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Design and development of a dry, textile-based wearable technology for body motion sensing

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8 January 2018

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Dear Sir:

This report, entitled "Design and development of a dry, textile-based wearable technology for body motion sensing", was prepared as my 4A Work Report for the University of Waterloo. This report is in fulfillment of the course WKRPT 400, as required by my BASc Nanotechnology Engineering degree.

I was employed at Myant, a Toronto-based company that designs, develops and produces wearable technology. I was part of the R&D team, which was responsible for early protoyping, rapid testing and long-term contract projects with clients. The purpose of this report is to present my results on the development of a novel, textile-based wearable for detecting motion precisely, as part of a client project contracted with the R&D team.

I would like to thank CEO of Myant Tony Chahine and Executive VP Ilaria Varoli for providing me with this great opportunity wherein I have gained valuable experience in the world of wearable technology. The working environment established by Tony and Ilaria has fostered a very capable and creative team of people, which I was fortunate to be a part of, allowing me to bridge my previously acquired research skills with this newly found biomedical industry setting and business understanding.

I would also like to thank the rest of the personnel at Myant, as the work was very cross-functional in nature, and required support from various individuals. In particular, I would like to thank my supervisor Dr. Milad Alizadeh-Meghrazi who was directly responsible for supervising my work throughout the project, and helped me get past many obstacles along the way. Additionally, I would like to thank Monica Nealis, one of the lead product developers, who worked closely with me in developing all the various iterations of the wearable sleeve. Furthermore, I would like to thank Dr. Rishabh Gupta and Dr. Abdul Javaid for aiding in the development of my analysis code. I hereby confirm that I have received no further help other than what is mentioned above in writing this report. I also confirm this report has not been previously submitted for academic credit at this or any other academic institution.

Sincerely,

Dhilan Bekah
ID 20512356

Contributions

The ensemble of very talented individuals at Myant is extremely multidisciplinary, comprising of mechanical, electrical, computer, and textile engineers, fashion, U/I and U/X designers, as well as data scientists, chemists, knit technicians and sport scientists. I was employed in the R&D Department at Myant, which studied everything from fibre level development to physiological influencers, tackling several problems in various fields of technology, such as biomedicine and healthcare, energy harvesting, and materials development. The R&D team consisted of approximately ten people, including my supervisor, a recent chemical engineering graduate, an electrical engineer with over ten years of experience, a few PhDs, and an administrative director tied to the company since its inception. It is important to note, however, that a handful of other individuals throughout the company had a role to play in the work I carried out. During my term at Myant, I was involved in several ongoing projects.

As a member of the R&D team, my leading responsibility was to design, develop, and produce a wearable technology system to detect motion using fully-textile sensors for a client. My secondary tasks involved supporting the development of high priority projects that were ongoing, including client projects, as well as in-house Myant products (primarily SKIIN). Key tasks involved acquiring and testing materials (washing, strength, durability), contacting vendors, rapid prototyping, and product testing and validation. These tasks sometimes superseded my primary project if deadlines for later-stage products were more imminent.

This report focuses on my primary task of developing a textile-based motion sensing wearable technology. More specifically, this report serves to discuss my design choices, materials selection, computer programming analysis, as well as testing and results. Furthermore, this report serves to demonstrate the novelty and potential of the product created, in comparison to current standard technology. Throughout the duration of my project, I gained a tremendous amount of knowledge into the world of textiles, electronics and the wearable technology industry. This knowledge is directly related to both my theoretical and practical experience as a nanotechnology engineer, since the project required multi-disciplinary expertise, examining concepts dealing with materials science, biology, programming, and electrical and chemical engineering. Moreover, being exposed to such a fast-paced environment with several very talented and equally multi-disciplinary individuals was great for my growth as a nanotechnology engineer, due to the rapidly-evolving nanotechnology industry.

In the broader scheme of things, my project is part of Myant's vision for seamless integration of technology for ameliorating day-to-day life for all demographics. The development of a motion-sensing technology that is textile-based and user-preferred is one of the many form factors that Myant's technology targets. Others include pressure sensing, temperature response, conductivity, and more. On a lower level, this work helped meet client demands for one of the major ongoing projects, pushing the commercialization of another Myant product closer to completion. On an even lower level, the results from the material property and signal quality tests contribute greatly to the overall material database that Myant is creating. The investigated textile electrode systems, for example, are easily transferable to other projects at Myant. In the long-term, this work helps obtain further clients and grants, and continue Myant's growth.

Summary

Wearable technology has shown rapid growth in combination with the increasing reliance on portable medical devices and POC treatment and monitoring in the modern healthcare system. Major sectors such as sport & fitness, healthcare, and wellness have greatly benefited from these new technologies. Electrodiagnostics in particular have found great prevalence in today's research and development. The scope of this report encompasses one of the major types of electrodiagnostics for body motion sensing through EMG. More specifically, this report serves to demonstrate the process behind the design and development of a dry, textile-based wearable technology for body motion sensing. Primary customer requirements such as real-time function and response, as well as a variety of wearable criteria are outlined and addressed.

The major points covered in this report concern the establishment of a control system for real-time EMG acquisition and processing, as well as the design, fabrication and testing of the wearable garment for the same purpose. A description of the control system is given by discussing the type of electrode, electronic module, and hardware / software system that is used. Ultimately, gel electrodes, an Olimex EMG Module, and Arduino and MATLAB were chosen to tackle the project problem statement. Following this, an initial design of the wearable is discussed, focusing on target muscles, electrode placement, and electrode type. Testing and refinement are discussed afterwards, addressing improvements made to the system. The final design consists of a double layer sleeve, embedded with terry' knit-structured electrodes made of silver yarn, and a design for an in-house on-body EMG module based on the Olimex shield.

The major conclusions of this report concern the success of the developed wearable technology with regards to the customer requirements. A fully-functional, dry, textile-based garment capable of real-time function and response was successfully created through EMG code, conductive textile electrodes, and an Olimex EMG module. Moreover, comfort, washability, and unobtrusiveness were well addressed by appropriate materials selection, and user-friendly product design. Therefore, customer requirements were well met, and a comparison of the newly developed technology with the existing technology is evaluated, demonstrating its advantages.

The major recommendations of this report are to develop the in-house electronics module for inconspicuous transmission and wearable capabilities, adapt the software and hardware systems for multi-channel acquisition, and further develop new materials, electrodes, and muscle-targeting garments for expanding the technology's capabilities.

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1 Introduction

Wearable technology refers to electronic devices that are designed to be worn / embedded into articles of clothing for the purpose of enhancing an individual's day-to-day life by contributing to the "Internet of Things" (IoT) [1]. Through these wearable devices, data can be exchanged between multiple platforms, and even between multiple individuals, both on and offline. They can perform many of the same tasks as mobile phones or computers, often demonstrating increased performance in their tasks [2]. Moreover, their functions generally allow them to be seamless and easily integrated into everyday life due to their communications capabilities [2]. In conjunction with advancements in technology and growth in global innovation, wearable technology has gained huge popularity over the past few years, showing prevalence in a number of sectors from healthcare to military & defense [1]. According to Techcrunch, the estimated market value for the wearable technology industry in 2017 was USD $30.5 billion [3]. It is expected to increase rapidly to about USD $95.3 billion by 2021, and an astonishing USD $150 billion by 2026, as depicted in Figure 1 [3].

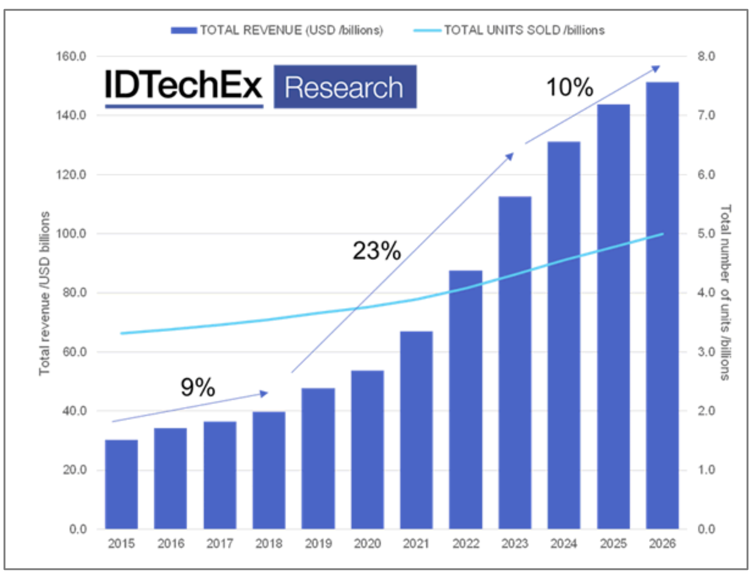


Figure 1. Global market trend for the wearable technology industry over the next 10 years [3]

With increasing expenditure and investment, there will be an increase in the range of devices available to consumers, expanding to new markets and new areas of the human body [3]. Fitness tracking, security profiling, and network monitoring are major contributing factors with regards to higher adoption of wearable technology [1]. Furthermore, today's increase in health awareness, portable and convenient usage of technology, and entry of large smartphone manufacturers will continue to drive the industry's growth [1].

1.1 Applications

There are several sectors in which wearable technology has penetrated, with a number of applications in each. In communications, wearables have found their use in inter-platform and inter-individual connectivity, integrating Wi-Fi and Bluetooth into clothing and jewelry. In lifestyle computing, wearables such as smartwatches, VR gaming headsets, and HUD glasses are used for schedule management, augmented reality, and displays. In business operations, information access and control has lead to the development of hotel key and event ticket bands. In security and safety, wearables have enabled situational awareness for military personnel, and trackable personal items such as wallets and bracelets for identity recognition and profiling. Wearables have even been used in fashion for decorative displays and reactive responses, such as light reactive and shape-shifting dresses. One of the major sectors that wearables has impacted greatly is the medical industry, offering unobtrusive patient monitoring capabilities, as well as biomedical implants and disease management (e.g. Diabetes). Expanding to the broader healthcare industry encompasses the wellness sector, which finds wearable technologies capable of physiological monitoring (sleep, diet, motion, weight, energy). Finally, wearables have had a tremendous impact on sports & fitness, offering performance monitoring and training, goal management, virtual coaching, and body temperature regulation [4]. It is clear that wearables have become increasingly cross-functional and diversified, since this is only a fraction of the possible applications and uses.

Body motion sensing is a very important use-case of wearable technology, falling into many of the aforementioned categories, and is the focus of this report. It is essential to the development of prostheses, sports medicine & fitness, radiotherapy, and biomechanical research [5]. For instance, head rotation and body orientation are the input signals for human balance prosthesis, chest wall movements must be monitored precisely during respiration support, and body motion characteristics must also be continuously evaluated during rehabilitation processes for physically disabled individuals [5]. Motion detection can be accomplished via mechanical or electronic means, but in both cases, sensors must be used. These sensors can function via a variety of mechanisms, including infrared (passive and active), optics (video and camera systems), microwaves and ultrasonic waves, sound, physical vibration, and magnetics [6]. Some of these mechanisms present greater difficulty for body motion sensing as compared to general motion detection due to their cost, complexity, and lack of body precision [5]. Given the modern healthcare system, which relies increasingly on portable medical devices and POC treatment and monitoring, wearable devices have become a prime area for exploration [5]. In fact, in recent years, many companies have created wearable technologies for body motion detection using a principle known as electromyography (EMG). EMG is part of the electrodiagnostic medicine family that comprises of techniques such as ECG, EEG, and EOG for investigating various biosignals. EMG in particular monitors muscle activity, corresponding to a signal that is measured from the electric potential generated by muscle cells during motion. Figure 2 illustrates this general concept.

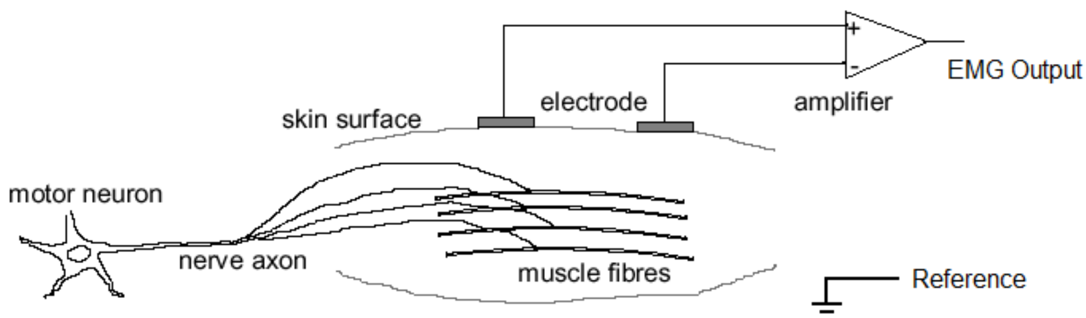


Figure 2. Schematic for how EMG functions generally [7]

The action potential generated from muscle movement is captured between the two measuring electrodes, and is baselined against a reference electrode (effectively acting as an electrical ground) placed on electrically unrelated tissue. There are two types of EMG techniques: intramuscular and surface [8]. The former (iEMG) involves inserting a needle electrode through the skin and into the muscle of interest, whereas the latter (sEMG) involves placing the electrodes on the surface of the skin [8]. iEMG is typically used for research or kinesiology studies due to its increased precision and added complexity as compared to sEMG. For the purposes of a wearable technology, however, sEMG is the clear choice due to its non-invasive nature and commonality in physiotherapy.

1.2 Textiles and Printed Electronics

With the advancements made in materials science over the last few years, textiles and printed electronics have greatly increased their range of functional properties. At Myant, a wide variety of knitted and printed solutions have been created for taking the functionality of conventional technologies, and embedding them into textiles and wearables. Using seamless circular, warp and flat knitting, the most natural solutions next to skin can been achieved, providing both passive and active functions in precise targeted locations within a garment or product. Examples include features such as moisture management, thermochromics, photochromics, electrochromics, biometric sensing and more [9]. As for printed solutions, Myant's technologies combine science, engineering, and design to develop and integrate flexible electronic devices, sensors, and actuators into textiles. Examples include temperature, pressure, stretch, and electrochemical sensors, as well as electrodes composed of flexible conductive polymers, and flexible superpacitors for energy storage [9]. Of particular interest for this body motion sensing project, however, are the in-house developed knitted textile electrodes. These electrodes are composed of conductive yarns (typically stainless-steel or silver), and offer several advantages over conventional electrodes [10]. First, conventional gel electrodes typically used for EMG are disposable and must be replaced after a few hours of usage to avoid skin irritation. In addition, they cannot be treated like regular clothing (i.e. washed), which is fairly important for a wearable garment. As a result, they cannot be integrated into a wearable technology. The development of dry, textile electrodes is no new venture, however, and while it has greatly improved in recent years, there are several accompanying challenges such as noise, interference, and motion artifact.

1.3 Project Objectives

This report focuses on the development of a body motion sensing wearable technology by integrating textile electrodes, using principles of EMG. To narrow the scope of the project and establish a strong systematic foundation, the focus was geared towards developing an arm sleeve garment, capable of detecting motion in one of the primary forearm muscles, through a textile-based approach (i.e. conductive yarn-spun electrodes). This project was commissioned as a result of a contract with a client. Therefore, this report will follow the design and development process of creating such a wearable, using the client-given guidelines. The customer requirements for this project can be thought of as follows:

1. Real-time function and response of body motion detection
2. Comfortable, washable, seamless and non-invasive

In order to address these requirements, an initial description of the software and hardware setup will be given. Then, the design considerations for the garment will be elaborated on, taking into account the functional specifications required. Testing and results will be provided in both the control and design sections of the report, in order to gauge the success of the design.

2 Establishing a Control System

In developing a novel EMG detection system that is textile-based, a standard must first be established in order to access the efficacy and overall success of the new design. To accomplish this, gel electrodes, a commercial EMG module for reading EMG signals, and MATLAB and Arduino software packages were used to gather and process the data.

2.1 Gel Electrodes

The performance of non-invasive surface electrodes in detecting biological signals (in this case EMG) is highly dependent on electrode-skin impedance [11]. Electrical impedance is the measure of a material's opposition to the flow of alternating electric currents of various frequencies. Electrode-skin impedance arises as a result of the highly resistant outer skin layer called the stratum corneum, which is in contact with the electrodes (see Figure 3) [11].

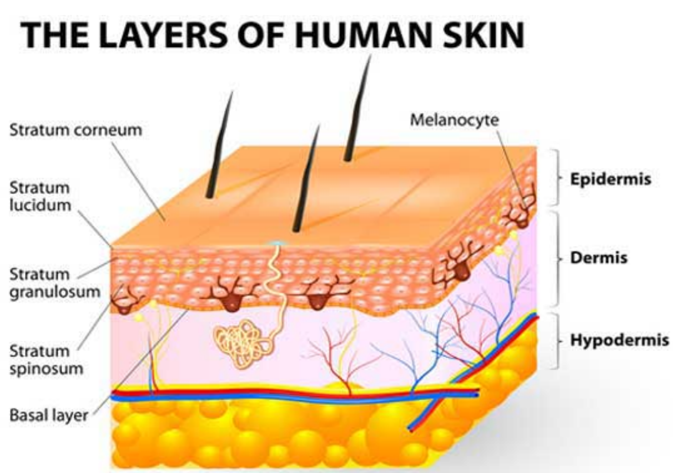


Figure 3. Layers of human skin [12]

This layer is largely composed of dead cells, and consequently impedes ion mobility, leading to low electron/ion exchange at the electrode sites [11]. As a result, there is weaker conductivity between the electrodes and the skin, reducing the biological signal amplitude, thereby contributing to a lower SNR. Therefore, high electrode-skin impedance is undesired. Conventional surface electrodes can either be dry or gelled. In the case of dry electrodes, recording surface electrical activity can be quite challenging due to this high electrode-skin impedance barrier. Gel electrodes use a conductive gel that is typically viscous and electrolytic in nature, lowering the skin-electrode impedance, thus allowing for easier conduction of ions and better signal quality. Surface Ag/AgCl gel electrodes are the most common, and are considered to be the universal electrodes for clinical measurements [11]. They generate low motion artifact due to their adhesiveness, low noise during biological signals recording, and significantly lower the skin-electrode impedance [11]. Moreover, they are easy to place for targeting a specific muscle, and lead to high reproducibility. As a result, these Ag/AgCl electrodes were chosen for establishing the control system.

2.2 EMG Module

In order to successfully measure EMG and gather meaningful data, an electronic module that is capable of processing EMG signals is required. Many systems have been created for this purpose, but since MATLAB and Arduino were pre-chosen as appropriate software and hardware platforms for this project, an Arduino-compatible EMG sensor was used. More specifically, the Myoware muscle sensor, depicted in Figure 4, was chosen. It is a 3-lead sensor capable of measuring the electrical activity of a muscle, outputting 0-V_s Volts depending on the intensity of the muscle action. In addition to being readily usable with the Arduino package, this product is especially suitable for the initial design of the wearable garment, due to its sleek wearable design that allows one to attach biomedical sensor pads directly to the board itself, bypassing the need for cables. In the case of establishing a control system, the sensor pads are the gel electrodes previously discussed. Moreover, the Myoware sensor possesses a battery pack shield for powering the board (in addition to powering it through an external source), as well as an LED visualization shield and an AUX-port shield with accompanying cables.

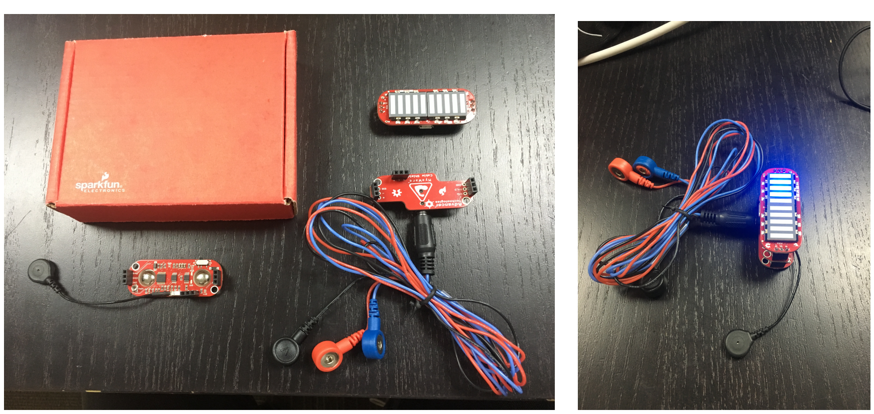


Figure 4. a) Full Myoware set ; b) LED shield signal intensity response

2.3 Arduino + MATLAB

Two software packages were used for EMG data handling and processing: Arduino and MATLAB. Arduino is an open-source electronics platform, based on hardware and software meant for rapid prototyping. The company designs and manufactures microcontrollers for building digital devices that can interact with the physical world. Through their software IDE, the Arduino platform offers great interfacing between hardware and software for data handling. MATLAB is a numerical computing environment which offers several capabilities that are pertinent to this project. First of all, it possesses the ability to integrate external libraries, which is very important as it allows for interfacing with Arduino to be possible and relatively seamless. MATLAB also contains a very large database of built-in algorithms for processing applications (i.e. signal, image, etc). It also allows for immediate testing of algorithms without recompilation, which is crucial for repeated data handling. There is also a large user community with lots of open-source code and knowledge sharing, as well as very well written documentation and examples.

2.3.1 Thresholding

Thus far, the detection and processing system comprises of a Myoware muscle sensor, connected to gel electrodes at the skin interface, and to cables feeding data to Arduino and MATLAB software on a computer. In order to first develop a real-time response system, establishing a thresholding system is necessary to determine triggering (i.e. when an action is turned on or off). In the case of this project, the triggering is determined from the on and offset of the muscle's contraction. This thresholding system was achieved by implementing a function in MATLAB that generates a signal envelope in real time based on the acquisition signal, and establishes a minimum and maximum threshold based on that envelope. The acquisition signal is received from the Arduino software/ hardware interface, using its MATLAB-compatible packages. The details, however, are not relevant to this report, and will be overlooked. A type of real-time digital filtering called Direct Form II IIR was used to produce the signal envelope. Again, while the premise behind this type of filtering is beyond the scope of this report, it is important to note that it essentially filters point-by-point, as data is being received. This is important to closely simulate real-time function and response. In terms of establishing thresholds, the minimum is calculated from the mean of the rest period, while the maximum is calculated from the mean of the maximum voluntary contraction period, once the acquisition period is terminated. The rest period is allocated more time (75%) in order to compensate for noise disturbances that could arise during the determination of the baseline (minimum). Furthermore, it is difficult to maintain the same maximum voluntary contraction for significant periods of time, due to muscle fatigue. This thresholding system is crucial to account for the inter-individual and inter-system variability, as it normalizes the action response.

2.3.2 Triggering

Upon having established a reproducible thresholding system, an EMG triggering and analysis function was created in MATLAB. In this function, the same Arduino hardware setup parameters are used as in the thresholding function in order to receive data. The previously obtained minimum and maximum values are passed into the function, wherein an "action" value is calculated based on their difference, divided by an arbitrary number of 'on' regions (in this case, five). Therefore, using the minimum, maximum, and action values, six distinct regions are established, including the 'off' region. Using the same type of filtering as in the thresholding function, the incoming data is point-by-point filtered and compared to the established threshold values for each distinct region. This system not only improves on the binary "on and off" system by improving resolution, but it also gives a qualitative measure of the signal intensity, and thus the muscle contraction.

2.3.3 Processing

After handling the real-time triggering, the data is stored in a text file, and the function proceeds to post-process the data, generating four key figures. The first is simply a plot of the raw EMG signal. The second plot contains the EMG signal after having been passed through two of the three IIR filters, which are fairly standard for EMG signal processing. The first is a high-pass filter allowing signals above 40 Hz, while the second is a low-pass filter allowing signals below 500 Hz [13]. The third plot contains the fully rectified and filtered signal, as well as the envelope that is generated from the third IIR low-pass filter, with a cut-off of 5 Hz. This method is one of the many ways of generating a signal envelope, and was chosen due to its simplicity and relative accuracy. Finally, the fourth plot illustrates the EMG signal magnitude as a function of frequency. This plot is the most important since it is used as the confirmatory figure of merit for verifying the presence and properties of EMG. An example of the results of this system, combining the gel electrodes, the Myoware sensor, and the MATLAB code can be seen in the 4-plot Figure 5. The frequency spectrum plot can be well compared to that of Figure 6, which is a canonical example of an EMG signal, found in literature. Note that the magnitude scale in Figure 5 is different from that of Figure 6, as different units were used, and the magnitude does vary between systems, individuals, and even tests. The figure's important feature is the shape in correspondence to the frequencies.

Finally, a script was made to call the thresholding and triggering/analysis functions, allowing for changes to be made to the acquisition time and sampling frequency for iterative purposes. A sampling frequency of 1070 Hz was ultimately chosen due to its high matching with the sampling frequency of the Arduino-MATLAB interface, and the minimum 1 kHz requirement for EMG detection. This final script, along with the accompanying functions, can be found in Appendix A of this report.

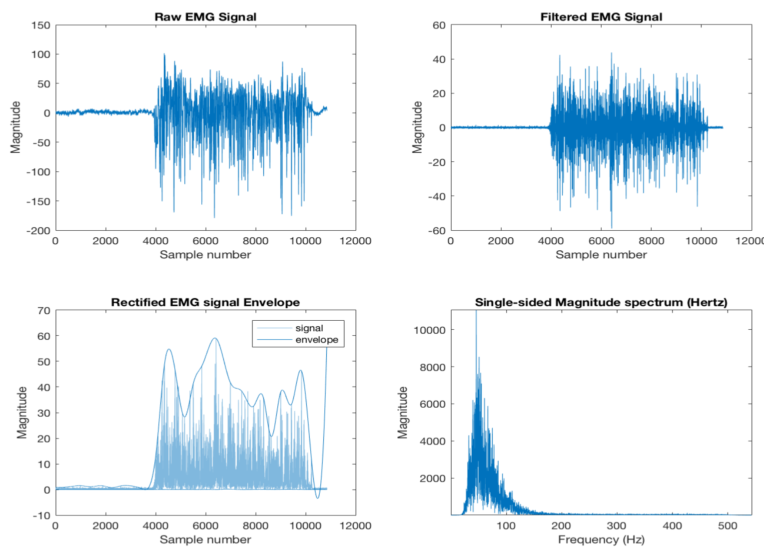


Figure 5. 4-plot generated from MATLAB EMG processing script for control system

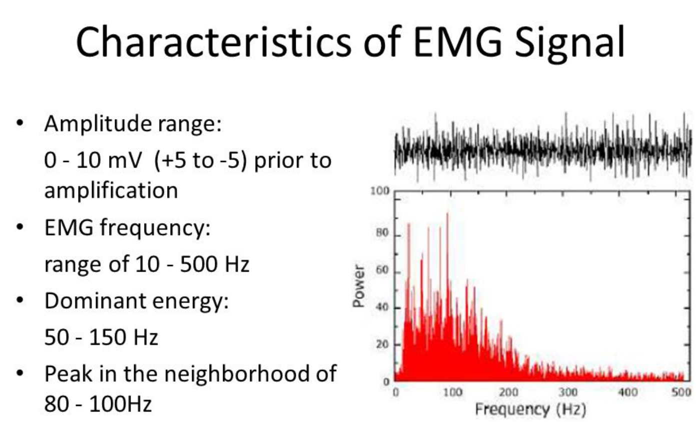


Figure 6. Typical frequency spectrum for an EMG signal [13]

3 Developing Wearable Technology

After establishing a working and reliable control system for testing EMG acquisition and processing, the next phase of development involved designing the wearable garment to incorporate the functionality of the system. To do this, a few functional considerations were made involving the target muscles, and the type and placement of electrodes primarily. First, the muscle to be studied – in this case one of the major flexor muscles of the forearm – was investigated in EMG literature. Second, various properties pertaining to the textile electrodes were examined and optimized for a wearable garment.

3.1 Target Muscles

In order to maximize the efficacy of sEMG, a proper understanding of the muscles from which the EMG signals are being extracted is required. The EMG electrodes should be placed at a proper location, with correct orientation with regards to the muscle fibers. More specifically, the detecting electrodes should be placed between the motor unit and the tendinous insertion of the muscle, along the longitudinal midline of the muscle (see Figure 7) [11]. Doing so allows for an improved superimposed signal, as a result of the detecting surfaces intersecting the high density of muscle fibers found in the belly of muscle [11]. Other variations of electrode placement, such as near-tendon or near to the muscle edge are sub-optimal. Muscle fibers become smaller and thinner as they approach the tendon, resulting in weaker EMG signals, while the chances of crosstalk from other muscles are considerably increased near the muscle edge. As a result, the distance between the centers of the electrodes should be kept to within 1-2 cm [11].

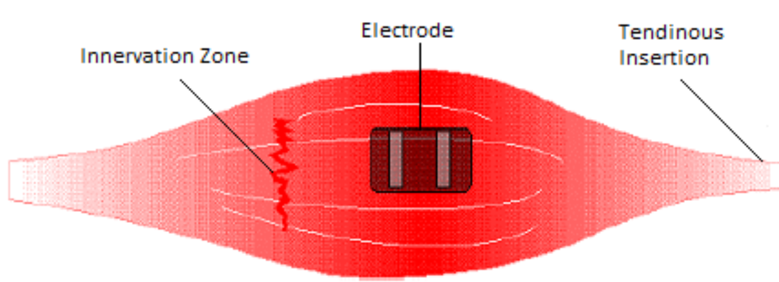


Figure 7. Electrode placement for maximal EMG detection [11]

Revisiting figure 2 of the report on the underlying working principle of EMG, a third electrode, termed the reference electrode, is required for providing a common reference to the differential input of the preamplier in the electrode system [11]. This reference electrode must be placed away from the target muscle, on an electrically neutral tissue (such as bone) in order to be most effective.

The muscle of interest for the initial design of the wearable garment is called the flexor digitorum superficialis, located on the anterior side of the forearm, and is maximally triggered via a "wrist down" motion, as depicted in figure 8. This muscle was chosen for its ease of testing, and well-documented literature.

Figure 8. a) Flexor digitorum superficialis (FDS) [14] ; b) Triggering the FDS [15]

3.2 Yarn / Electrode Types

After ensuring placement details, the next major factor that was considered was material properties. Since the textile electrodes of the garment aim to replace gel electrodes, they need to mimic the functional capabilities of gel electrodes with regards to EMG as closely as possible, while also accounting for the limitations / requirements of a wearable. Encapsulating this within the customer requirements, the textile electrodes need to have high enough conductivity, have good skin contact, and be both washable and comfortable.

As a result, an investigation into the different electrodes that were being fabricated at Myant was conducted. There are primarily two types of methods for fabricating textile electrodes for biomedical purposes at Myant, including knitting and polymer coatings. Knitting involves spinning different yarns with different material properties into textiles of various shapes, using highly advanced machinery such as STOLL knitting machines. The second method involves creating a textile base using knitting, and then coating the area(s) of interest in a polymer, possessing properties relating to the goals of the technology. At the time of this project, a parallel study was being conducted to effectively test and characterize all the in-house electrodes, in order to classify their use, and select the best ones for each application. Due to its infancy and concurrency with this project, however, the study had not yet been fully carried out. Fortunately, there were significant tests and iterations that had been done previously as part of other projects which more than sufficed for the initial design. Since Myant had previously worked on knitting solutions the most, in particular with conductive yarns, a knitting approach was chosen for the forearm sleeve wearable. Typical choices for conductive yarns were either silver or stainless steel. Silver showed the best overall results, including washability, conductivity, and its ability to be used in the knitting machines. While a variety of silver yarns with different conductivities and tensile strengths were available, a silver yarn from Noble Biomaterials was chosen since it was the most tested for human usage, and demonstrated good reproducibility as well as structural integrity.

3.3 Initial Design, Fabrication, and Testing

3.3.1 Version 1

Once the choice of materials, muscles, and general setup was established, a visual schematic for the design of the wearable technology sleeve was created, as demonstrated by Figure 8. This design was based on personal measurements for the initial testing, since they were meant for the client.

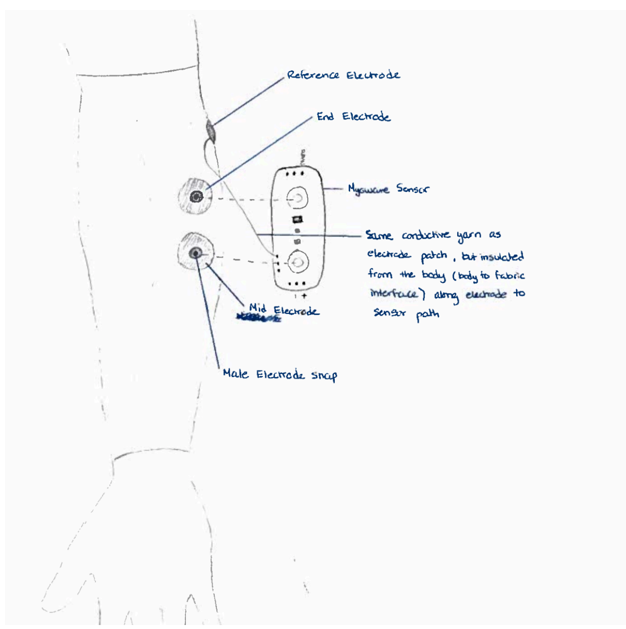


Figure 9. Initial design of wearable sleeve with integration of Myoware module

The sleeve was knitted using the silver Noble yarn, along with this design, including client dimensions and feature dimensions (i.e. electrode spacing and sizing). Moreover, the connection method for the electronics was addressed by integrating medical grade male electrode snaps onto the textile electrodes. These snaps are important as there is standard sizing in the field of medicine for electrode snaps, meaning regular clothing snaps cannot be used. They are typically a direct part of gel electrodes, but a vendor was successfully contracted to sell them as standalone parts for very cheap (~$0.03 per snap). On their own, they can be used just like snaps for regular clothing. The fabricated sleeve based on this initial design is shown in figure 10.

Figure 10. Fabricated sleeve (version 1), demonstrating knitted electrodes and integrated snaps

3.3.2 Challenges and Improvements

Initial testing of this sleeve, however, presented several difficulties. The first was bad skin contact due to lifting of the Myoware module, and consequently the lifting of the fabric, during attachment. As previously mentioned, poor skin contact directly contributes to a weaker (if not non-existent) EMG signal during recordings. This is amplified by the dryness of the textile electrode as opposed to the conventional gel electrodes, as there is more noise and a higher skin-electrode impedance present in the system. Since the system involving the Myoware module was unable to detect EMG from textile at all, and the same textile electrodes proved effective in other biosignal applications, it was hypothesized that the Myoware module was not capable enough to detect the low signals from textile electrodes. Moreover, the Myoware sensor was intended to be a DIY, low-cost version of an EMG system, furthering the notion that its design may not be complex enough to function properly for this application. To verify this hypothesis, alternative EMG modules were explored. Keeping in mind Arduino-compatibility, the Olimex SHIELD-EKG-EMG prototype board, depicted in Figure 11, was chosen to replace the Myoware sensor.

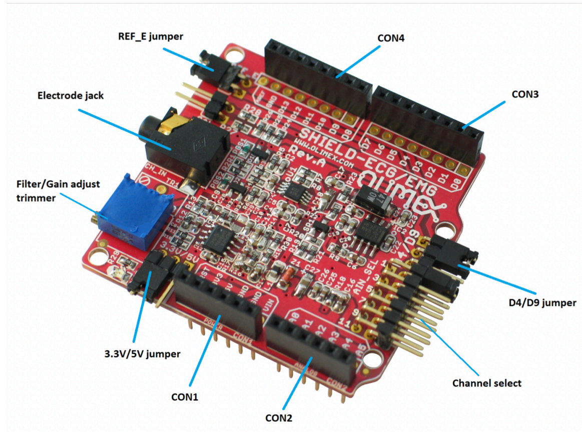


Figure 11. Olimex SHIELD-EKG-EMG module [16]

While not nearly as sleek or small, rendering it a poorer choice for wearability, this module offers increased performance due to greater specs, which was quickly confirmed through initial testing of the system using the textile electrodes. Additionally, the circuitry of the board allows for stacking of multiple channels for multi-channel acquisition, which is a key feature for the future of the technology. With multi-channel acquisition, EMG signals can be gathered from multiple muscles at once. Most importantly, however, is the fact that the documentation for the board includes an Eagle CAD schematic of the board's circuitry, which is included in Appendix B. By possessing the schematic to reproduce the board, this bulky module can be circumvented by creating an in-house Myant module. These modules, as depicted in Figure 12, are much smaller and more compact, and have been similarly designed for other projects. They provide increased function such as wireless capabilities, and can be more easily integrated onto the body for a seamless and non-invasive wearable technology, due to their size and shape. Therefore, for the purposes of testing the system, the Olimex EMG module was used as pictured above, with the understanding to develop the technology as discussed after being fully established.

Figure 12. In-house Myant modules comprising of the necessary technology components

3.3.3 Testing + Results

For testing, four different electrode patterns using Noble silver yarn were investigated, using the Olimex shield system. These patterns are termed ‘flat’, ‘fluffy’, ‘puffy’, and ‘terry’, and are illustrated in Figure 13 below. These names indicate the kind of knitting structure that is used to create the electrodes. The ‘flat’ structure is the most basic, as it is just a single layer of silver yarn. The ‘fluffy’ and ‘puffy’ structures provide the texture that the name implies, while the ‘terry’ structure is a pattern that one of the machines can make uniquely. EMG results for each type are given in Figures 14 and 15. These results are broken down into wet and dry results, corresponding to the application of conductive gel or not. Triplicate tests were conducted for each subdivision, but only the most characteristic results are presented below.

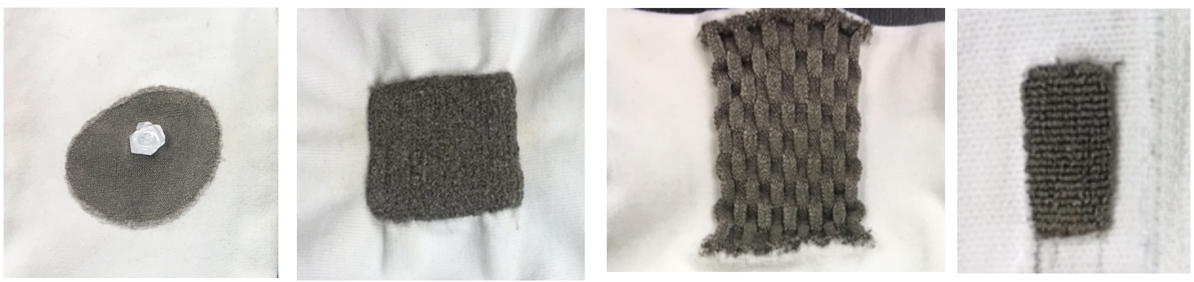


Figure 13. Textile electrodes: a) Flat b) Fluffy c) Puffy d) Terry

Dry Electrodes

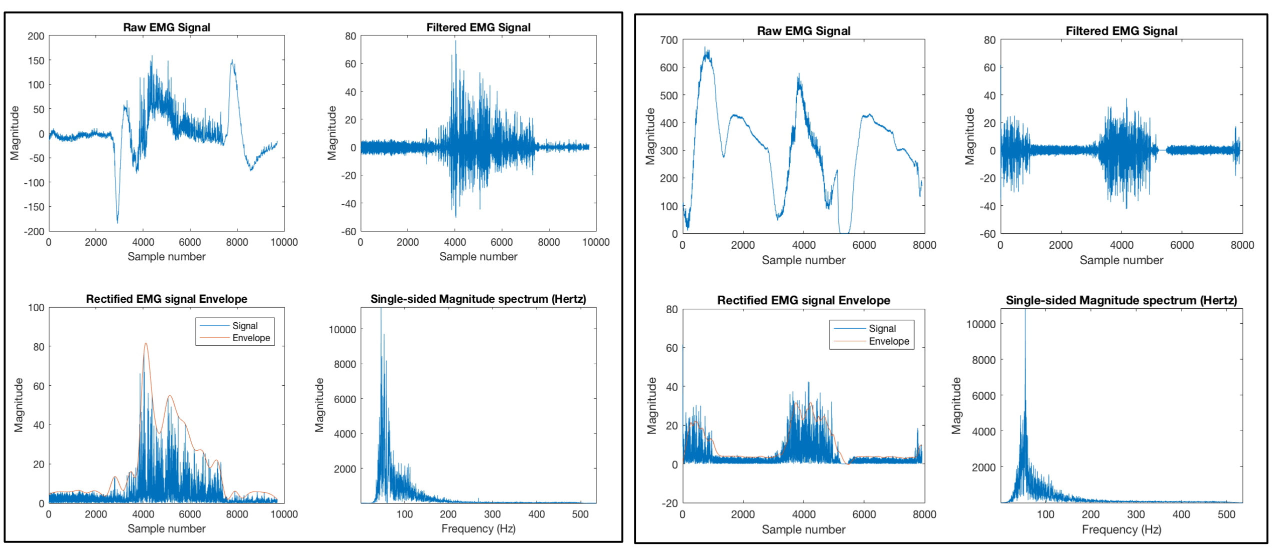


Figure 14. EMG results for dry textile electrodes. a) Fluffy. b) Terry

Wet Electrodes

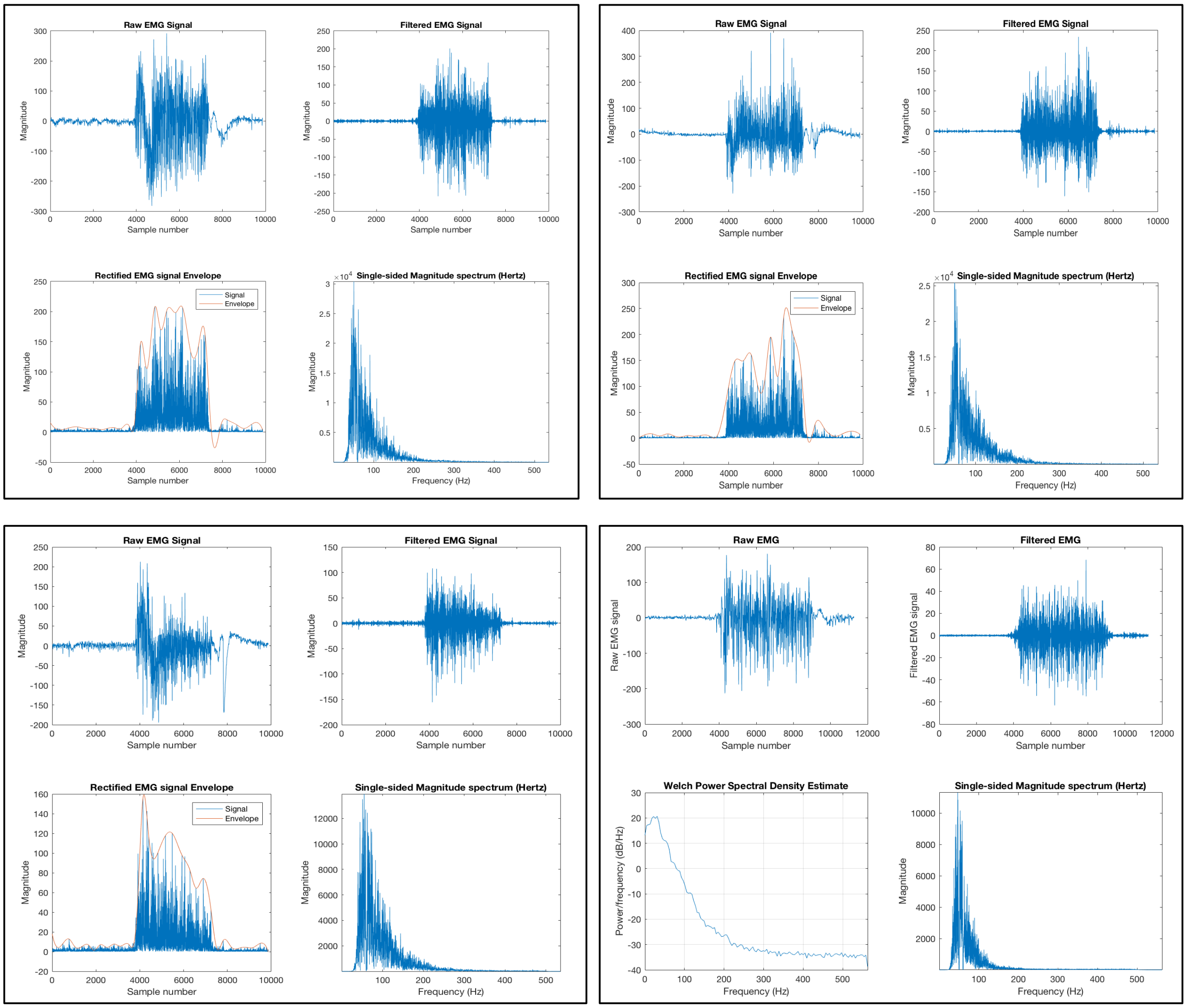


Figure 15. EMG Results for wet textile electrodes. a) Flat. b) Fluffy. c) Puffy. d) Terry

It is important to note that the third plot of the terry results from Figure 15 is different from the others due to its capture prior to the final changes that were made to the code described in section 2. Nonetheless, the second and fourth plots are the most important as they provide information with regards to SNR and the typical EMG frequency spectrum. Based on these results, all four knitting structures appear to show similar function when wetted by a conductive gel. A key distinction between the electrode setups must be made however, in order to correctly interpret the results. For the first three knit structures (i.e. all but terry), a gel electrode was used for the reference electrode, instead of a third of the same kind, since a sleeve could not be knit with circular shaped electrode patches for those three. This is because the terry pattern was formed on a different machine than the one for the other three patterns. Therefore, only the terry electrode was tested and capable of working as a standalone system. Furthermore, only the fluffy and terry knit structures were able to acquire EMG when dry. As a result, the terry knit structure was chosen for the further development of the sleeve due to these scalability and form-to-function issues. Finally, all of this testing was conducted with electrode patches of approximately 1.5 inch in diameter, which is larger than the standard gel electrode size. As a result, the patches were shrunk to ¾ inch diameters to more closely mimic the gel electrode size, and to reduce muscle crosstalk in order to improve SNR.

3.4 Final Design, Fabrication, and Testing

In addition to incorporating the aforementioned changes to the system, the sleeve was also redesigned to incorporate a double layer of fabric, as illustrated in Figures 16 and 17. This double layer serves to not only increase skin contact for better signal quality (higher SNR), but also to isolate the backing of the metal snaps from the skin.

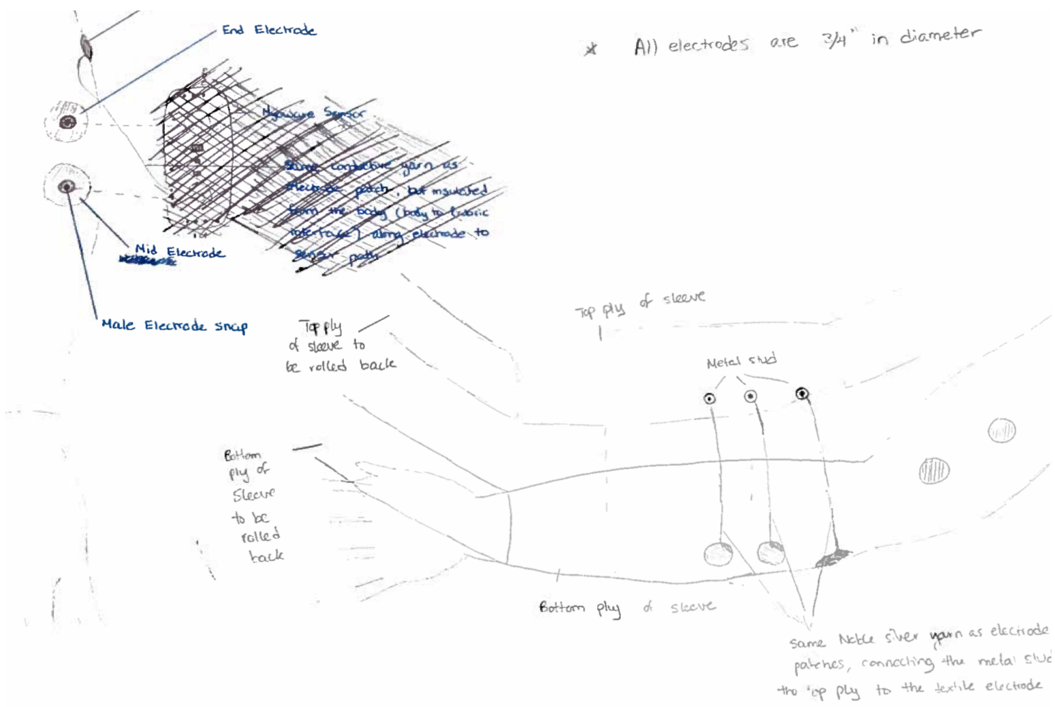


Figure 16. Sleeve design (version 2), including double layer

Figure 17. Fabricated sleeve (version 2)

This was achieved by fabricating two identical sleeve layers with the same three Noble 'terry' electrode patches, sewn through to each other using a conductive yarn thread to enable permanent contact of both the top and bottom patches. In doing so, the snap connectors for the EMG recording and electronics modules are attached through the top patch only, and do not make contact with the skin directly. This prevents the results from being skewed due to the conductance of the metal snap itself, and increases comfort for the user. More importantly, however, the double layer provides a snugger fit, and allows for cleaner end seems, again increasing user wearability and comfort. The results for testing this double layer sleeve, both wet and dry, are found below in Figure 18.

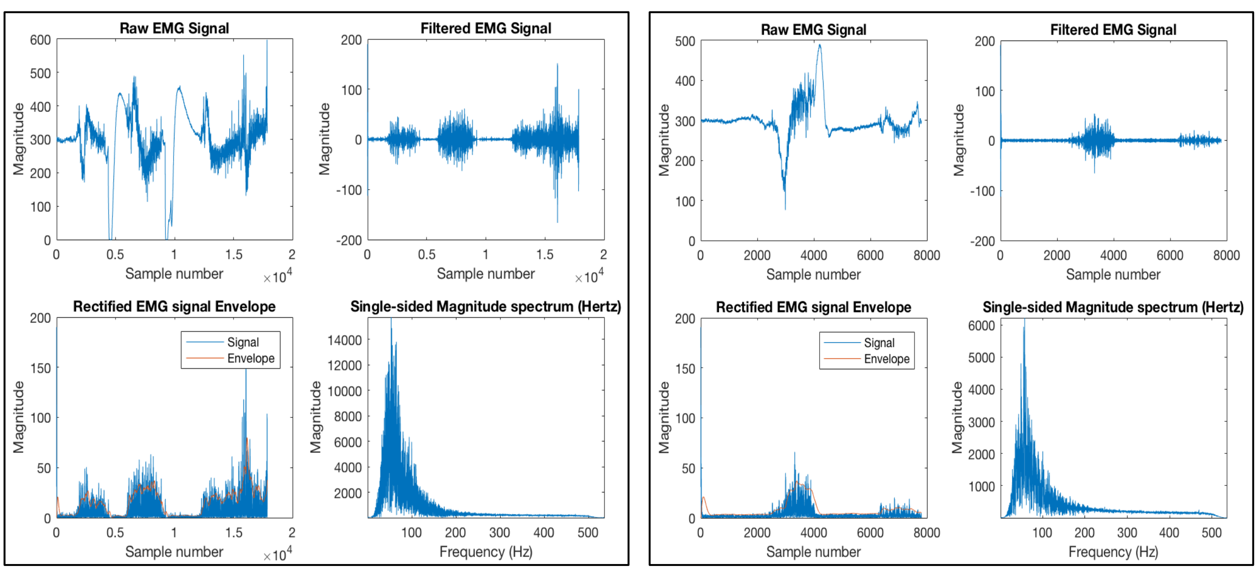


Figure 18. EMG Results for a) Wet Electrode ; b) Dry Electrode

It is clear from these results that the dry electrode system is fully functional. While the real-time function is confirmed from these results (as the real-time script was used), it is even more important to note that the triggering that was previously discussed in section 2.3 of this report was successfully visualized simultaneously during the acquisition period, which was the primary goal of the technology. More specifically, 5 LEDs were mapped to turn on (light up) for each region that was established, by connecting the system to a breadboard as illustrated in Figure 19. The more LEDs that were turned on, the more intense the EMG signal. Following the same logic, a lack of muscle contraction or movement turned off all LEDs. Moreover, this LED system demonstrated less than 300 milliseconds of delay, as verified by the tic and toc functions of MATLAB, which is acceptable for real-time emulation given the inherent communication and hardware limitations.

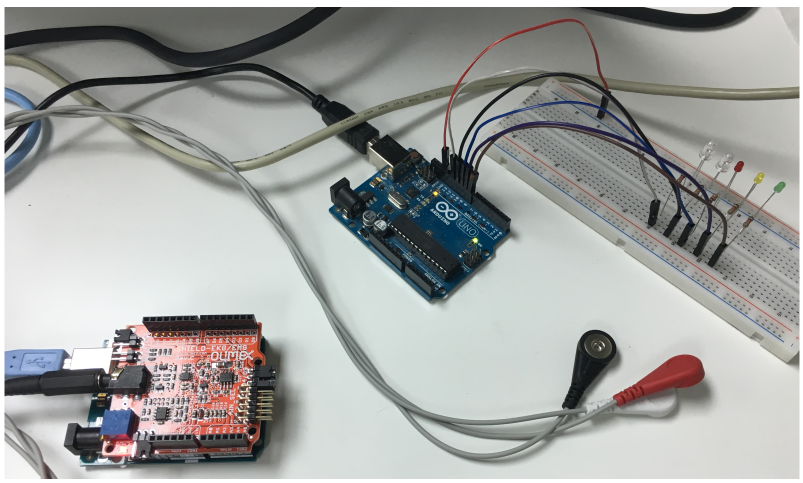


Figure 19. Visualization setup for EMG detection, including 5 LEDS for intensity response

Table 1 below provides a comparison of the existing conventional method for EMG sensing and the newly developed textile one discussed in this report. A few important criteria were chosen for this comparison, and evaluated from a perspective of augmenting the current system through a wearable technology approach. It is clear from the total weighted scores for each that the wearable EMG sensing provides several benefits over the current technology.

Table 1. Decision Matrix: Wearable vs. Conventional EMG Sensing

Category	Score Weight	Conventional (/10)	Wearable (/10)	Conventional Weight	Wearable Weight
Cost	4	8	6	32	24
Wearability	5	3	8	15	40
Application Versatility	5	6	9	30	45
Reproducibility	3	9	5	27	15
Aesthetics	3	5	8	15	40
Novelty	4	4	9	16	36
Total				**135**	**200**

4 Conclusions

Based on the analysis reported above, it can be concluded that the design and development of a dry, textile-based wearable technology for motion sensing was successfully achieved in relation to the project objectives initially outlined in the introduction.

In terms of the technology's ability to function in real-time and trigger an action with adequate precision, a software and hardware system was setup using Arduino and MATLAB to detect and process EMG signals. This system consisted of acquisition, thresholding, triggering, mapping, and processing stages, accomplished via digital filtering and typical EMG analysis techniques. On the equipment side, conventional surface electrode function was mimicked by conductive yarns that were knit into textile electrodes. Furthermore, switching to the Olimex EMG module proved to be much more effective for this wearable system as compared to the initial Myoware sensor, due to its increased capabilities in detecting signals from textile electrodes. Finally, the integrated medical electrode snaps provided the necessary surface-to-textile connection that was required for transmitting the signals to the module. These all led to a better functioning system than initially designed, improving precision, and allowing for dry-textile recording.

With regards to the wearable aspects of the product, namely comfort, washability, seamlessness and non-invasiveness, many functional specifications were investigated. Textile electrodes in general were chosen over the conventional electrodes due to their lack of irritability, non-invasive surface function, integration into fabrics, and structural durability through washes and abrasion. The 'terry' knit structure was chosen out of the four that were tested as it demonstrated the highest reproducibility from previous tests, and was the only true functioning standalone system of all three textile electrodes in a dry setting. Finally, while not practically implemented yet, the idea of an in-house Myant electronics module to replace the bulky Olimex EMG module was investigated and is being pursued. These choices are key contributors to the overarching theme of being wearable (i.e. mimicking normal clothing), user friendly, and aesthetically appealing, which are all crucial for wearable technology and everyday IOT integration. More generally, this greatly increases the commercialization potential of the product, thereby increasing company value through revenue and investors / clients.

Overall, the customer requirements were well met, and based on the decision matrix from Table 1, it is clear that the newly developed technology offers significant advantages over the existing methods, primarily in terms of user preference, commercial potential, and range of application. Given more time to develop, the wearable sleeve discussed in this report will incur significant improvement as better materials, and more advanced processing systems will be used.

5 Recommendations

Based on the analysis and conclusions drawn in this report, the main recommendations of this report are to further develop the acquisition and processing system, as well as continue optimizing the wearability and seamless integration of the technology into the user's life. Furthermore, the wearable sleeve should then be expanded to incorporate new features as the technology continues to progress. More specifically, the key steps to be taken moving forward are:

1. Develop in-house electronics module by:
    a. Investigating hardware requirements (amount of power, size of battery).
    b. Mimicking Olimex module CAD schematic to modify the multi-channel acquisition capabilities through multiplexing (common filtering system, common referencing).
    c. Incorporating wireless module(s) for Bluetooth and Wi-Fi interfacing.
2. Modify the wiring system for signal acquisition to be less dense and less obtrusive by investigating new connection methods to replace electrode cables. One possibility

includes creating embedded textile ‘wire’ lines on the sleeve, leading directly to the on-body electronics module.

3. Fabricate new garments targeting different muscles, with the end goal being to create a full body garment, capable of detecting a variety of muscle movements.
4. Adapt software system to manage multiple muscle channels for simultaneous EMG acquisition and processing.
5. Continue cross-sectional electrode study to explore and investigate new materials for conductive yarns, textile electrodes, and polymer coated electrodes.

Glossary

CAD: Computer-Aided Design – Use of computer systems to aid in the creation, modification, analysis, or optimization of a design.

Crosstalk: A phenomenon in electronics by which a signal transmitted on one circuit or channel of a transmission system creates an undesired effect in another circuit or channel. Muscle crosstalk refers to the picking up of muscle action signals from non-targeted muscles.

DIY: 'Do it yourself' – Method of building, modifying, or repairing things without the aid of experts or professionals.

ECG: Electrocardiography – Process of recording the electrical activity of the heart.

EEG: Electroencephalography – Process of recording the electrical activity of the brain.

Electrolyte: A substance that produces an electrically conducting solution when dissolved in a polar solvent, such as water.

EMG: Electromyography – Process of recording the electrical activity of skeletal muscles.

EOG: Electrooculography – Process of recording the electrical activity of the eyes.

HUD: Head-up Display – Transparent display that presents data without requiring users to look away from their usual viewpoints.

IIR: Infinite Impulse Response – Used as a type of digital filtering for signal processing.

IoT: Internet of Things – Network of physical devices, vehicles, home appliances and more that are embedded with electronics, software, sensors, and actuators, allowing for easy exchange of data.

IDE: Integrated Development Environment – Software application, typically consisting of a source code editor, build automation tools, and a debugger.

LED: Light-Emitting Diode – A two-lead semiconductor light source.

Motion artifact: A structure or appearance that is not naturally present in the system, introduced as a result of undesired movement during signal acquisition.

POC: Point of care – Point in time when clinicians deliver healthcare products or services to patients at the time of care.

SKIIN: Myant's leading in-house product, marketed as a comfortable and washable smart garment, capable of detecting body signals such as heart rate, respiration, temperature, and more.

SNR: Signal-to-noise ratio – A measure that compares the level of a desired signal to the level of the background noise.

VR: Virtual Reality – A computer technology that generates realistic sensations that stimulate a person's physical presence in a virtual or imaginary environment.

References

[1] "Wearable Technology Market - Global Opportunity Analysis and Industry Forecast, 2014-2022", *Prnewswire.com*, 2017. [Online]. Available: https://www.prnewswire.com/news-releases/wearable-technology-market---global-opportunity-analysis-and-industry-forecast-2014-2022-300460342.html.

[2] A. Michael, "Wearable Technology and Wearable Devices: Everything You Need to Know", *Wearable Devices*, 2014. [Online]. Available: http://www.wearabledevices.com/what-is-a-wearable-device/.

[3] J. Wade, "Wearable Technology statistics and trends 2018 | Smart Insights", *Smart Insights*, 2017. [Online]. Available: https://www.smartinsights.com/digital-marketing-strategy/wearables-statistics-2017/.

[4] "Wearable Technology Application Chart :: Beecham Research", *Beechamresearch.com*, 2018. [Online]. Available: http://www.beechamresearch.com/article.aspx?id=20.

[5] H. Zeng and Y. Zhao, "Sensing Movement: Microsensors for Body Motion Measurement", *Sensors*, vol. 11, no. 12, pp. 638-660, 2011.

[6] T. Agarwal, "Different Types of Motion Sensors with Applications", *Buy Electronics & Electrical Projects in Europe*, 2014. [Online]. Available: https://www.efxkits.co.uk/motion-sensors-detectors-with-applications/.

[7] *EMG Biology*. [Online]. Available: http://www.nrsign.com/wp-content/uploads/2014/02/image15.png.

[8] A. Jawahar, "Specialized Nerve Tests: EMG, NCV and SSEP", *Spine.org*, 2011. [Online]. Available: https://www.spine.org/KnowYourBack/Treatments/AssessmentTools/SpecializedNerveTests.aspx.

[9] "Technologies - Myant", *Myant*. [Online]. Available: http://www.myant.ca/technologies/.

[10] R. Davis, "Wearable Electronics | Textile World", *Textileworld.com*, 2017. [Online]. Available: http://www.textileworld.com/textile-world/features/2017/03/wearable-electronics/.

[11] A. Albulbul, "Evaluating Major Electrode Types for Idle Biological Signal Measurements for Modern Medical Technology", *Bioengineering*, vol. 3, no. 3, p. 20, 2016.

[12] K. Cherney, "Beyond Skin Deep: Understanding the Layers of the Skin", *Consumerhealthdigest.com*. [Online]. Available: https://www.consumerhealthdigest.com/beauty-skin-care/understanding-the-layers-of-the-skin.html.

[13] C. De Luca, *SURFACE ELECTROMYOGRAPHY: DETECTION AND RECORDING*. DelSys Incorporated, 2002 [Online]. Available: https://www.delsys.com/Attachments_pdf/WP_SEMGintro.pdf.

[14] *Anterior Forearm*. [Online]. Available: http://www.pelhamrehab.com/sitebuildercontent/sitebuilderpictures/anteriorforearm5.gif.

[15] *EMG Actions*. [Online]. Available: http://research.levelup.webfactional.com/wp-content/uploads/2015/04/Gestures.png.

[16] *SHIELD-EKG-EMG bio-feedback shield USER'S MANUAL*. Olimex, 2014 [Online]. Available: https://www.olimex.com/Products/Duino/Shields/SHIELD-EKG-EMG/resources/SHIELD-EKG-EMG.pdf.

Appendix A

Final Script:

```
function [] = emg_Sunflower()
% This function gathers user threshold data for their EMG signals, and then
% begins real-time acquistion and signal processing using Direct Form II
% IIR filtering, as well as the previously acquired threshold data to
% trigger an intensity-based LED response.
%   Written by: Dhilan Bekah

% Thresholding

clear; clc;

prompt = {'Please enter the desired duration for the signal recording'};
dlg_title = 'Input';
answer = inputdlg(prompt,dlg_title);

t_out = str2double(answer); % Define time duration in seconds for recording data
Fs = 1070; % Sampling Frequency [=] Hz
Fn = Fs/2; % Nyquist Frequency [=] Hz
[min,max] = get_thresh_2(t_out, Fs, Fn);
pause;

% Data

emg_realtime_2(t_out,Fs,Fn,min,max);

end
```

Triggering + Analysis Function:

```
function [n, EMG] = emg_realtime_2(t_out, Fs, Fn, min, max)
% This function uses Matlab to receive EMG signal data from the serial port.
% In this case, an Arduino board connected to an EMG shield (Olimex) is used
% to detect and send EMG data to the port. The signal is real-time filtered
% using Direct Form II implementation, and triggers an action based on the
% established thresholds. The difference between the max and min is used
% as the "Action" region, which is then arbitrarily subdivided into 5
% regions, demonstrating the relative intensity of the signal.
% The structure for this version 2 uses If statements.
%   Written by: Dhilan Bekah

% References:
% https://ccrma.stanford.edu/~jos/fp/Direct_Form_II.html
% https://ocw.mit.edu/courses/mechanical-engineering/2-161-signal-processing-continuous-and-
discrete-fall-2008/study-materials/filterstructure.pdf


% Setup to connect to Arduino
delete(instrfindall)     %Resets the Com Port
delete(timerfindall)     %Delete Timers

led = arduino('/dev/cu.usbmodem1411','UNO') % LED action for response of signal. Comment out if
another action is preferred

sp = serial('/dev/cu.usbmodem1421');

command = 'A' ;  %Needed for fprintf function

        %Parameters for object in port. Good Housekeeping practice
        BaudRate = 57600;
        InputBufferSize = 57600;
        Timeout = 5;
        set(sp, 'BaudRate', BaudRate);
        set(sp, 'InputBufferSize', InputBufferSize);
        set(sp, 'Timeout', Timeout);


% Begin Acquisition and Processing
```

```
[b1,a1] = butter(4,40/Fn,'high'); % Create highpass filter at 40Hz
[b2,a2] = butter(2,500/Fn,'low'); % Create lowpass filter at 500Hz
[b3,a3] = butter(2,5/Fs,'low'); % Create lowpass filter for signal envelope

iirdf2_high('initial',b1,a1); % Initialize filters
iirdf2_low('initial',b2,a2);
iirdf2_env('initial',b3,a3);

status = 1;
action = (max-min)/5;
count = 1;

fopen(sp); % Open Serial Port
tic
while toc <= t_out

    fprintf(sp,command);
    raw = fgetl(sp); % Get raw signal bit data from serial port
    n(count) = count;
    EMG(count) = str2double(raw);

    y_out(count) = iirdf2_high(EMG(count));
    y_filt(count) = iirdf2_low(y_out(count));
    rec_y(count) = abs(y_filt(count));
    y_env(count) = 2*iirdf2_env(rec_y(count));

    if y_env(count) <= min
        if status == 1 || status == 2 || status == 3 || status == 4 || status == 5
            %fprintf('Off \n')
            writeDigitalPin(led,8,0); % LED action for response of signal. Comment out if
another action is preferred
            writeDigitalPin(led,9,0);
            writeDigitalPin(led,10,0);
            writeDigitalPin(led,11,0);
            writeDigitalPin(led,12,0);
            status = 0;
        end
    elseif (y_env(count) > min && y_env(count) <= (min + action))
        if status == 0 || status == 2 || status == 3 || status == 4 || status == 5
            %fprintf('Very Low \n')
            writeDigitalPin(led,8,1); % LED action for response of signal. Comment out if
another action is preferred
            writeDigitalPin(led,9,0);
            writeDigitalPin(led,10,0);
            writeDigitalPin(led,11,0);
            writeDigitalPin(led,12,0);
            status = 1;
        end
    elseif (y_env(count) > (min + action) && y_env(count) <= (min + 2*action))
        if status == 0 || status == 1 || status == 3 || status == 4 || status == 5
            %fprintf('Low \n')
            writeDigitalPin(led,8,1); % LED action for response of signal. Comment out if
another action is preferred
            writeDigitalPin(led,9,1);
            writeDigitalPin(led,10,0);
            writeDigitalPin(led,11,0);
            writeDigitalPin(led,12,0);
            status = 2;
        end
    elseif (y_env(count) > (min + 2*action) && y_env(count) <= (min + 3*action))
        if status == 0 || status == 1 || status == 2 || status == 4 || status == 5
            %fprintf('Mid \n')
            writeDigitalPin(led,8,1); % LED action for response of signal. Comment out if
another action is preferred
            writeDigitalPin(led,9,1);
            writeDigitalPin(led,10,1);
            writeDigitalPin(led,11,0);
            writeDigitalPin(led,12,0);
            status = 3;
        end
    elseif (y_env(count) > (min + 3*action) && y_env(count) <= (min + 4*action))
        if status == 0 || status == 1 || status == 2 || status == 3 || status == 5
            %fprintf('High \n')
            writeDigitalPin(led,8,1); % LED action for response of signal. Comment out if
another action is preferred
            writeDigitalPin(led,9,1);
            writeDigitalPin(led,10,1);
```

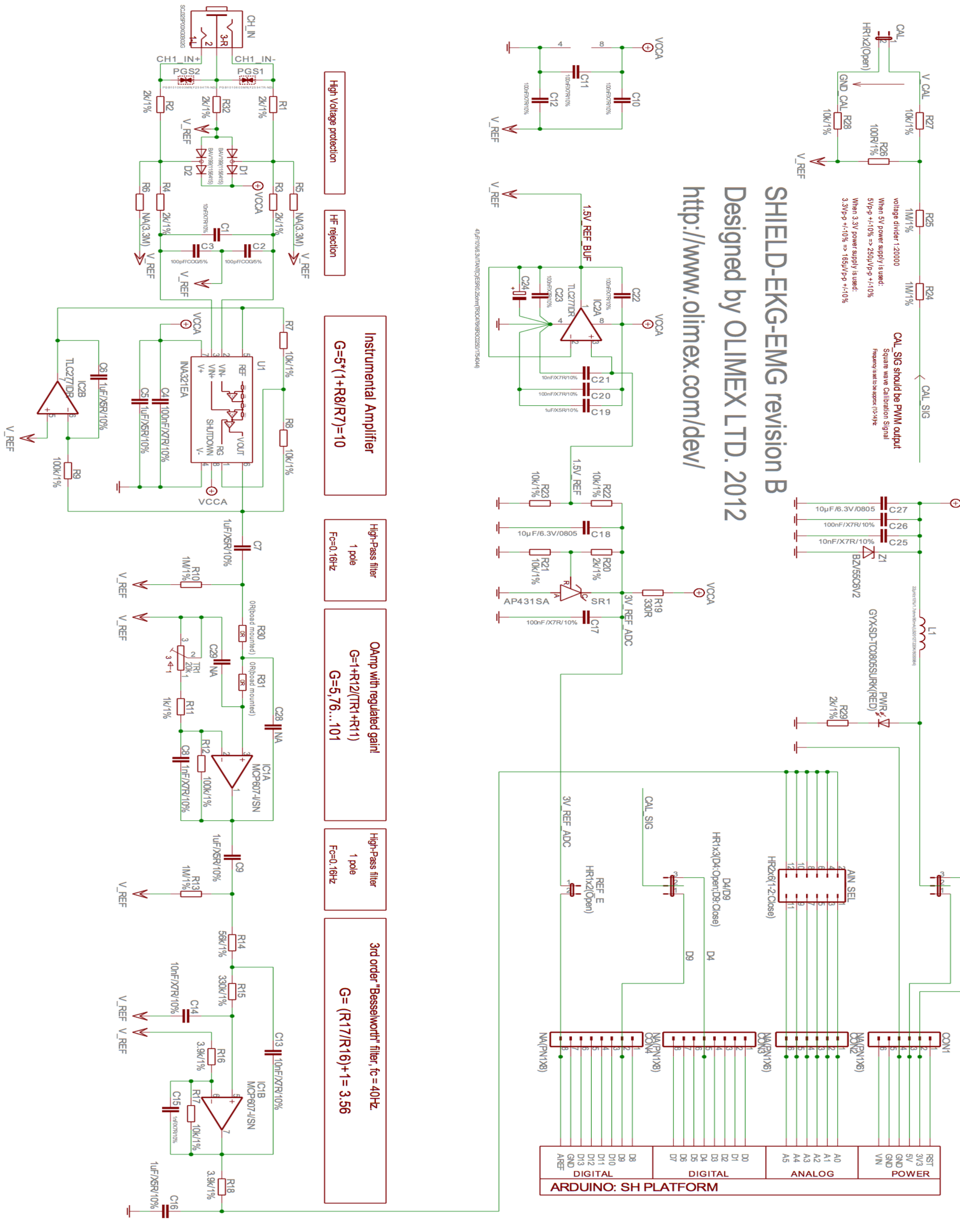
```
            writeDigitalPin(led,11,1);
            writeDigitalPin(led,12,0);
            status = 4;
        end
    elseif (y_env(count) > (min + 4*action))
        if status == 0 || status == 1 || status == 2 || status == 3 || status == 4
            %fprintf('Very High \n')
            writeDigitalPin(led,8,1); % LED action for response of signal. Comment out if
another action is preferred
            writeDigitalPin(led,9,1);
            writeDigitalPin(led,10,1);
            writeDigitalPin(led,11,1);
            writeDigitalPin(led,12,1);
            status = 5;
        end
    end

    count = count + 1;

end

fclose(sp); % Close Serial Port

A = [n;EMG];
fid = fopen('EMG_RT_data.txt','w+');
fprintf(fid, '%f %f \n', A);
fclose(fid);

figure;
subplot(2,2,1);
plot(n,EMG);
xlabel('Sample number');
ylabel('Magnitude (Raw Bit)');
title('Raw EMG Signal');

subplot(2,2,2);
y_volt_filt = (((y_filt/1024)*5)/2084)*1e6;
plot(n,y_volt_filt);
xlabel('Sample number');
ylabel('Magnitude (\muV)');
title('Filtered EMG Signal');

subplot(2,2,3);
rec_volt_y = (((rec_y/1024)*5)/2084)*1e6;
y_volt_env = (((y_env/1024)*5)/2084)*1e6;
plot(n,rec_volt_y,n,y_volt_env);
xlabel('Sample number');
ylabel('Magnitude (\muV)');
legend('Signal','Envelope');
title('Rectified EMG signal Envelope');

FEMG = abs(fft(y_filt));
L = length(FEMG);
f_Hz = (Fs/L)*(0:L-1);
L_2 = ceil(L/2);
subplot(2,2,4);
plot(f_Hz(2:L_2+1), FEMG(1:L_2))
xlabel('Frequency (Hz)')
ylabel('Magnitude');
title('Single-sided Magnitude spectrum (Hertz)');
axis tight

end
```

Thresholding function:

```
function [min, max] = get_thresh_2(t_out, Fs, Fn)
% This function uses real-time filtering (Direct Form II IIR) to produce a
% signal envelope, which is then used to trigger an action based on the
% established threshold. The method of determining the different
% intensities of the signal is done using a rest period for the first 3/4
% of the acquisition period, and then a maximum voluntary contraction for
% the last 1/4. The means of both periods are returned as the min and
% max thresholds, respectively.
% Version 2
```

```matlab
%   Written by: Dhilan Bekah

delete(instrfindall)     % Resets the Com Port
delete(timerfindall)     % Delete Timers

sp = serial('/dev/cu.usbmodem1421');

command = 'A' ;  % Needed for fprintf function

        % Parameters for object in port. Good Housekeeping practice
        BaudRate = 57600;
        InputBufferSize = 5760;
        Timeout = 5;
        set(sp, 'BaudRate', BaudRate);
        set(sp, 'InputBufferSize', InputBufferSize);
        set(sp, 'Timeout', Timeout);

[b1,a1] = butter(4,40/Fn,'high'); % Create highpass filter at 40Hz
[b2,a2] = butter(2,500/Fn,'low'); % Create lowpass filter at 500Hz
[b3,a3] = butter(2,5/Fs,'low'); % Create lowpass filter for signal envelope

iirdf2_high('initial',b1,a1); % Initialize filters
iirdf2_low('initial',b2,a2);
iirdf2_env('initial',b3,a3);

count = 1;

fopen(sp); % Open Serial Port
tic
while toc <= t_out

    if count == 1
        fprintf('Rest \n'); % First 75% is rest
    elseif count == (Fn*t_out)*3/2
        fprintf('Maximum flex \n'); % 25% remaining is contraction
    end

    fprintf(sp,command);
    raw = fgetl(sp); % Get raw signal bit data from serial port
    n(count) = count;
    EMG(count) = str2double(raw);

    y_out(count) = iirdf2_high(EMG(count));
    y_filt(count) = iirdf2_low(y_out(count));
    rec_y(count) = abs(y_filt(count));
    y_env(count) = 2*iirdf2_env(rec_y(count));

    count = count + 1;

end

fclose(sp); % Close Serial Port

plot(n,y_env);

min = mean(y_env(1:(Fn*t_out)*(3/2)));
max = mean(y_env(((Fn*t_out)*(3/2))+10:end));

end
```

Appendix B


SHIELD-EKG-EMG revision B
Designed by OLIMEX LTD. 2012
http://www.olimex.com/dev/
High Voltage protection
HF rejection
Instrumental Amplifier
G=5*(1+R8/R7)=10
High-Pass filter
1 pole
Fc=0.16Hz
OAmp with regulated gain!
G=1+R12/(TR1+R11)
G=5,76...101
High-Pass filter
1 pole
Fc=0.16Hz
3rd order "Besselworth" filter, fc=40Hz
G= (R17/R16)+1= 3.56
ARDUINO: SH PLATFORM
DIGITAL
DIGITAL
ANALOG
POWER