Development of a 3D-Printed EMG Wristband

by

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Abstract

Electromyography (EMG) has been widely used in robotics and biomedical applications for sensing and diagnostic purposes. Because of the complex shape of the human limbs and the uneven surface of the human skin, EMG sensing often faces the challenge of stable signal detection. As manufacturing technology advances, Additive manufacturing has shown its potential to improve the existing EMG sensing system further. 3D printing offers the advantage of customized fabrication to fit the designated locations of EMG detection. In this work, a 3D-printed EMG wristband was developed for the control application. The wristband demonstrated its stability in EMG signal processing with a single-sensor system, which differed from other multi-sensor system devices. 3D-printed serpentine electrodes have improved signal detection by providing conformal contact to the skin compared with the existing rigid dry electrodes. This work shows 3D-printed EMG device's potential for the advanced control system in the prosthetic fields.

Keywords: 3D printing; wristband; electromyography; surface dry electrodes; sensing; serpentine structure; prosthetics

Dedication

I dedicate this thesis to my family for their love and support in my life.

Acknowledgments

I would like to express my sincere gratitude to the following individuals and groups for their support and guidance during my master's thesis project:

- My supervisor, Dr. Woo Soo Kim, for his support and encouragement throughout the project. His guidance and comments have been valuable to me. I am truly grateful to learn under his supervision during these two years.

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List of Acronyms

ADC	Analog to Digital Converter
CB	Carbon Black
DIW	Direct Ink Writing
DLP	Digital Light Process
EcoFlex	Platinum-Catalyzed Silicones
EHD	Electrohydrodynamic
EMG	Electromyography
FDM	Fused Deposition Modeling (FDM)
GTL	Gerber File
iEMG	Intramuscular Electromyography
LIG	Laser Induced Graphene
MWCNT	Multiwalled Carbon Nanotube
PCB	Printed Circuit Board
PETG	Polyethylene Terephthalate-glycol
PLA	Polylactic Acid
RMS	Root Mean Square
sEMG	Surface Electromyography
SLA	Stereolithography
SNR	Signal-to-Noise Ratio
SVM	Support Vector Machine
TPU	Thermoplastic Polyurethanes
TXT	Drill File
UV	Ultraviolet

Chapter 1.

Introduction

1.1. Literature Review

Today, millions of patients worldwide are suffering from arm amputation [1] and the challenges associated with a lack of limbs [2]. Past data has shown that in 2005 there were 1.6 million amputees living in the United States, and that number has been steadily increasing since [3]. As a result, the demand for prosthetic limbs has been growing over time [4]. In the last decade, statistically 1 in 190 people are living with missing limbs, and the number is projected to double by 2050 [5]. However, the high cost (ranging from USD $9,000 to $40,000) and system complexities are not suitable or accessible for most patients [6]. Further, most prosthetic hands are cosmetic, or only limited to the basic functionality of grabbing and gripping light objects. Surface EMG signal processing enables patients to have direct control of their prosthesis by utilizing electrical signals from their forearms, which are obtained by the usage of surface EMG dry electrodes [7]. Through physical contact, the patients could also interact with the sensing robots by gestures [8]. Due to the curvature of the arm, flexible and stretchable electrodes are desired to be developed in order to provide better signal acquisition [9]. Then, the electrical signals generated from forearm muscles can be securely detected and used to recreate any hand movements by using the portable EMG sensor [10]. This advantage improves the suitability of the prostheses and reduces control difficulty for the users [11]. Therefore, several designs of the EMG sensing system have been developed for signal detection and gesture recognition purposes [12-14]. However, they are not suitable to be implanted in a wearable device on the arm due to the system's complexity. Generally, prosthetic hands can perform grasping and releasing motions controlled by a group of EMG signals measured in different arm locations [15]. The muscle force is also related to the intensity of the EMG signal [16]. Even though the electrode has simple implementations on the signal acquisition, there are also some limitations. For instance, the electrode position must be stationary for ideal measurement and can be easily dislocated or shifted during arm movement [17]. Therefore, a wearable wristband can provide consistent EMG signal monitoring at the designated forearm locations [18]. The wristband does not limit the user's exercise postures when monitoring [19]. Currently, several EMG acquisition system products have been commercially released in the market, such as the Myo armband, gForcePro+ armband, and Delsys Trigno Avanti. The Myo armband consists of 8 channels that offer human-computer interaction with basic hand gestures (USD $200) [20]. The gForcePro+ armband is a surface EMG (sEMG) system with up to 18 customizable gestures. However, the increased number of gestures also increased the cost of the product to USD $1500 compared to the Myo armband [21]. The Delsys Trigno Avanti is a medical-graded EMG device. A summary of the main wearable EMG devices and the wristband from this study is presented in Table 1.1. However, the device requires its own data acquisition software which limits its application outside of medical purposes [22]. The dry electrodes are used on all these devices for EMG signal detection. Compared with the wet electrodes [23] and the conventional Ag/AgCl electrodes [24], the dry electrodes provide an overall preferable performance. As 3D printing manufacturing technology develops, the material cost and manufacturing times continue to decrease [25]. Thus, the wristband can be fully 3D printed to become a cost-effective device for most users and patients. Several methods of 3D printing have been widely used for prototyping, such as stereolithography (SLA), direct ink writing (DIW), and digital light process (DLP). Among these, DIW is the common method for printing electrodes [26]. Due to the low weight of the printing material, it does not require additional strength from the user to operate and wear [27]. In addition to this, 3D printing technology allows the device to be customized since the arm sizes vary as per patient.

Table 1.1. Properties of the main wearable EMG devices

Name	Number of Channels	Bandwidth Filter Range (Hz)	Electrode Material	Sampling Rate (Hz)	Price ($USD)
Myo Armband	8	20-500	Stainless steel	200	200
gForcePro+ Armband	8	20-500	Stainless steel	1000	1,250
Delsys Trigno Avanti	16	20-450	Silver	2000-3000	20,000
3D-printed Wristband	5	20-500	Flexible silver ink	1000	70

1.2. Motivation

Currently available EMG wearable devices for detecting hand movements are limited in their ability to be customized due to their fixed designs and high cost. To enable cost-effective customization of EMG devices, the development of flexible manufacturing processes such as 3D printing is crucial. In this research project, we aim to develop an affordable and customizable 3D-printed surface EMG wristband. This prototype will reduce the difficulties of prosthetic control and become an easy-to-wear device that can aid patients in their daily lives.

1.3. Objectives

The objective of this research project is to design and fabricate a wearable wristband with conformal dry electrodes using 3D printing technology that can perform EMG signal detection on the user's forearm. The project aims to achieve the following objectives:

1. To design a custom EMG wristband using 3D printing technology for tracking muscle activities.
2. To design dry electrodes using 3D printing technology to improve signal detection.
3. To develop an EMG signal control system that can quantitatively analyze muscle activities.
4. To fabricate an EMG wristband system using advanced 3D printing technology.

1.4. Contribution

The Additive Manufacturing Laboratory (AML) at Simon Fraser University has extended its studies on 3D printable sensing systems for prosthetic applications, laying the foundation for 3D-printed EMG-controlled wearable devices. In this research, a 3D-printed wristband was developed that demonstrated the ability to acquire valid EMG signals from muscle activities. The following publications have supported this research:

- **H. Su**, T. H. Kim, H. Moeinnia and W. S. Kim, “A 3-D-Printed Portable EMG Wristband for the Quantitative Detection of Finger Motion.” *IEEE Sensors Journals* 23 (7), 7895-7901 (2023).

- **H. Su**, H. Moeinnia and W. S. Kim, “3D Printed Electromyography Sensing Systems.” *Advanced Sensor Research (in press)* (2023).
- H. Moeinnia, **H. Su**, and W.S. Kim, “Novel Grasping Mechanisms of 3D-Printed Prosthetic Hands.” *Advanced Intelligent Systems* 4 (11), 2200189 (2022).
- C. Bao, **H. Su**, S. K. Seol and W. S. Kim, “A 3D Printed Portable Neuromorphic System.” *IEEE Sensors Letters* 6 (2), 5500203 (2022).

1.5. Thesis Outline

This thesis aims to explore the design and fabrication of an EMG wristband device that includes surface dry electrodes and a signal processing system. Chapter 2 provides background information on current 3D printing techniques used in EMG sensing. Chapter 3 details the development of the 3D-printed wristband and the fabrication of each component. Chapter 4 presents the testing and validation of the wristband. Finally, Chapter 5 summarizes the research study and provides potential avenues for future work and improvements for the next wristband iteration.

Chapter 2.

3D-Printed Electromyography Sensing System

2.1. Introduction to Current 3D Printing Techniques

An EMG sensing system includes several components which must be fabricated with different materials. These materials vary from polysiloxane (silicone) and thermoplastic elastomer as a substrate to silver ink for printing flexible electrodes [28]. 3D printing technology has been growing significantly for rapid prototyping manufacturing. It is now feasible to perform multi-material printing within the layers [29-30]. In addition, specific 3D printing methods have achieved multi-material fabrication on a microscale [31-32]. Thus, 3D printing is becoming a promising technology for fabricating EMG sensing systems.

2.1.1. Fused Deposition Modeling (FDM)

Fused deposition modeling (FDM) is a 3D printing technique where the deposition of filaments forms an object through a heated nozzle on a building plate. The nozzle extrudes a thin layer of melted filament to build the structure with stacked layers [33]. This method is widely used among researchers to study and develop. FDM can use various materials, from soft to rigid, such as plastics/polymers, metals, and ceramics [34]. Both iEMG (Intramuscular Electromyography) and sEMG electrodes can be fabricated through the FDM technique. Researchers have demonstrated the capability of FDM in fabricating iEMG sensors. In recent research, a wired microelectrode using polyethylene terephthalate-glycol (PETG) as the filament was printed by Prusa i3 MMU2S (Prusa Research, Czech Republic), shown in Figure 2.1.a [35]. The insulted body was half-printed first, and the metal wires were then loaded into the body. The remaining half body was printed to complete the electrode structure. However, due to the fabrication complexity of iEMG electrodes, researchers have widely studied the sEMG. Material selection is a crucial step in designing a functional sEMG sensor. sEMG electrodes can be fabricated with various materials with the help of current advances in additive manufacturing techniques. Vaněčková et al. [36] manufactured a 3D printed electrode using polylactic acid (PLA)/carbon black (CB) filaments, shown in Figure 2.1.b. The 3D printed electrode showed nearly ideal electrochemical behavior comparable to the conventional metallic and carbon electrode. In addition to the rigid and brittle materials, soft and flexible are also could be used for the sEMG electrode fabrication. Dijkshoorn et al. [37] designed a conductive sensor by embedding copper tape during the FDM process. The sample showed robustness to the contacts in tensile testing, which has potential for biomedical applications. Wolterink et al. [38] used thermoplastic polyurethanes (TPU) to print a low-cost, flexible sEMG electrode, shown in Figure 2.1.c. The conductive and insulated parts are printed concurrently by the conductive and non-conductive TPUs on Flashforge Creator Pro (FlashForge Corporation, China). By using the electronically conductive composite filaments, the FDM can also 3D print electronics [39]. 3D printing, also, can fabricate substrates with dielectric and electric properties essential to electronics fabrication [40]. Wu et al. [41] constructed a passive wireless sensor to monitor the quality of liquid food. The structures were built with a multi-nozzle FDM printer. The system allows the electronics to be integrated into a 3D object at a lower cost than traditional manufacturing technology [42].

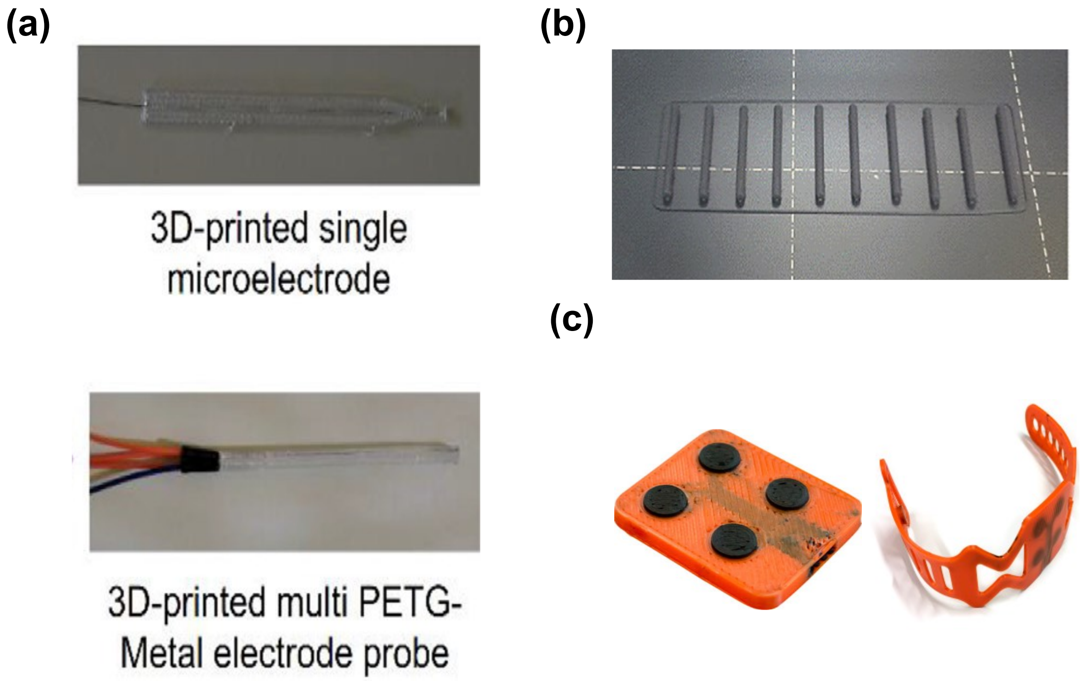

Figure 2.1. **(a) 3D printed microelectrodes and multi-electrode probes with embedded etched metal and conductive wires. Reproduced with permission.[35] Copyright 2021, Elsevier. (b) cylindrical shape of 3D printed electrodes using PLA-CB filaments by FDM printing. Reproduced with permission.[36] Copyright 2020, Elsevier. (c) 3D-Printed electrodes using conductive TPU (black) and non-conductive TPU (orange), which are flexible and elastic. Reproduced with permission.[38] Copyright 2017, IEEE.**

2.1.2. Direct Ink Writing (DIW)

Direct ink writing (DIW) is an extrusion-based printing technique in which material is deposited in a controlled pattern. It can also perform multi-material printing to create complex 3D shapes [43]. Studies have shown that wood, metals, and ceramics can be deposited through this method [44-46]. Several DIW printing mechanisms can be used for material extrusion. The pneumatic extrusion uses air pressure through a flexible tube to control the extrusion rate [47]. Chen et al. [48] utilized the pneumatic extrusion with multi-nozzle to fabricate a sensing pad, shown in Figure 2.2.a. It has a soft and rigid structure that can be applied to the flexible finger for robotic study. Screw extrusion is another mechanism that translates a rotary screw to a pumping action to extrude materials [49]. Instead of filaments, pellets are commonly used in screw extrusion. The rotating screw feeds the pellets into the nozzle after being melted from the melting zone [50]. The plunger extrusion is the most common mechanism used in DIW printing, which uses a plunger to push the feedstock into the nozzle [51]. It was also referred to as piston extrusion or ram extrusion. Christ et al. [52] designed a highly elastic sensor printed by multi-material plunger extrusion, shown in Figure 2.2.b. The thermoplastic polyurethane/multiwalled carbon nanotube (TPU/MWCNT) was a flexible material with excellent piezoresistive behavior.

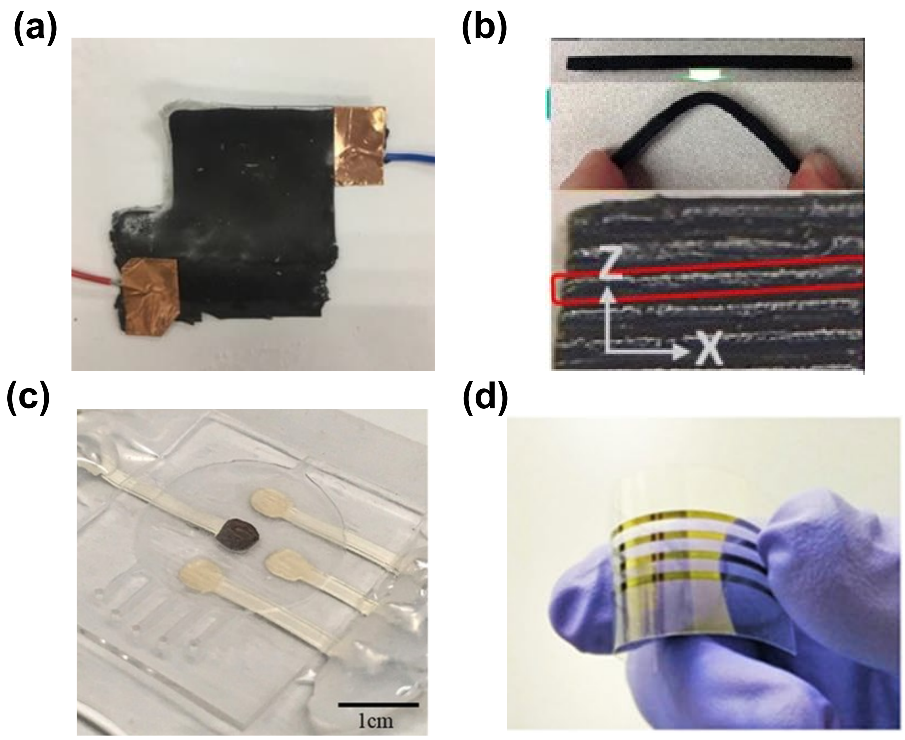

Figure 2.2. **(a) A piezoelectric sensing pad using PVDF (Polyvinylidene difluoride)/graphene by UV light curing. Reproduced under the terms of the CC-BY license.[48] Copyright 2020, MDPI. (b) FDM 3D printed flexible TPU/MWCUNT (multiwalled carbon nanotube) nano composites as an elastic strain sensor. Reproduced with permission.[52] Copyright 2017, Elsevier. (c) a wearable patch with 3D-printed flexible ion sensors and a microfluidic unit. Reproduced with permission.[54] Copyright 2021, Elsevier. (d) a flexible strain sensor based on the EHD-printed circuits on TPU substrates for motion monitoring. Reproduced under the terms of the CC-BY license.[56] Copyright 2021, Springer.**

2.1.3. Other 3D Printing Techniques

Aside from the two standard 3D printing techniques, some other techniques can be applied to fabricate an EMG sensing system, such as inkjet printing and electrohydrodynamic (EHD) jet printing, and digital light processing (DLP) printing. Inkjet printing ejects ink droplets onto a substrate and solidifies them with a curing process, such as ultraviolet (UV) curing [53]. Kalkal et al. [54] reported that wearable sensors with polymer substrate had been developed using inkjet printing, shown in Figure 2.2.c. The sensors can be directly applied to the biomedical field, including EMG sensing. The EHD jet printing creates a fluid flow using an electric field to deposit inks onto a substrate [55]. Khan et al. [56] fabricated a soft wearable sensing system utilizing EHD jet printing, shown in Figure 2.2.d. The system was designed to monitor the muscle motion of humans, such as finger bending. DLP printing uses liquid resin to create objects layer by layer, cured by ultraviolet (UV) light from a digital light projector [60]. Lopez-Larrea et al. [61] developed 3D-printed hydrogels with conductive inks using poly (3,4-ethylenedioxythiophene) (PEDOT). The printed hydrogels demonstrated high conductivity and long-term reproducible detection for EMG recording.

Table 2.1. Summary of 3D-printing technique for EMG sensing system

Technique	Materials	Strengths	Limitations
Fused deposition modeling (FDM)	Thermoplastics	Low-cost; material availability; easy to use; customizable	Low accuracy; rough surface finish; require the use of supports
Direct ink writing (DIW)	Polymer; hydrogels; ceramics; bioprinting	Freeform; material flexibility; complex geometries	Low build volume; slow printing speed; ink viscosity
Inkjet printing	Polymer; composites; metal; ceramics	High speed; low waste; High scalability	Limited strength; limited accuracy; more energy consumption
Electrohydrodynamic (EHD) jet printing	Polymer; conductive ink; composites	High resolution; high precision; biocompatibility; high porosity	Complexity; post-processing; quality control; alignment issue

2.2. EMG Detection in Printed Sensing System

Electromyography (EMG) is a technique that measures the electrical activity generated from the muscle fibers during muscle contraction [57]. Currently, there are two main methods of measuring muscle activity which are intramuscular EMG (iEMG) and surface EMG (sEMG), as shown in Figure 2.3. [58]. The iEMG demonstrated a more specific measurement of muscle signals than the sEMG from four different muscles. However, the sEMG provided a similar result of collected signals with a less invasive performance. The iEMG detects the signals by placing the needles or wires into the body [59]. It can acquire EMG signals from muscle fiber sources [60]. However, iEMG could cause potential risks after the operation [61]. On the other hand, sEMG has been commonly used in clinical applications and research studies [62]. It collects the EMG signals on the skin surface by the surface electrodes [63]. With the advantage of non-invasive electrodes, the sEMG is suitable for developing wearable devices in muscle activity measurement, such as EMG armbands [64-65]. Therefore, sEMG has become preferable to iEMG in the bio-medical field. By utilizing the EMG signals, several human diseases can be detected. Also, injury prevention could be provided by measuring and analyzing sEMG data during physical activities, including sports, occupational and daily activities. For example, the gait is generated by analyzing EMG signals, which are commonly applied to diagnose 'Parkinson's disease and spondylosis [66]. Furthermore, EMG signals can also be used to evaluate muscle dysfunction through neurophysiological evaluation [67].

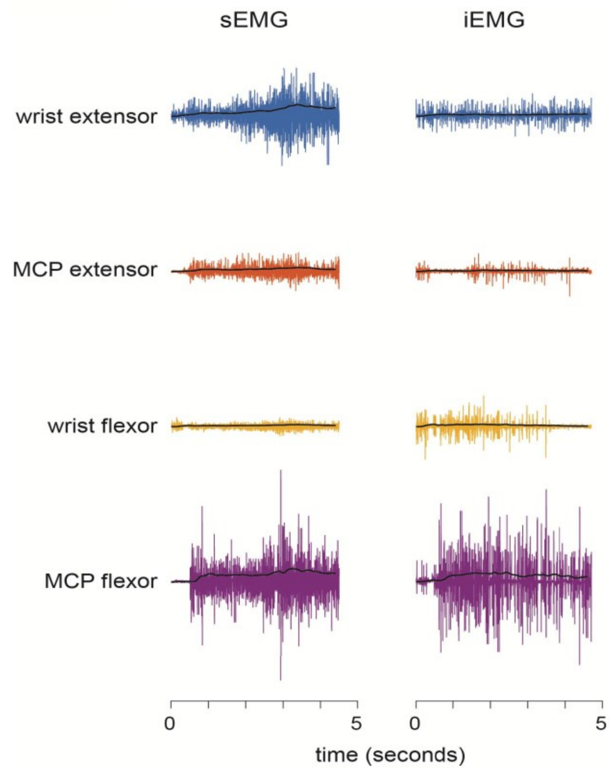

Figure 2.3. **Raw and filtered (black line) EMG signals of sEMG and iEMG were collected from selected muscle movements. Reproduced with permission.[58] Copyright 2018, IEEE.**

Nowadays, there are several EMG devices available in the market. The EMG electrodes play an essential role in the sensing system to acquire muscle activity in EMG signals. Commercial EMG devices mainly use solid shapes with rigid dry electrodes, such as gForcePro+ (Oymotion) and Myo armband (Thalmic Labs) [68]. In clinical practice, conventional wet electrodes (Ag/AgCl) are used to form a conductive layer on the skin for diagnosing [69]. The inserted electrodes are designed to perform the iEMG data acquisition. These electrodes are fabricated through a micromachining process with a laser on small scales [70]. Compared to wet electrodes, the inserted electrodes can record more distinguishable and more stable biological signals [71]. In recent years, stretchable dry electrodes have been developed to improve the user's comfort and address the skin adhesion issues. Due to the outstanding material property of polydimethylsiloxane (PDMS) in providing flexibility and printability, it has become a suitable substrate material choice in stretchable wearable devices [72]. Through the electroplating and standard photolithography fabrication process, the Au and Cu are patterned on the PDMS substrate and accordingly an inexpensive stretchable dry electrode could be fabricated [73]. Moreover, a different approach to nanofiber carbon electrodes was developed through electrospinning [74].

As additive manufacturing (3D printing) technology develops, it enables customizable structures for the EMG electrodes. Compared with the manufacturing process, 3D printing allows researchers to create suitable electrodes based on the experimental demands [75]. In particular, electrodes with complex internal structures can be fabricated in a shorter time using 3D printing [76]. Moreover, 3D printing expands the use of materials in manufacturing the electrodes, such as polymer, carbon, and ceramics [77]. This has reduced the cost of production to fulfill different applications [78]. Therefore, additive manufacturing has become preferable to traditional manufacturing techniques [79-80]. Many research studies on 3D printable electrodes have recently been conducted [81-83]. To utilize the EMG signals in a real-world application, it is critical to process the data using a reliable signal acquisition system [84]. By combining the 3D printed electrodes with EMG data acquisition systems, the EMG signals can be used in the advanced field, such as the control of prosthetics [85].

The 3D-printed EMG sensing system collects the raw EMG signals from the muscle contractions. By using 3D printing techniques, the devices can be flexible to fully cover the human skin compared with conventional rigid devices [86]. However, the raw EMG signals must be preprocessed to acquire interpretable data for further applications. Filtering [87] and amplifying [88] are two essential steps after obtaining raw EMG data. First, the raw signals are filtered to remove the noise signals within a range of the assigned high and low pass frequencies. Then, the filtered signals are increased in amplitudes through the amplifier to be recorded. Traditionally, the signal preprocessing was performed following encoded electronic circuits [89]. The 3D-printed electronic circuit boards advanced a suitable and customizable platform for EMG sensing applications. The printed sensing system has been widely used in wearable applications [90]. Due to the adaptive customization ability of 3D printing, the sensing system can be developed individually for compatibility. Bao et al. [91] presented a 3D-printed sensing system with signal conversion and transmission functionality, shown in Figure 2.4. Using 3D printing technology, they fabricated all the components, including inductors, capacitors, and resistors.

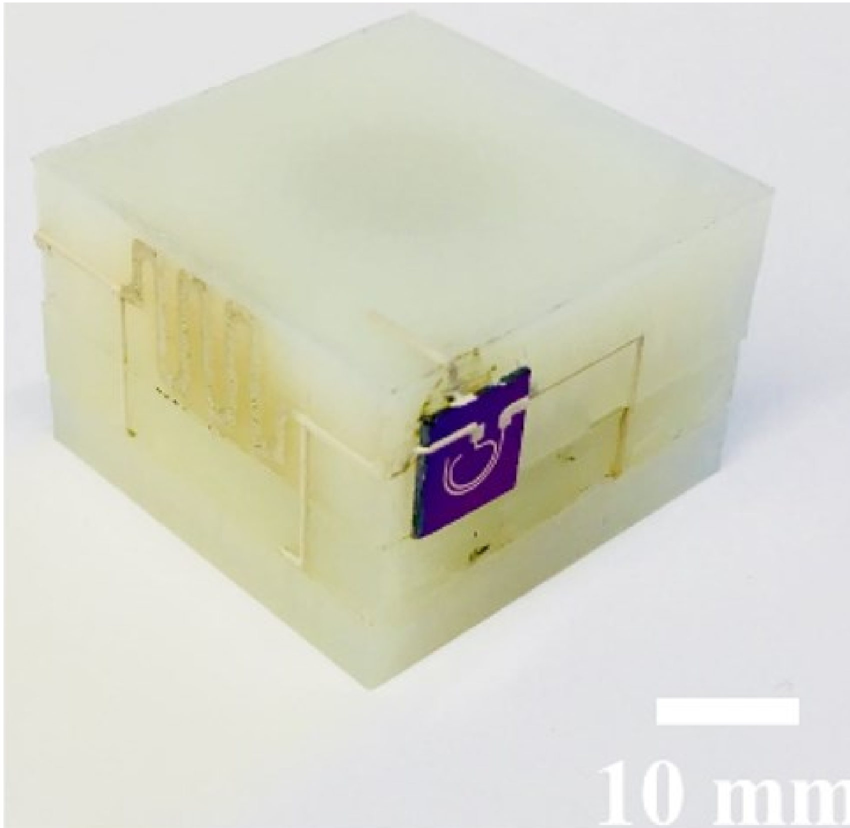

Figure 2.4. **3D-printed sensing system for mimicking sensory receptors, neurons, and synapses using printed components. Reproduced with permission.[91] Copyright 2021, Elsevier.**

2.3. Applications of Printed Sensing System

As one of the critical components in the EMG sensing system, different patterns and textures of printed EMG electrodes have been developed to improve signal detection for EMG analysis. Pani et al. [92] designed a circular textile electrode where the conductive ink is printed on a piece of cotton fabric, shown in Figure 2.5.a. After printing, the electrode is cured in an oven at 70 °C for 15 minutes as post-treatment. The printed electrode was compared to the conventional gelled electrode in signal acquisition. The noise and raw signals detected by both electrodes were similar, which suggested that the printed electrode can be used as an alternative in EMG detection. Spanu [93] also reported a stretchable textile electrode for EMG detection on the leg muscles. The movement of leg muscles requires significant stretching, which is challenging for maintaining consistent surface conductivity. The stretchable textile electrode remained stable after 500 stretch cycles while providing accurate signal readings compared with AG/AgCl electrodes. The number of ink layers was also optimized to increase the conductivity of the printed electrodes. As a result, the three layers of printed electrode showed the best conductivity. In addition, the size of the printed electrode was reduced due to closer contact with the skin. Scalisi et al. [94] used inkjet print to fabricate a flexible electrode matrix based on conductive silver ink. The sample created a lower skin-electrode impedance which further improved the detection of EMG signals in sEMG. In particular, the electrode matrix performed well in the low-pressure, high-frequency recording scenario. The novel printing method allows the electrode matrix to be customized to devices with minimum modifications. Toral et al. [95] introduced laser-induced graphene (LIG) electrodes as a low-cost material with a simple fabrication process, shown in Figure 2.5.b. The stretchable LIG electrode collected a higher root mean square (RMS) value than the traditional AgCl electrode, which demonstrated that the LIG electrode removed more noise signals overall. Furthermore, the structure of the printed electrode gave a more comfortable fit for the user compared to the commercial ones. With the 3D printed electrodes having similar or better performance than commercial electrodes, they have been implanted for the analysis of gesture recognition. Abass et al. [96] employed 3D-printed conductive PLA electrodes to replace the snap-on flat electrodes with flexibility in customization. In the study, the EMG data from printed electrodes could distinguish five different gestures through a support vector machine (SVM) classifier. Five PLA electrodes were fabricated to detect the EMG signals on the user's forearm. The PLA electrodes performed the gesture classification with an average of 85% accuracy. In another study, Rosati et al. [97] approached gesture recognition with the printed electrode matrix structure. The electrode consists of eight channels and is positioned on the forearm for EMG detection. Through a series of tests with multiple volunteers, the accuracy of gesture classification was 93-95% consistently across the volunteers. These studies have shown the stability of the 3D printed electrodes as reliable sensing devices in EMG detection and analysis.

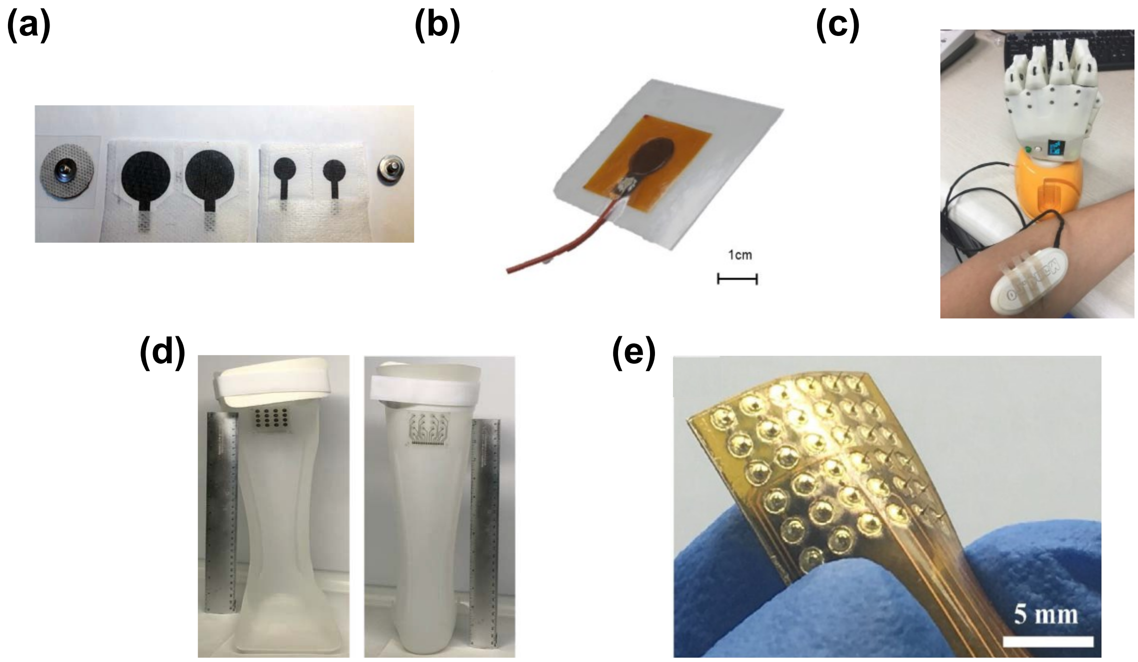

Figure 2.5. **(a) Printed circular textile electrodes on cotton fabric (black) compared with Ag/AgCl electrodes (silver) at 24mm and 10mm in diameter. Reproduced under the terms of the CC-BY license.[92] Copyright 2019, IEEE. (b) a developed stretchable LIG electrode with a multi-layer structure, including electrode (black), wire, isolation layer (orange), and flexible adhesive (white). Reproduced under the terms of the CC-BY license.[95] Copyright 2020, IEEE. (c) prosthetic hand connection with EMG electrodes, the detected signals from forearm muscle controls prosthesis movement. Reproduced under the terms of the CC-BY-NC 4.0 license.[100] Copyright 2019, Thieme. (d) electrodes and tracks printed on the orthosis at the position of the gastrocnemius muscle for EMG signal detection. Reproduced with permission.[102] Copyright 2021, IEEE. (e) flexible micro-needle array electrode pad reinforces the contact with skin to ensure its compatibility and conductivity. Reproduced with permission.[105] Copyright 2017, Elsevier.**

Other than injury prevention and diagnostic applications, EMG data can be used to detect the amputee's desired gesture in prosthetic devices. Currently, the demand for prosthetic hands has continually increased to support hand amputees all around the world [4]. However, prosthetic hands are not affordable for most patients [6]. With the advancement of 3D printing technology, 3D-printed prosthetic hands are low-cost and customizable [98]. Therefore, a reliable system for the users to control the prosthetic hands is needed. 3D printed sEMG sensing systems show a promising performance for detecting grasping gestures in prosthetic hand [85] as it has several advantages over conventional methods [99]. Having a precise and user-customized sEMG sensor could reduce the post-processes after the signal measurement. Recently, AI-based data processing methods, including machine learning algorithms, have been widely used to detect the amputee’s grasping gesture from EMG data.

Nevertheless, these methods require lots of computational costs and time. 3D-printed electronics could provide a flexible and patient-specific solution, so fewer post-processing steps would be needed for gesture detection purposes. Ku et al. [100] reported a prosthetic hand utilizing the myoelectric interface for motion control, shown in Figure 2.5.c. The EMG electrodes were directly connected to the prosthetic hand. By detecting the voltage changes of the EMG signals through the sensing system, the prosthetic hand was controlled to the related motion based on the user's muscle contraction on the forearm. Cognolato et al. [101] evaluated a 3D-printed dexterous prosthetic hand for sophisticated gesture recognition in EMG control. The performance was tested by the Myo armband. After the calibration, the prosthetic hand was tested on real-time control, which minimized the movement time gap after the muscle contraction. As a result, the prosthetic hand could perform several gestures in various response times, such as fisting, grasping, and pinching. Depending on the complexity of the gesture, the motion performance time was varied with an average of two seconds. These studies have proven the feasibility of 3D printing in prosthesis applications. Therefore, the printed electrodes and prosthesis can create a fully customizable EMG sensing device, reducing the manufacturing time and being affordable to more users with the increased number of demands. Cantù et al. [102] proposed a prosthesis integrated with a printed EMG multi-electrodes matrix for rehabilitation purposes, shown in Figure 2.5.d. The aerosol jet printing technique fabricated the electrode matrix with conductive silver ink. The 3D printing allowed the direct integration of the EMG sensing system as it was printed on the prosthesis. The electrode matrix layout refers to the EMG detection locations of the commercial electrodes. The customized sensing system provided favorable feedback on the EMG signal acquisition during muscle contractions. Compared with the standard electrode, the printed electrode matrix showed similar results with only a slightly low amplitude of the time features. The advancement of 3D printing creates the capability of reducing the difficulty in the fabrication of the EMG-based prosthesis. The sensing system can be implanted into the prosthesis together as a single manufacturing process [103]. The customizable sensing system allows the designers to develop suitable devices of different shapes to fit the prosthesis based on its application. Although 3D-printed prosthesis has great potential, there are still disadvantages, such as the relatively slow printing speed, limited printing dimension, and low fabrication resolution.

3D printing has further enabled printed EMG sensing in the biomedical/clinical field [104]. The customized and flexible structures are suitable for tracing the irregular shape of the human body parts, providing the maximum amount of electrode-skin contact area to increase the sensor's precision. Ren et al. [105] developed a 3D-printed flexible microneedle electrode for multiple biomedical applications, such as drug delivery, EMG signal acquisition, and medical diagnosis, as shown in Figure 2.5.e. The microneedle electrode was fabricated using stereolithography (SLA) and digital light processing (DLP) 3D printing. The microneedle electrode is printed on relatively small scales, ranging from 25 to 100 micrometers. The microneedle electrodes were compared with conventional electrodes during the EMG test on the biceps brachii and elbow. The custom electrodes with 36 needles recorded the highest amplitude of the EMG signals from the muscle contractions. In addition, the number of needles can be increased to reduce the impedance of the electrode skin, which further improves the signal quality [83]. This iEMG sensing system can be applied for muscular dystrophy diagnosis in the clinical field [103].

Wearable health monitoring devices have recently been widely applied to discover potential diseases/disorders for clinical purposes [106]. Due to the complex process of bio-signal monitoring in clinical practice, conventional wet/dry electrodes will provide a poor signal recording in daily life [107]. Therefore, the printed flexible and stretchable electrodes can resolve the listed problems. The close skin contact with soft materials provides a comfortable user experience for wearable EMG monitoring [108]. Huang et al. [74] fabricated a flexible carbon electrode for EMG monitoring. The electrode demonstrated high conductivity and flexibility for long-term sensing applications. The electrode was integrated into a smart wearable device in the study. The sensing system measured the EMG signals from both leg and arm muscles and successfully distinguished the motion difference and strength level. The study proposed the stability of a printed wearable EMG system for health monitoring. In addition to daily monitoring, the wearable EMG can also be utilized in different situations. For example, vehicle drivers with unforeseen medical issues during driving can cause serious disasters and danger to them and others [109]. Therefore, Said et al. [110] reported a 3D-printed wearable bracelet to detect potential health risks. The bracelet consists of three electrodes located at the wrist and elbow. By feature extraction, the EMG signals can be divided into two groups, safe and dangerous signals. The system would consistently monitor the driver's EMG signals. When a hazardous signal was triggered, the bracelet would warn the driver to confirm his/her health condition. If the driver did not respond to the system, an emergency protocol would be activated to avoid a vehicle collision.

Aside from the prosthesis and clinical applications, the 3D printed sensing system has also been applied in robotics and motion sensing. Kim et al. [8] developed a sensing robot with specific 3D-printed origami-structured EMG electrodes placed on the robotic hand's fingertips, as shown in Figure 2.6.a. The robot was designed to provide medical assistance by touching the users to acquire EMG data. The soft robot hand structure provided a relatively safe interaction condition than most commercial rigid robot hands. Zhang et al. [111] proposed a 3D-printed eyeglass temple with implanted EMG sensing system to measure the signals generated from temporalis contraction, shown in Figure 2.6.b. The temple would trace the shape of the face to give close skin contact. This prototype has the potential to be applied for facial motion sensing and control in the smart glasses area.

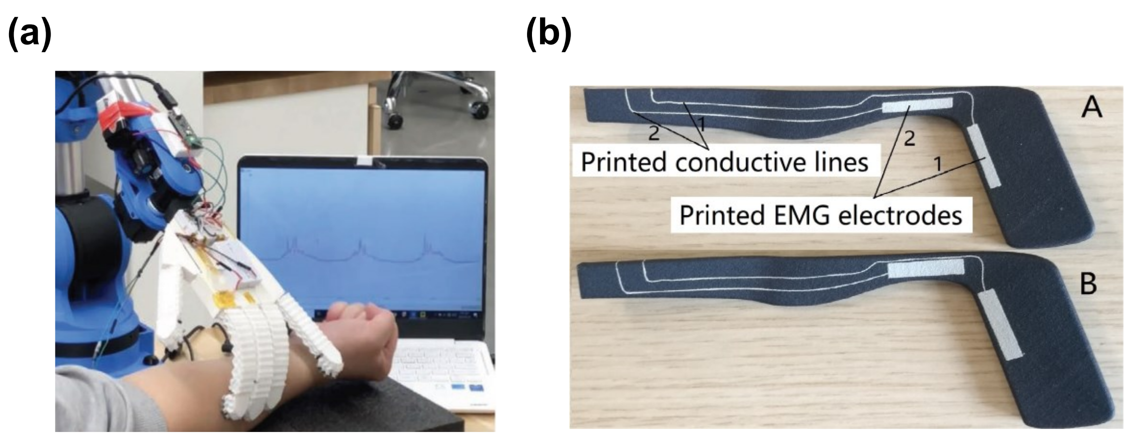

Figure 2.6. (a) Sensing robot hand measuring EMG signals on the forearm muscle with electrodes printed on the index, middle, and ringer fingers of the hand. Reproduced with permission.[8] Copyright 2021, John Wiley and Sons. (b) Printed eyeglass temple prototypes with printed conductive lines and electrodes for the EMG measurement on the head. Reproduced under the terms of the CC-BY license.[111] Copyright 2018, ACM.

Chapter 3.

Design & Fabrication of 3D-Printed EMG Wristband

3.1. Preparation of Surface Dry Electrodes

To detect the EMG signals generated in muscles from the residual arm, custom stretchable serpentine surface electrodes were created for signal acquisition. These custom-made electrodes are designed to be stretchable, which allows them to conform to the shape of the residual arm and maintain good contact with the skin, even as the arm moves. The custom-made electrodes are also designed to maintain their stable shape from bending and stretching [112]. This is crucial for accurate and reliable recordings, as any changes in the shape of the electrodes can affect the quality of the EMG signals.

3.1.1. Serpentine Electrode Design

The serpentine electrode is a type of surface dry electrode that is designed to provide a more accurate and complete recording of human muscle activities. The serpentine electrode allows for a wider area of muscle to be covered by the surface electrodes. As EMG signals can vary depending on the detection location of the electrode, the serpentine electrode creates a more comprehensive EMG detection with a larger electrode coverage area on the muscle. The serpentine electrode is also a suitable solution for uneven or curved skin surfaces, which can be difficult to access with conventional electrodes. The serpentine structure can wrap around the surface and make close contact with the muscle. In addition, the customizable serpentine electrode can be adjusted to different sizes for fitting with various muscle sizes as well. Serpentine electrodes have several advantages over conventional electrodes, such as:

1. Increased coverage: Serpentine electrodes can cover a larger area.

2. Reduced motion artifact: The shape of serpentine electrodes allows for more flexibility to the skin, which can reduce the motion artifact caused by the movement between electrodes and skin.

3. Greater sensitivity: Serpentine electrodes are generally more sensitive than conventional electrodes, which can detect lower EMG signals from muscles.

4. Improved signal-to-noise ratio: Serpentine electrodes can detect more accurate signals over conventional electrodes

5. Simple access: Serpentine electrodes do not require specialized equipment to be used for applications.

In this research study, the flexible serpentine electrode was designed shown in Figure 3.1. It contains three rows of serpentine-patterned structures and embedded conductive wires to create stable conductivity and high sensitivity in signal detection on the muscles.

Figure 3.1. 3D printing of the serpentine electrode.

3.1.2. DIW Printing of Serpentine Electrodes

The serpentine electrodes were fabricated by the DIW printing method using the Musashi printer [113], shown in Figure 3.2. Conductive flexible silver ink (Flex 2 Ink, Voltera) was used for the electrodes, and the substrates were made from platinum-catalyzed silicones (EcoFlex). The fabrication process is shown in Figure 3.3. Prior to the printing, conductive wires were embedded into the substrate to create a steady connection with the EMG sensor.

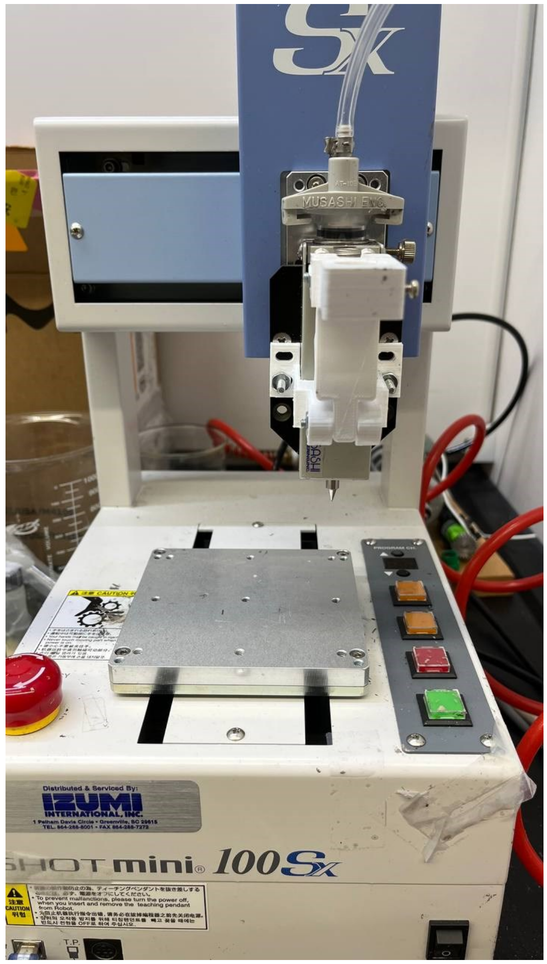

Figure 3.2. Overview of the Musashi printer for DIW printing.

The fabrication process was accomplished through several steps. First, the flexible ink was transferred into the syringe on the Musashi printer. A 0.3mm nozzle was used for the ink extrusion. Prior to printing, the silicon substrate was adjusted to be parallel to the printing platform. Then, the substrate was fixed to prevent an unintended dislocation. The serpentine pattern was then printed on the substrate. The printing speed was set to 0.5mm/s, and the pressure of the compressed air was at 110kPa. After the DIW printing, the printed electrode sample was placed on a heater and cured for 15 minutes at 110 ºC. Once the sample was fully cured, a thin layer of the EcoFlex was added on top of the sample to protect the printed electrode. Finally, the sample was cured again for 15 minutes at 110 ºC. Steps 2-5 were then repeated for each serpentine electrode fabrication. The complete set of serpentine electrodes is shown in Figure 3.4.

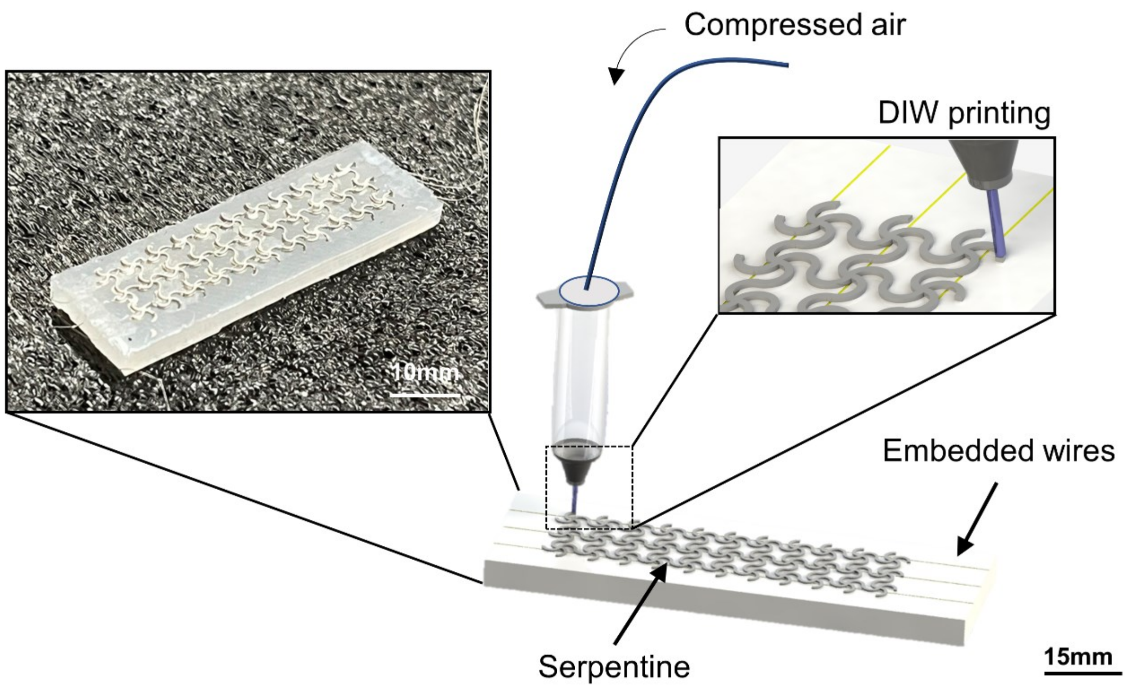

Figure 3.3. 3D printing of the serpentine electrode.

Figure 3.4. Five electrodes for the detection of EMG from five fingers.

3.1.3. Characterization of Serpentine Electrodes

The surface dry electrode consists of 3 rows of serpentine structures as demonstrated in Figure 3.5. This structure can provide maximum bending and stretching without any fracture [114]. Two layers of conductive ink were embedded to further reinforce the connection of the serpentine pattern. The structure of a continuous serpentine shape provided stretchability against deformation. The double-layer printed electrodes had a 100% strain limit of 35 mm, which was sufficient for the reliable performance of the wristband. Our previous study validated that the number of serpentine layers would directly affect both conductivity and stretchability on the electrode. The thicker layers showed improved reliability of conductivity during the stretching, which required higher bending force to reach the failure state [115]. The EcoFlex substrates further strengthened the electrodes for resistance against deformation during arm movement. Both bending and stretching tests were performed to validate the connection stability. In the bending test, the length of the serpentine set was shortened from 40mm to 20mm, and the impedance dropped from 0.72Ω to 0.60Ω.

During the stretching, the length was stretched up to 100% more than its original length. At 50% of the stretching, the impedance is maintained within the range of 1.0 ± 0.2Ω. When the length was increased by 60%, the serpentine pattern started to break. Once the sample reached 100% of its original length, the impedance increased up to 2.5Ω, as shown in Figure 3.6. When the surface dry electrode was inserted into the actual wristband device for practical purposes and testing, its limit was to be stretched by about 20% of its original length, which was significantly lower than the maximum capability of 60%. Therefore, the characteristics of the serpentine pattern allowed the EMG electrodes to become stretchable and bendable while maintaining steady functionality on the EMG signal detection on the forearm. The surface electrodes printed using the serpentine pattern were able to resist the damage and deformation from the arm movement, such as arm twisting and skin rubbing.

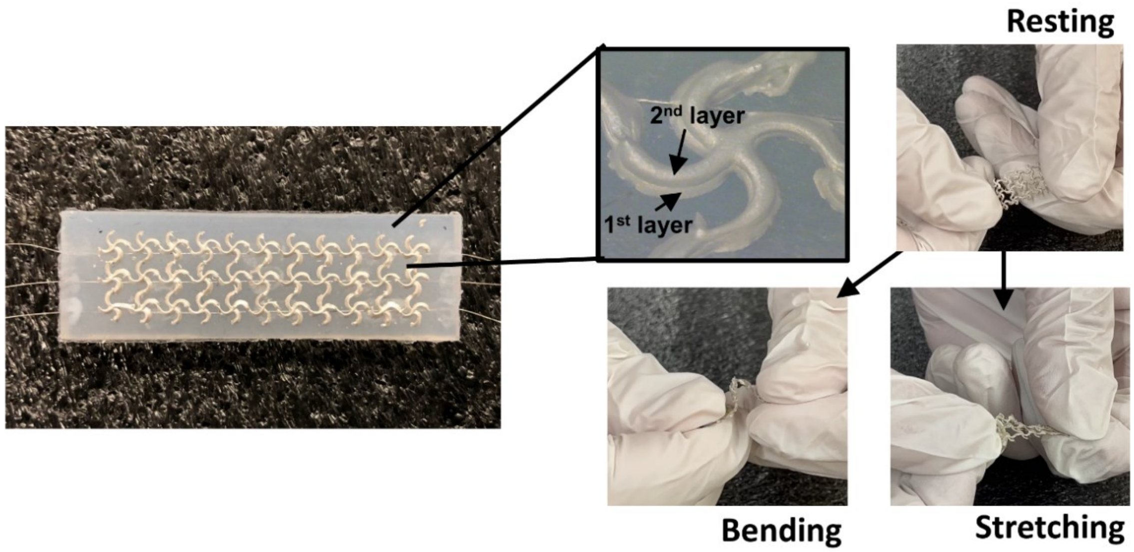

Figure 3.5. Left: Cross-serpentine pattern on the dry electrode sample. Right: Sample image of double-layered conductor and demonstration of stretching samples.

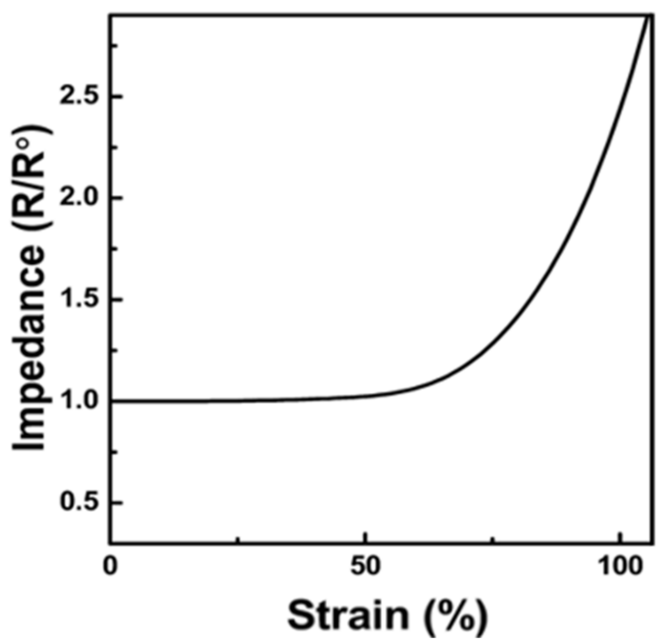

Figure 3.6. Impedance change as a function of strain.

3.2. 3D-Printed Wristband

Using 3D printing technology, it is possible to fabricate a wristband prototype as a cost-effective device for EMG detection. 3D printing allows for the creation of customized designs that can be tailored to the specific needs of the application. In this case, 3D printing was used to create a wearable wristband that incorporates serpentine electrodes and other necessary components.

3.2.1. Design of Wristband

The wearable wristband is designed to acquire and process EMG signals simultaneously, as shown in Figure 3.7. It is intended to be worn on the forearm and equipped with five pairs of electrodes to collect EMG signals from all finger muscle movements. The electrode pairs have independent positive and negative electrodes while sharing one common reference electrode. The direction of the electrode pairs is aligned with the flexor pollicis longus for the thumb and the flexor digitorum profundus for all other fingers. The wristband design is to ensure accurate and reliable recordings of muscle activity. The wristband holder contains the EMG signal processing system which is connected to all electrodes through internal wirings. This allows for the collection, amplification, and processing of the EMG signals in real time. The signals can then be transmitted to a computer or other devices for further analysis. All parts of the device, including the wristband holder and electrodes, were fabricated using 3D printing technology with Thermoplastic Polyurethane (TPU) material. TPU is a flexible, durable, and biocompatible material that is well-suited for use in wearable devices, making it an excellent choice for a wristband prototype.

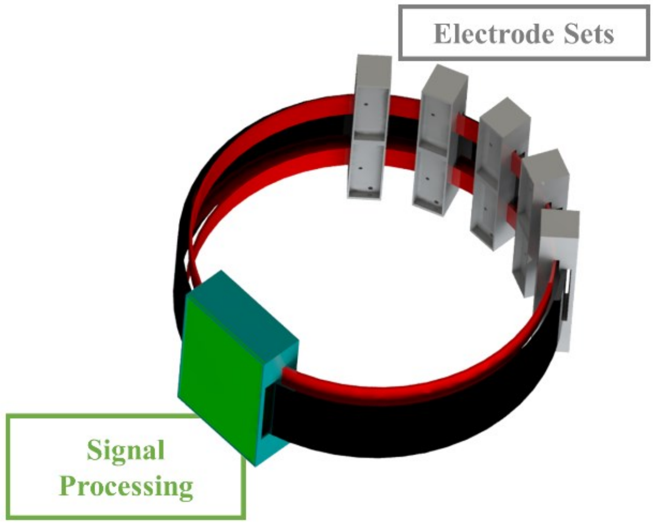

Figure 3.7. Overview of the wristband system.

3.2.2. FDM Printing of Wristband

FDM printing is an efficient and cost-effective method for fabricating a TPU-based wristband prototype. This also allows for the creation of customized designs for different users as the size of the wristband can be adjusted accordingly. FDM printing offers a stable printing process when using flexible TPU as the printing material. The printed wristband is flexible that traces the shape of the user's forearm, ensuring that the attached electrodes are kept in close contact with the skin. This is cruel for accurate and reliable EMG signal detection. It is also durable with repeated bending and stretching while maintaining its original shape. This makes it an excellent choice for wristbands that will be worn for extended periods of time. The TPU material is also a safe material for contact with skin, which is suitable for the wearable wristband prototype. In addition, the wristband is adjusted in terms of the location of the electrodes. As each user has a different arm size, the wristband can still be calibrated to be suitable for everyone. Figure 3.8 demonstrates the printed wristband worn on the user's forearm.

Figure 3.8. Printed wristband on the forearm.

3.3. Signal Processing System

3.3.1. Structure of Signal Processing System

As a portable sensing system, a double-sided Printed Circuit Board (PCB) was chosen to fit into the wristband holder with its small size. The use of a double-sided PCB also allows for increased circuit density and improved trace routing, which are important factors for a portable sensing system. Additionally, the use of a 3D Direct Ink Writing system (V-one printer) for the fabrication of the PCB using conductive silver ink, allows for a cost-effective and eco-friendly solution. The PCB signal processing system is also designed to be compact and efficient, consisting of a single EMG sensor, a microcontroller unit, and multiplexers, instead of multiple independent EMG sensors. The PCB was fabricated using the conductive silver ink by 3D DIW systems (V-one printer, Voltera, Canada) [116]. The PCB signal processing system consisted of one EMG sensor (Myoware muscle sensor, Sparkfun, USA), a microcontroller unit (Arduino Nano 33 IoT), and two multiplexers (TC74HC4051AP, Toshiba, Japan). The use of multiplexing in this portable sensing system allows for a more compact and efficient design, as it eliminates the need for multiple independent EMG sensors. Instead, all EMG signals are collected and sent to a single sensor, which contains an amplifier, bandpass filter, and rectifier. This sensor can then process the signals from all five electrode groups in a cyclical manner, using the multiplexer. This not only reduces the size and complexity of the system but also the cost of the system. Additionally, it also makes the system more reliable and easier to maintain. The multiplexer essentially allowed five signal inputs to be processed in a cycle from the analog to digital converter (ADC), shown in Figure 3.9.

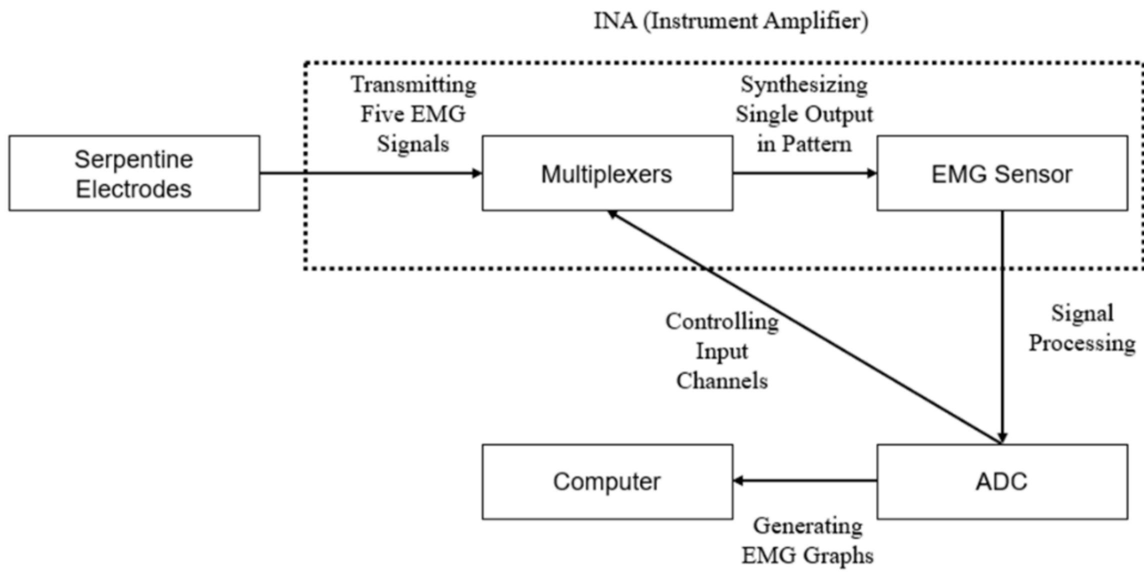

Figure 3.9. Operation flowchart of the wristband system.

3.3.2. Design of Printed PCB

The double-sided PCB was selected as an ideal component of the portable sensing system. The circuit schematic diagram was designed using Eagle software by Autodesk, shown in Figure 3.10. One multiplexer is linked to all the positive electrodes, and the other one is linked to all the negative electrodes. The outputs of both multiplexers were connected to the positive and negative channels of the EMG sensor. Since all the finger muscles shared one common reference electrode, the electrode was connected to the reference channel of the sensor. This approach has reduced the complexity of the system compared with the traditional multi-sensor system. It is also more reliable, as it reduces the number of components and the risk of failure in the system. In order to achieve real-time EMG data, it is crucial to minimize the delay between the time the EMG signals are collected and the time they are processed. By reducing the delay time of the multiplexer input switch, the multiplexers allow the EMG signals to be processed at a speed that is as close to simultaneous as possible. The use of multiplexers with low delay times allows for a more efficient system, as it reduces the time it takes for the EMG signals to be collected, processed, and analyzed. This is particularly important for real-time monitoring and analysis of muscle activity, as it allows for quick and accurate data collection and processing. The EMG signals generated from the finger muscles were detected by the electrodes and sent to the PCB as the raw EMG signals. The first step in processing the raw EMG signals is to remove noise by using low and high bandpass filters from the EMG sensor. These filters are used to eliminate unwanted noise and interference from the signals, which can be caused by various factors such as electromagnetic interference or movement artifacts. The low and high bandpass filters remove frequencies that fall outside of the determined range of the biological signals, leaving only the relevant signals for analysis. After the noise is removed, the data is processed through the ADC to convert the signals from analog to digital format. This allows for the data to be represented as EMG graphs on the computer, which can be analyzed to understand muscle activity.

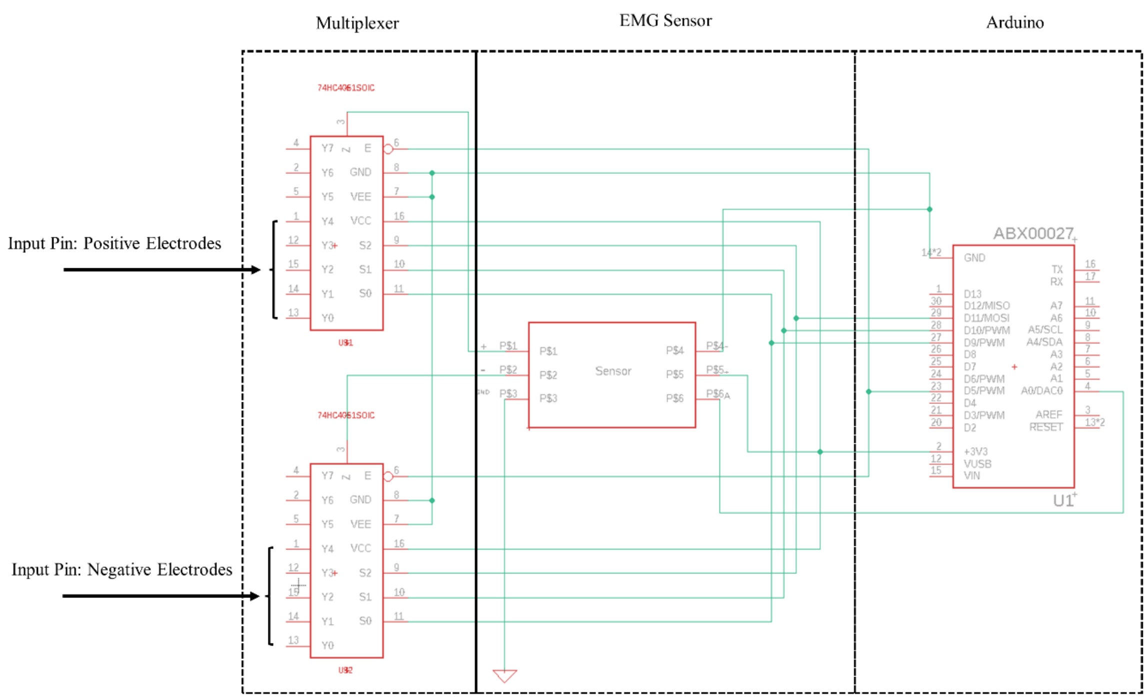

Figure 3.10. Schematic of PCB circuit design using Autodesk Eagle.

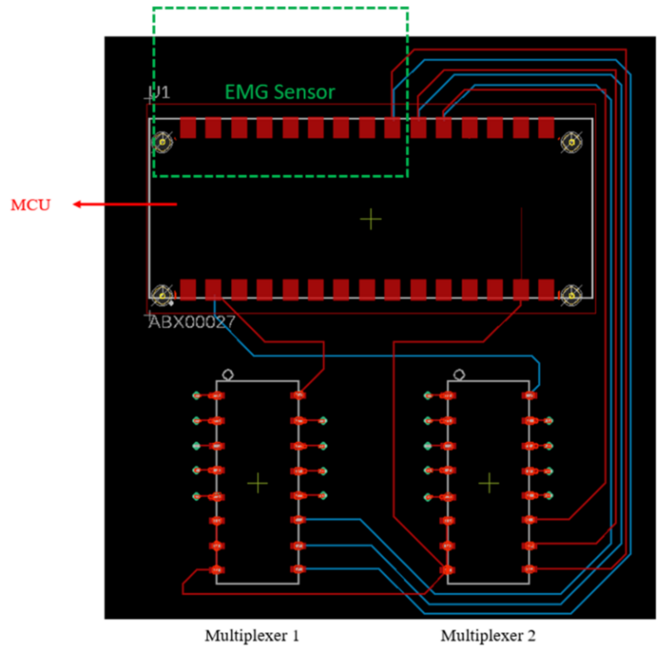

Figure 3.11. The board layout of double-sided PCB (red: top, blue: bottom).

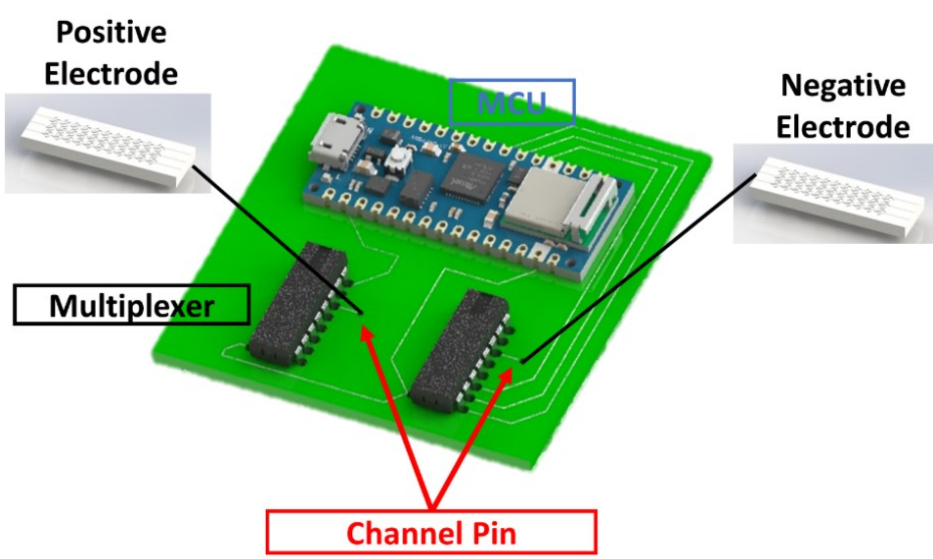

Figure 3.12. Assembly of the PCB.

The board layout of the double-sided PCB is shown in Figure 3.11. The Arduino Nano 33 IoT and two multiplexers were designed to be placed on the PCB top layer while the EMG sensor is attached to the bottom layer. The 50mm * 50mm board would connect all the electrodes to their corresponding multiplexer pins. As an example, shown in Figure 3.12, the positive and negative electrodes would connect to one of the pins on each multiplexer.

3.3.3. Fabrication of Printed PCB

The V-one printer from Voltera is chosen for the PCB printing, shown in Figure 3.13. The printer has a build area of 150mm * 100mm, which is designed for small-scale prototyping. It uses a direct inkjet printing process to extrude conductive materials onto a PCB substrate layer. The building area also includes an oven for the post-printing process, which allows the PCB to be cured shortly after printing. The lead-screw extrusion head drives the movement of the filament for material extrusion. By using a 0.2mm diameter nozzle, the circuit trace can be printed precisely to avoid overlapping. In addition, the small diameter nozzle also minimizes material waste and reduces the printing time. Commercial flexible conductive ink was used as the printing material. Its high conductivity and durability ensure the printed circuits can have a longer lifespan.

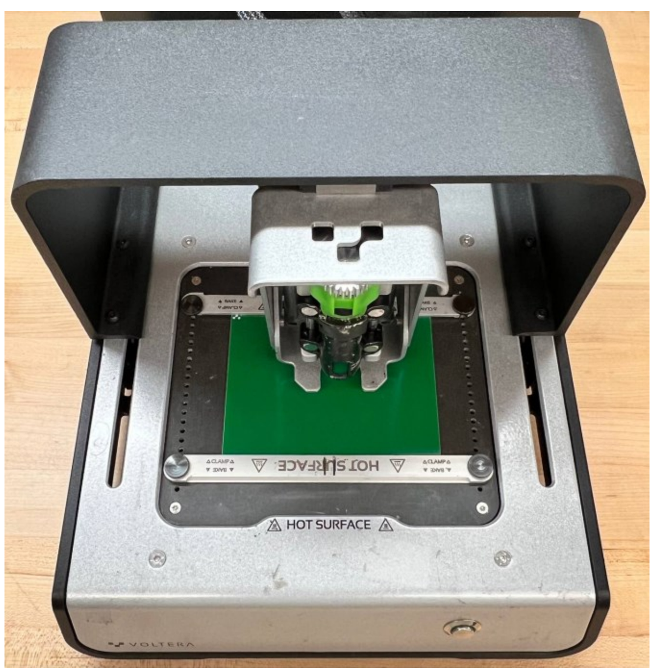

Figure 3.13. Voltera V-one DIW printer.

Before printing, the conductive ink was loaded into the lead-screw extrusion head, and the PCB substrate was placed on the build plate with two clamps to ensure its fixed position. In order to fabricate the double-sided PCB created from Eagle, the design was converted into a Gerber file (GTL) and drill file (TXT). After calibration, the V-one printer would receive the GTL and TXT files to begin the printing process. The printer would print the top layer and drill holes first. The drilling holes were necessary as they were to ensure the correct alignment between the top and bottom side circuits. Once the top layer is printed, the PCB substrate is flipped over and placed on the clamps for curing. This process would solidify the ink and create functional circuits. After inspecting the printed circuits for proper connection, the PCB substrate was placed on the build area again for bottom-layer printing. The alignment was calibrated to have the drilled vias aligned with the top layer. Then, by the same printing procedure, the bottom layer was fabricated and cured as well. The connection for the double-sided PCB was tested and no defects were found. Next, the Arduino Nano 33 IoT and two multiplexers were mounted onto the designated locations on the PCB, shown in Figure 3.14. The serpentine electrodes and EMG sensor were connected to the PCB with conductive thin copper wires. In the end, the PCB signal-processing system was placed into the wristband holder.

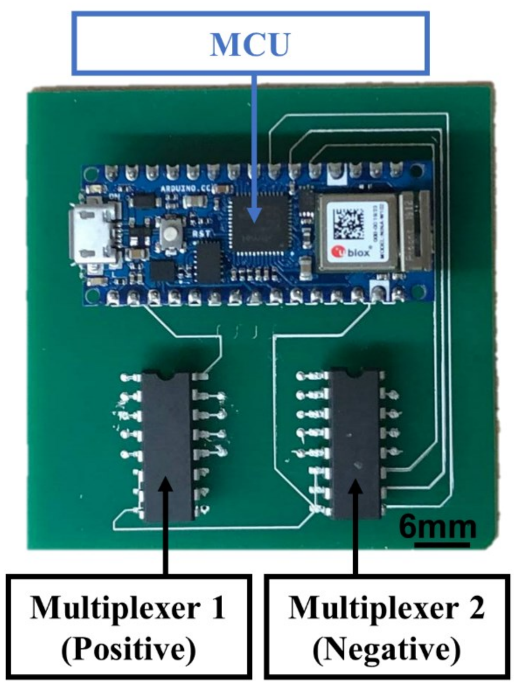

Figure 3.14. Printed PCB control unit.

Chapter 4.

Result & Discussion

4.1. Research Ethics Approval

All procedures to perform the research study of the 3D printed wristband have been approved by the research ethics committee at Simon Fraser University in Canada: Minimal Risk Approval 30001017. The participant signed the approved consent forms before participating in the tests. According to the legal term of Canada, the confidentiality of participants is strictly respected.

4.2. Experimental Setup

The volunteer completed the testing on the same day for data consistency. First, the muscles which were corresponding to all five fingers were located for EMG signal detection, shown in Figure 4.1. The locations with visible changes in EMG signal intensity were selected for the placements. Further verification was performed during the experiment to ensure the electrodes were acquiring signals from their corresponding finger. The wristband was then positioned on the volunteer's forearm-marked area, shown in Figure 4.2. Five pairs of dry electrodes were adjusted accordingly to detect EMG signals generated during the muscle movements of all fingers. Secondly, the volunteer sat with the forearm positioned horizontally on the flat surface table while holding a hydraulic hand dynamometer (Jamar, Performance Health, Canada). While performing the hand-gripping process, both the wristband and dynamometer recorded the EMG signal and applied force simultaneously.

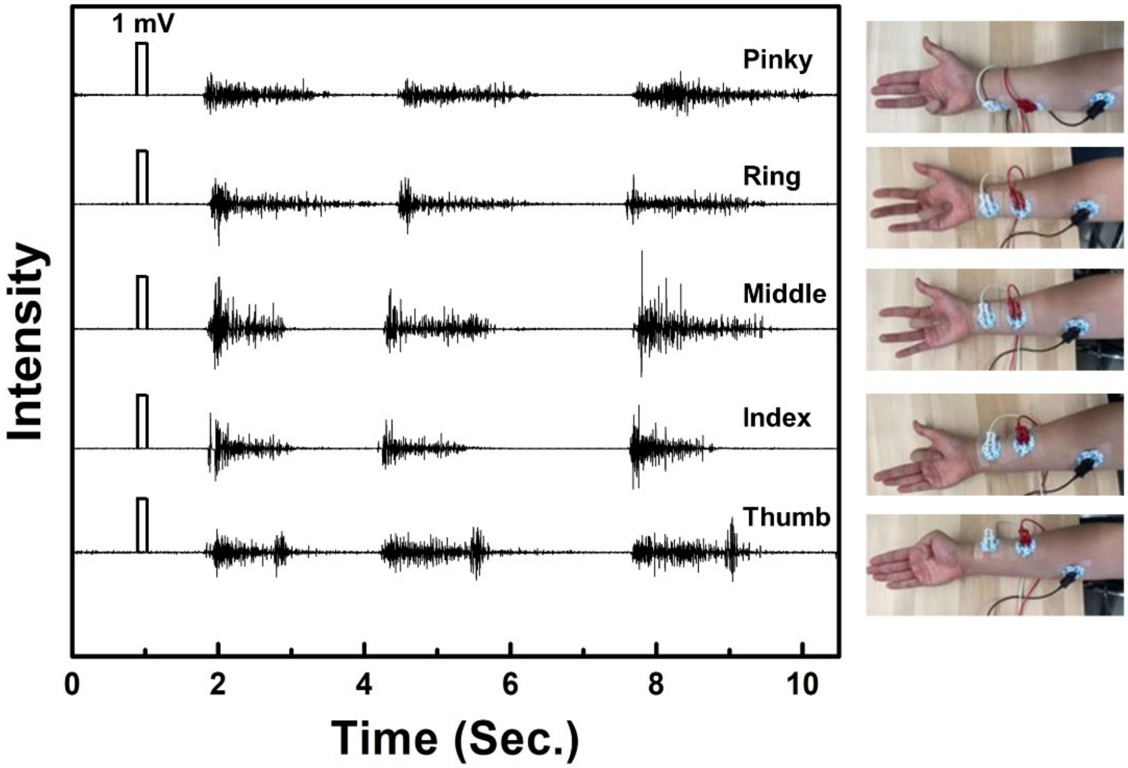

Figure 4.1. **EMG signal detection at different muscle locations during finger movement.**

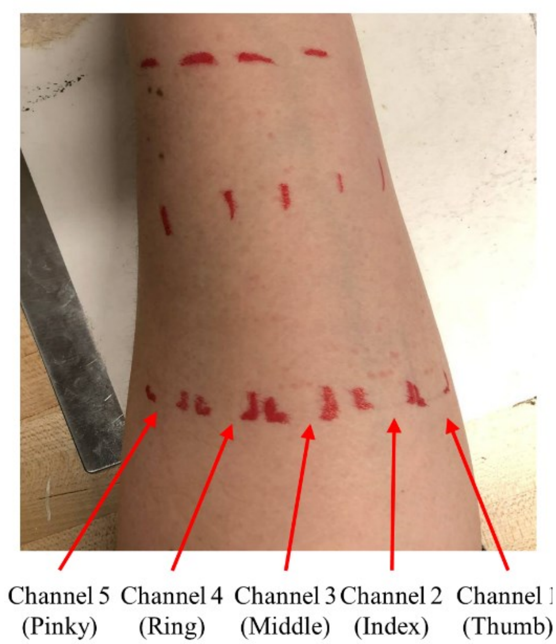

Figure 4.2. **Five selected locations for the EMG wristband electrode placements.**

4.3. Locations of EMG Signal Detection

Due to the difference between human arm sizes, the wristband placement was verified on the volunteer first. The EMG signals for five fingers were detected individually along the flexor pollicis longus and flexor digitorum profundus on the volunteer's forearm. Since the EMG electrodes were only placed on the location of one specific finger, the EMG signals received by the EMG sensor were mainly generated from this finger movement. As an example in Figure 4.3, the EMG electrodes were placed on the middle finger detection location on the forearm. The volunteer performed the finger flexion and extension movement three times on each finger. As the result, the middle finger created the highest intensity of the EMG signal during flexion and extension. Because of the EMG electrode placement, other fingers showed significantly lower signals from the same detection spot. Since all muscles were connected, it was inevitable to receive the EMG signals from a different finger movement. Compared with other devices, our wristband is targeted to detect the precise motion of each finger. For instance, the Myo armband is focused on the five standard gesture interactions by default [117]. Thus, the hand motion and functions are limited to performing assigned controls. However, the independent finger signal detection in our wristband provides the ability to carry out many more gestures selectively from the user.

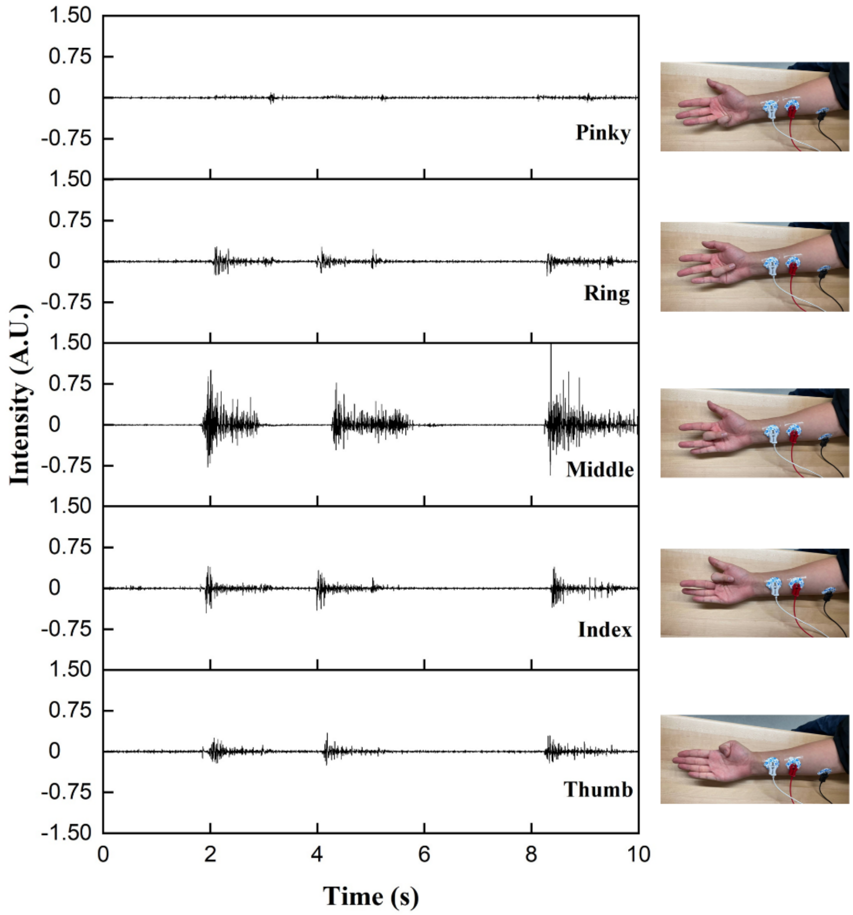

Figure 4.3. **Muscle-specific EMG detection for the middle finger.**

4.4. EMG Signal Monitoring by the Wristband

Once the EMG detection locations were selected, the wristband was equipped and adjusted to fit on the volunteer's forearm. The volunteer performed a series of tests to confirm the EMG signals from specific finger movements as shown in Figure 4.4. Initially, five fingers were flexed with all channels connected by the multiplexers. Secondly, each channel was then opened to verify the EMG signals from individual finger movements. At last, the wristband detected pinching movement with two different fingers while all channels remained active. The multiplexers classified the EMG signals from different channels, which allowed the single EMG sensor to process the data generated at all five signals simultaneously.

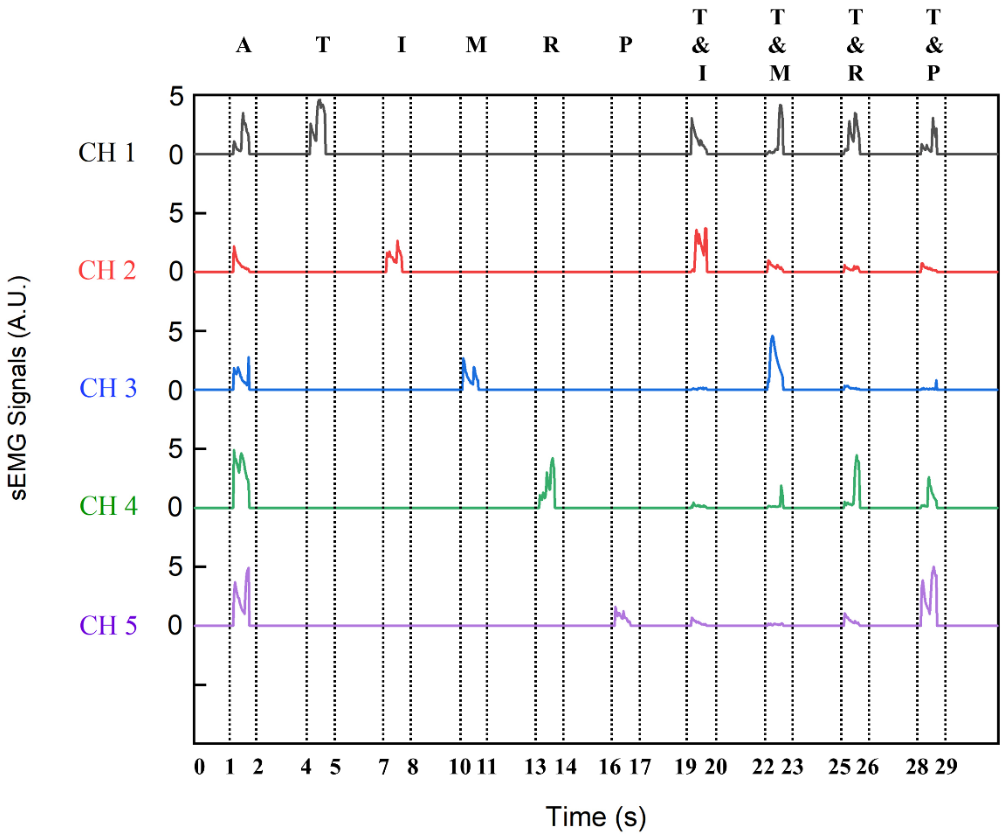

Figure 4.4. Confirmed specific location-based EMG signals (A: all fingers; T: thumb; I: index; M: middle; R: ring; P: pinky).

4.5. Quantitative Analysis of EMG Signal

The correlation between the EMG signal intensity and force exerted by forearm muscles was studied. The volunteer was asked to wear the wristband and hold the hydraulic dynamometer on the hand. The test setup was illustrated in Figure 4.5. During the test, the volunteer applied five different levels of force from low to high: 15 lb, 30 lb, 45 lb, 60 lb, and 75 lb. From the dynamometer, the gripping forces were displayed and recorded. When the force reached the target value, the volunteer released the hand with 2-4 seconds of rest before the next gripping. The EMG signals and applied forces were recorded concurrently. Quantitative analysis was demonstrated in Figure 4.6. The EMG values were presented in percentage of the maximum muscle contraction force, which was 3.55 mV at 100%. Since the gripping force was controlled by the hand, the volunteer had to adjust the force value manually during the test. Therefore, the EMG signals showed two different peaks at lower force levels due to the rapid force adjustment. The signal-to-noise ratio (SNR) was monitored as:

$$SNR_{dB} = 20log_{10}\left(\frac{A_{signal}}{A_{noise}}\right),$$

where A_{signal} is the signal power, and A_{noise} is the noise power [118]. The Myo armband shows a SNR of 7.6622dB by filtering of signals [119]. The wristband's filter ranges between 20-500 Hz, which is similar to the gForcePro+ EMG armband. This provides consistent and stable signal readings for the EMG application. To determine the SNR of the wristband, the collected result from Fig 4.6 was used as a reference. The noise power was established at 0.15 mV throughout the analysis. The peak signal power of 3.55 mV was applied to the calculation. Thus, the SNR of the wristband was at 27dB, which indicated the satisfactory performance of the EMG sensing. The signal sensitivity was evaluated as:

$$Sensitivity = \frac{\Delta V}{\Delta F},$$

where ΔV is the signal amplitude difference, and ΔF represents the exerted force of the muscle contraction. By the linear approximation, the signal sensitivity was calculated as 0.556 mV/lb through the test. The results had a coefficient of determination value of 0.97, which showed that the model fit the observed data.

Figure 4.5. Gripping test setup.

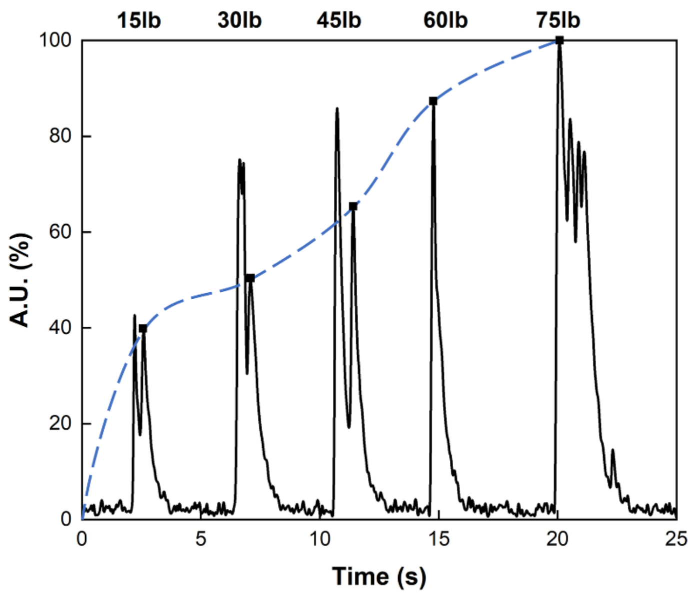

Figure 4.6. The EMG signal intensity at different levels of muscle force.

The test demonstrated a non-linear relationship between EMG and force. While the muscle force increased each time from the gripping movement, the EMG signal peak amplitude continued to rise accordingly. As a result, the EMG signals collected from the wristband could represent the muscle force of its detected area quantitatively based on the signal intensity.

Chapter 5.

Conclusions and Future Work

5.1. Conclusions

The reliability of 3D printing technology in the fabrication of EMG devices has been demonstrated in this study. The stretchable EMG dry electrodes offer a more comfortable experience for users during EMG detection. The flexibility of the electrodes enables them to conformally attach to uneven skin surfaces and improve skin adhesion. 3D printing also allows for the customization of the EMG system, which can be modified to improve its suitability for different EMG acquisition designs. The versatility of 3D printing has enabled the adaptation of EMG sensors to various fields of practical application, including prosthetics and health monitoring. With 3D printing, sensing systems can be fully integrated for long-term use. The entire sensing system can be fabricated with a single process using plastic and conductive 3D printing technologies, reducing the dimensions of the devices compared to traditional processes.

In this research study, the primary milestone is to design and demonstrate a customizable EMG wristband that could be fabricated using 3D printing technology. These are summary of the accomplished results in this thesis.

1. Implementation of custom stretchable serpentine electrodes

The serpentine structure provides high flexibility, which is essential for proper contact with the skin on the user's forearm. It is also stretchable to maintain a consistent contact for accurate signal recording during muscle movements. In addition, the high conductivity ensures that the signals can be detected with a high degree of sensitivity.

2. Development of the signal processing system

Compared with the traditional EMG signal processing systems that require multiple sensors, our design is able to capture and process signals from multiple locations using only a single sensor. This significantly reduces the number of sensors to be inserted to the wristband device, which also reduces the cost and complexity of the fabrication.

3. Design and fabrication of the 3D printed EMG wristband

The wristband is designed to contain the signal acquisition system and all electrodes all in one device. The wristband is fabricated using FDM 3D-printing technology. Therefore, it can be customizable to various sizes for different users. The TPU printing material also assists in creating a close contact between the skin and serpentine electrodes.

4. Demonstration and validation of the prototype

To demonstrate the performance of the 3D printed EMG wristband, several tests were conducted. The device showed its capabilities in detecting multiple signals quantitatively, which can be further applied to the field of prosthetics and assistive technology.

The proposed wristband can fully contact a skin surface with the 3D printed dry electrodes, which increases the signal quality compared with standard dry electrodes of other wearable devices. The wristband also collects inputs from 5-channels to individually record the EMG signals generated from five fingers' motion. The demonstrated single-sensor system utilizing multiplexers reduces the number of components, which decreases the difficulty and cost of component replacement. Thus, the wristband was achieved as an accessible device. The user could also generate different intensity levels of the EMG, which can be classified into multiple finger movement commands. The wristband has demonstrated the stability of the EMG signal processing through quantitative analysis. The developed signal acquisition system was able to process five independent sets of data simultaneously. Furthermore, the result showed that the muscle contractions were reflected as different levels of signal intensity through the wristband. The stretchable dry electrodes further reinforced the surface EMG detection to acquire reliable signals. Furthermore, through data analysis, we determined that the amplitude of the EMG signal was related to the intensity of the force exerted by the muscle contractions. At the muscle contraction of 75 lb, the result was shown that the wristband possessed a signal sensitivity of 0.556 mV/lb and 27 dB SNR. The wristband structure was also able to be customizable for wearing with no additional components or add-ins using 3D printing technology. As the first prototype, the wristband is demonstrated under steady-state or mild motion to minimize the motion artifacts, which may affect the collected data during the experiment [120].

Overall, the developed EMG wristband device has demonstrated its ability to utilize EMG signals from the forearm muscles, which will enable different applications controlled by EMG signals in supporting amputees in their daily life. This research study creates new opportunities for more advanced assistive technology, such as prosthetic devices.

5.2. Future Work

This study presented a 3D-printed EMG wristband, which offers a new perspective on the development of EMG control and monitoring devices. 3D printing technology presents a high potential for future EMG devices as an affordable approach to manufacturing. Further studies on frequent motion will expand the wristband's application, making it capable of interfacing with different types of hand prostheses. The wristband can detect EMG signals at various intensity levels, reflecting different bending degrees of the individual finger, such as fisting and pinching. Unlike existing devices, which focus on pattern recognition, the wristband aims to improve the precision of finger motions. The single-sensor system allows for independent control of finger movement to accurately mimic intended movement by the hand prosthesis. The control system can optimize the performance of current prosthetic hands with the user's directed control. Additionally, the wristband can be utilized with a 3D-printed hand prosthesis as a completely customizable device. Finally, the prototype design can be improved by reducing the size of the surface electrodes and the wristband device based on the further study of EMG detection area of the finger motion.

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Appendix A.

Serpentine Electrode Flexibility

Demonstration of the serpentine structure under bending and stretching condition.

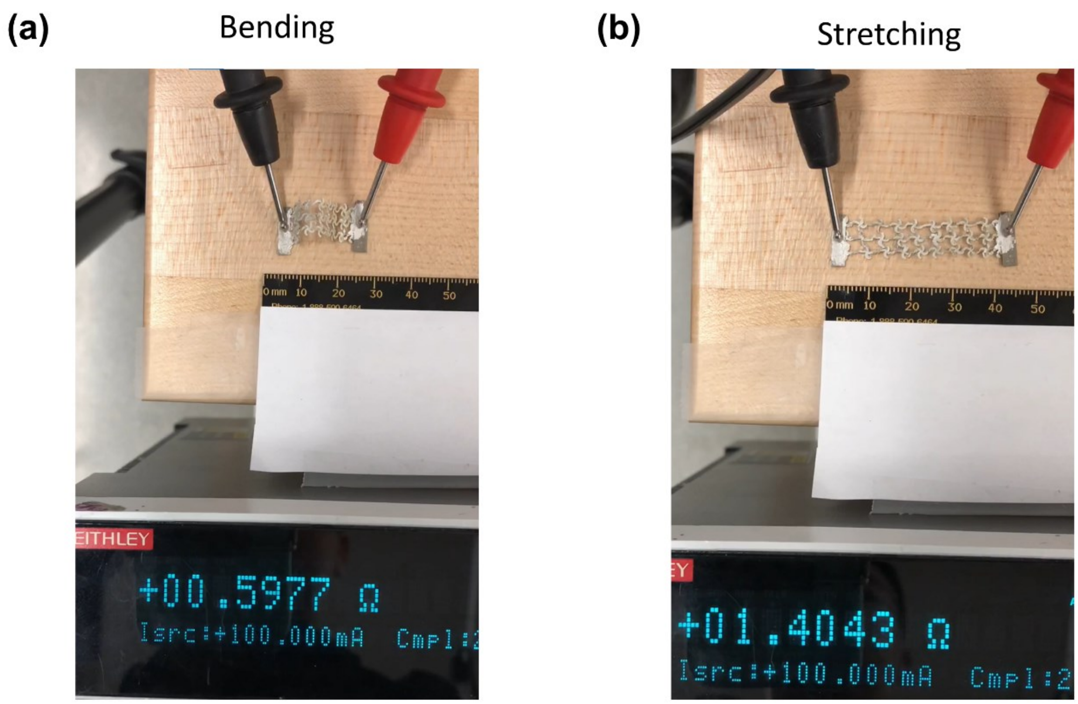
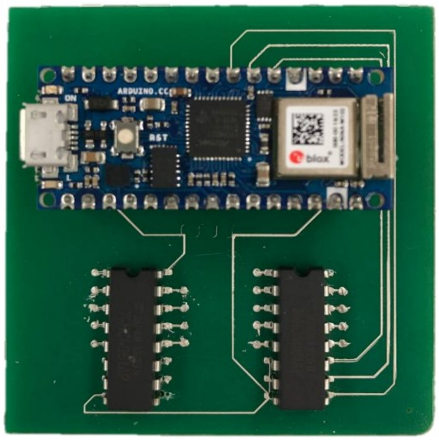


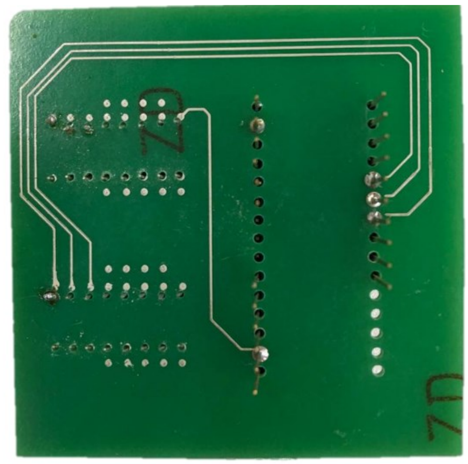
Figure A.1. Serpentine strucutre (a) bending; (b) stretching.

Appendix B.

Printed PCB



Front View



Back View

Figure B.1. Printed double-sided PCB.

Appendix C.

Multiplexer & Truth Table

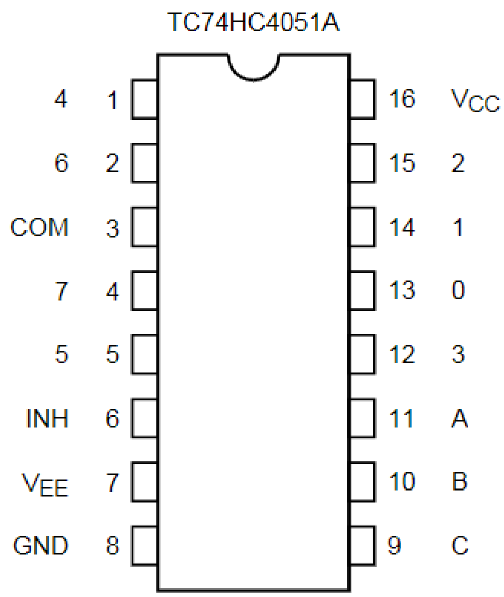


Figure C.1. TC74HC451A multiplexer.

Table C.1. Truth Table

A	B	C	Channel
L	L	L	0
L	L	H	1
L	H	L	2
L	H	H	3
H	L	L	4
H	L	H	5
H	H	L	6
H	H	H	7

Appendix D.

Circuit Workflow of the EMG Sensor (Myoware Muscle Sensor)

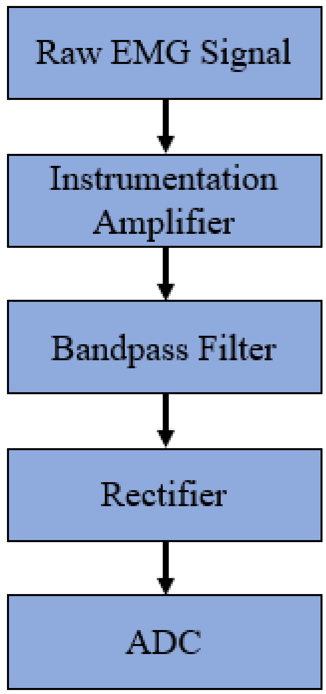


Figure D.1. EMG sensor circuit workflow.

Appendix E.

Calculation of the Signal Sensitivity on the Wristband

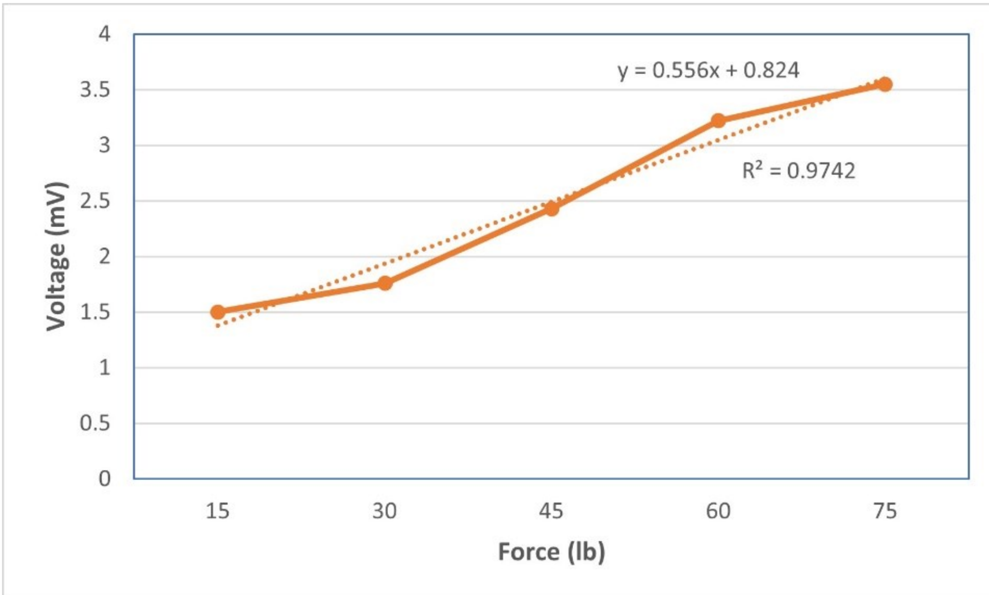


Figure E.1. Signal sensitivity calculation.

Appendix F.

Microcontroller Code

```
int A = 2, B = 3, C = 4;

int electrode0, electrode1, electrode 2, electrode 4, electrode 6;

unsigned long time_previous, time_current;

void setup() {

pinMode(A, OUTPUT);

pinMode(B, OUTPUT);

pinMode(C, OUTPUT);

time_previous = millis();

Serial.begin(9600);

}

void loop() {

 time_current = millis();

 if (time_current - time_previous >= 500){

 time_previous = time_current;}

 else if (time_current - time_previous >= 400){

 digitalWrite(C, LOW);

 digitalWrite(B, HIGH);

 digitalWrite(A, LOW);
```

```
int voltage = analogRead(A0);

float electrode2 = (voltage * (5.0 / 1023.0) + 50);

electrode4 = (40);

electrode0 = (30);

electrode6 = (20);

electrode1 = (10);

Serial.print(electrode2);

Serial.print(" ");

Serial.print(electrode4);

Serial.print(" ");

Serial.print(electrode0);

Serial.print(" ");

Serial.print(electrode6);

Serial.print(" ");

Serial.println(electrode1);

}

else if (time_current - time_previous >= 300){

digitalWrite(C, LOW);

digitalWrite(B, LOW);

digitalWrite(A, HIGH);

int voltage = analogRead(A0);
```

```
float electrode4 = (voltage * (5.0 / 1023.0)+ 40);

electrode2 = (50);

electrode0 = (30);

electrode6 = (20);

electrode1 = (10);

Serial.print(electrode2);

Serial.print(" ");

Serial.print(electrode4);

Serial.print(" ");

Serial.print(electrode0);

Serial.print(" ");

Serial.print(electrode6);

Serial.print(" ");

Serial.println(electrode1);

}

else if (time_current - time_previous >= 200){

digitalWrite(C, LOW);

digitalWrite(B, LOW);

digitalWrite(A, LOW);

int voltage = analogRead(A0);

float electrode0 = (voltage * (5.0 / 1023.0)+ 30);
```

```
electrode2 = (50);

electrode4 = (40);

electrode6 = (20);

electrode1 = (10);

Serial.print(electrode2);

Serial.print(" ");

Serial.print(electrode4);

Serial.print(" ");

Serial.print(electrode0);

Serial.print(" ");

Serial.print(electrode6);

Serial.print(" ");

Serial.println(electrode1);

}

else if (time_current - time_previous >= 100){

digitalWrite(C, LOW);

digitalWrite(B, HIGH);

digitalWrite(A, HIGH);

int voltage = analogRead(A0);

float electrode6 = (voltage * (5.0 / 1023.0) + 20);

electrode2 = (50);
```

```
electrode4 = (40);

electrode0 = (30);

electrode1 = (10);

Serial.print(electrode2);

Serial.print(" ");

Serial.print(electrode4);

Serial.print(" ");

Serial.print(electrode0);

Serial.print(" ");

Serial.print(electrode6);

Serial.print(" ");

Serial.println(electrode1);

}

else if (time_current - time_previous >= 0){

digitalWrite(C, HIGH);

digitalWrite(B, LOW);

digitalWrite(A, LOW);

int voltage = analogRead(A0);

float electrode1 = (voltage * (5.0 / 1023.0) + 10);

electrode2 = (50);

electrode4 = (40);
```

```
 electrode0 = (30);

 electrode6 = (20);

 Serial.print(electrode2);

 Serial.print(" ");

 Serial.print(electrode4);

 Serial.print(" ");

 Serial.print(electrode0);

 Serial.print(" ");

 Serial.print(electrode6);

 Serial.print(" ");

 Serial.println(electrode1);

 }

delay(1);

}
```

Appendix G.

Ethics Approval

SFU RESEARCH ETHICS

Minimal Risk Approval – Delegated

Study Number: 30001017
Study Title: 3D Printed Wearable EMG System

Approval Date: June 13, 2022
Principal Investigator: Woo Soo Kim
Faculty/Department: Mechatronic Systems Engineering

Expiration Date: June 13, 2023
SFU Position: Faculty

Student Lead: N/A
SFU Collaborator(s): N/A
Research Assistant: Haotian Su
External Collaborator(s): N/A

Funder: NSERC
Funding Title: 3D Printed Electronics
Funding Number: N/A

Document(s) Approved in this Application:

- Consent Form, version 2 dated May 25, 2022
- EHS Approval Letter

The application for ethical review and the document(s) listed above have been reviewed and the procedures were found to be acceptable on ethical grounds for research involving human participants.

The approval for this Study expires on the **Expiration Date. An Annual Renewal must be completed every year prior to the Expiration Date. Failure to submit an Annual Renewal will lead to your study being suspended and potentially terminated.** The Board reviews and may amend decisions or subsequent amendments made independently by the authorized delegated reviewer at its regular monthly meeting.

This letter is your official ethics approval documentation for this project. Please keep this document for reference purposes.

This study has been approved by an authorized delegated reviewer.

SIMON FRASER UNIVERSITY ENGAGING THE WORLD Page **1** of **1**