
- Interface: Analog
- Peso: 55 grams
- Torque a 4.8 volts: 10.00 kg/cm
- Operation voltage: 4.0 a 7.2 volts
- Rotation speed at 4.8 volts: 0.2 sec / 60 °
APPENDIX C

Voltage Divider Diagram and SPI interface Write Sequences and timing diagram and potentiometer measurements for TI’s TPL051.

C.1. Equivalent Circuit for Rheostat Mode with Terminal H to Terminal W Resistance

![Rheostat Mode Circuit Diagram]

C.2. SPI Digital Interface

Table 1. Default Value 0x80H

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<tr>
<th>BIT7</th>
<th>BIT6</th>
<th>BIT5</th>
<th>BIT4</th>
<th>BIT3</th>
<th>BIT2</th>
<th>BIT1</th>
<th>BIT0</th>
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<td></td>
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<td></td>
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<td>LSB</td>
</tr>
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<td>D6</td>
<td>D5</td>
<td>D4</td>
<td>D3</td>
<td>D2</td>
<td>D1</td>
<td>D0</td>
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![SPI Write Sequence Diagram]

*Figure 15. SPI Write Sequence*
Table C.1 – shows the common values used for the digital potentiometer and the binary input to obtain such values.

<table>
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<tr>
<th>Binary</th>
<th>$R_{HW} (k_\Omega)$</th>
<th>$R_{LW} (k_\Omega)$</th>
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<tr>
<td>00000000</td>
<td>100</td>
<td>0</td>
</tr>
<tr>
<td>10000000</td>
<td>50</td>
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<tr>
<td>11111101</td>
<td>1.17</td>
<td>98.83</td>
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APPENDIX D

Table D.1 – Example page to register the *SGU response under muscle contraction*

<table>
<thead>
<tr>
<th>Number in list</th>
<th>Observed result</th>
<th>Number in list</th>
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