DEVELOPMENT OF A CONFORMAL TEXTILE ELECTRODE LINER FOR THE PURPOSE OF MYOELECTRIC PROSTHESIS CONTROL

by

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Abstract

In upper-extremity myoelectric prostheses, hard surface electromyography (sEMG) electrodes are typically placed in a rigid or semi-rigid socket worn over the residual limb. The resulting interface does not offer optimal suspension and cannot adapt to volume changes that the residual limb inevitably experiences. These volume changes can lead to electrode lift-off and electrode shift, resulting in weaker control or loss of control of the prosthesis.

The work in this thesis focuses on the development of an alternative electrode interface that we will refer to as the “MyoLiner”. The MyoLiner comprises textile electrodes embedded directly into a roll-on silicone liner to create a secure and conformal interface. By using textile as the electrode substrate, the electrodes can also be embedded and sealed in a silicone liner while maintaining the liner’s suspension integrity.

We have developed both remote and active textile electrode prototypes. This thesis demonstrates the potential of these newly developed textile electrodes and establishes the groundwork for the development of a fully-integrated version of this
ABSTRACT

interface.

Impedance testing with the remote textile electrodes has shown that electrode-skin impedance similar to that of conventional metal dome electrodes can be achieved with the presence of sweat or water. Noise analysis has shown that the active textile electrodes can effectively minimize power line and cable interference. Preliminary functional test results have also suggested that the textile electrodes can reliably control a prosthesis. A Cue Test showed non-significant differences in time needed to achieve 3 out of 4 conventional sEMG triggers for direct control, and a case study with a pattern recognition system has shown comparable separability of patterns during training and comparable cue completion during evaluation.

An envelope SNR study has suggested the sEMG envelope quality of the textile electrodes is not matched to the envelope quality of conventional Otto Bock electrodes. An active textile electrode design utilizing a high front-end analog gain can address this issue. Future work involves developing this next generation active textile electrode prototype and fully integrating the electrodes into a silicone liner and ultimately into a prosthesis.

Primary Reader: Nitish V. Thakor, PhD
Secondary Reader: Albert Chi, MD
Tertiary Reader: Youseph Yazdi, PhD
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Dedication

This thesis is dedicated to my family: my creative and talented sister, my successful and driven brother, and my wonderful and loving parents. You have been my role models from the start and I only hope that I can make you proud. Thank you for being there every step of the way. Love you all!
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Chapter 1

Introduction

There are over an estimated 1.7 million people living with lower-limb or upper-limb loss in the United States today, with the projection that this number will jump to 3.6 million by the year 2050 [1]. This number includes both congenital absence at birth and amputations. With respect to amputations, the two leading causes are vascular disease and traumatic injury. In vascular-related amputations, termed dysvascular amputations, the limb must be removed because of complications that arise from a defective blood supply to that appendage. Complications with the blood vessels in the limb can eventually progress into vascular disease; those with diabetes have a much higher risk of developing such severe vascular disease. Traumatic-related amputations are the second-leading cause, and are necessary when a limb or appendage has been injured beyond repair. This can occur as a result of car accidents, severe burns, combat injury, and other situations in which the limb becomes significantly severed.
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<table>
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<tr>
<th></th>
<th>Lower-limb</th>
<th>Upper-limb</th>
<th>Total</th>
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<tr>
<td>Dysvascular disease</td>
<td>51.40%</td>
<td>2.49%</td>
<td>53.89%</td>
</tr>
<tr>
<td>Trauma</td>
<td>13.20%</td>
<td>31.76%</td>
<td>44.96%</td>
</tr>
<tr>
<td>Cancer</td>
<td>0.89%</td>
<td>0.19%</td>
<td>1.09%</td>
</tr>
<tr>
<td>All etiologies</td>
<td>65.50%</td>
<td>34.50%</td>
<td>100%</td>
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Table 1.1: Prevalence percentage estimates of limb loss based on level and etiology of amputation. (2005, U.S.) [1].

If the health of the limb cannot be salvaged, there is a risk that infection can spread to the rest of the body, necessitating the removal of that limb. A much smaller percentage of amputations are due to cancer-related causes. Table 1.1 shows the percentage estimates of all amputations that are lower-limb or upper-limb based on etiology from 2005 in the United States [1]. As shown in Table 1.1, dysvascular disease including diabetes and arterial disease are estimated to account for about 54% of all amputations. Traumatic-related amputations are estimated to account for 45%, with a much smaller percentage attributed to cancer (<2%) [1].

While the work in this thesis can be somewhat applicable to lower-limb amputees, the studies in this thesis are focused on the use of upper-extremity prostheses. As can be seen in Table 1.1, upper-limb amputations represent an estimated 34% of all amputations in the U.S. Table 1.2 shows the breakdown of percentage estimates of lower-limb versus upper-limb amputations within each etiology category. The majority of dysvascular disease-related and cancer-related limb loss instances are
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<table>
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<tr>
<th></th>
<th>Lower-limb</th>
<th>Upper-limb</th>
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<tbody>
<tr>
<td>Dysvascular disease</td>
<td>95.27%</td>
<td>2.49%</td>
</tr>
<tr>
<td>Trauma</td>
<td>29.40%</td>
<td>70.74%</td>
</tr>
<tr>
<td>Cancer</td>
<td>77.78%</td>
<td>16.67%</td>
</tr>
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Table 1.2: Percentage estimates of limb loss attributed to lower-limb vs. upper-limb amputations based on etiology (2005, U.S.) [1].

accounted for by lower-limb amputations, about 95% and 78%, respectively, while the majority of traumatic-related limb loss instances are attributed to upper-limb amputations (about 70%) (Table 1.2) [1–3].

As technology advances, the available options for prosthesis control for those living with upper-limb loss have evolved, giving amputees the ability to more easily re-adopt a normal lifestyle. Prosthesis types range from body-powered hooks to myoelectric multi-articulated hands; depending on the individual, a specific type of prosthesis may be more beneficial over the other.

1.1 Upper-Limb Prostheses Types

Minor limb loss in the upper-extremity sense is typically defined as the amputation of the hand or fingers, whereas major limb loss in the upper-extremity sense refers to amputations that are more proximal to the body both below-elbow and above-elbow. The majority of all upper-limb loss instances are minor amputations, with
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only approximately 8% of all upper-limb loss instances being major amputations [1].
As such, there are a variety of upper-limb prostheses available for amputees based on their level of amputation, individual needs and desired functionality. For minor limb loss, there are a range of prosthetic hands as well as partial hands and fingers available on the market. For major limb loss, in addition to prosthetic hands, there are also prosthetic elbows (both mechanical and electrical) that are available to help regain elbow function. Electric wrist rotators are also an option for some, and can be attached to the prosthetic hand to help restore wrist function. While there are a wide range of devices meant to restore different functions of the hand or arm, these devices can essentially be grouped into four different categories: cosmetic, body-powered, myoelectric or hybrid.

1.1.1 Cosmetic Prostheses

Cosmetic devices, also termed passive devices, can be used for basic functionality such as pushing and pulling light objects, or for stability and support. Overall they are not very functional, however, and they serve more of an aesthetic purpose with the aim of making the appearance of an amputation look less obvious. These types of devices, although passive, still offer a boost in the quality of the life for an amputee. The cosmesis can help an amputee overcome some of the negative, self-conscious feelings that accompany limb loss. In many cases, lower-limb amputations can be hidden under clothing and shoes, whereas upper-limb amputations are more apparent. This
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Figure 1.1: livingskin™ passive prostheses by Touch Bionics, Inc. designed using an imaging system that can detect and simulate the amputee’s own skin features. [5]

makes the design of a prosthetic hand or arm challenging, since users would also like the prosthesis to be acceptable from a cosmetic viewpoint [3]. While more advanced prostheses have become available over the decades, studies have shown that there has been no major decline in user acceptability of passive prostheses. In fact, studies have suggested that cosmetic devices have a higher permanent rate of use [4]. This may be because while a passive prosthesis doesn’t necessarily offer much added functionality, it also elicits a smaller number of maintenance issues. Advances in cosmetic prostheses by companies such as Touch Bionics, Inc. (Livingston, U.K.) have led to very life-like prostheses that have synthetic skin features which can closely resemble the amputee’s own skin features (Fig. 1.1). These realistic-looking devices make passive prostheses even more appealing to the user who is worried about the aesthetics of their device.
1.1.2 Body-Powered Prostheses

Body-powered devices are sometimes referred to as cable-driven devices and typically use a system of cables and pulleys to control a terminal device such as a mechanical hand or hook. These devices comprise a body harness, a hard socket that encases the residual limb, a control cable-pulley mechanism and a terminal device (Fig. 1.2).

![Body-powered prosthesis diagram](image)

**Figure 1.2:** Body-powered prosthesis with a split hook terminal device. [6]

The body harness is worn over the opposing shoulder and is connected to the cable-pulley mechanism. This mechanism then relies on movements of the back and shoulders to drive the terminal device. In the case where the terminal device is a split hook, if the amputee pulls his or her arm back, the cable pulls the hook open; when the amputee returns to a relaxed position, the hook closes. While these movements can be tiring and cumbersome over time, these types of devices also tend to be reliable. This is because, if worn correctly, a specific movement will always cause the device to react the same way. Because of this, despite recent advances in the field of electrical prostheses, body-powered devices still remain popular among amputees [4].

Typically for body-powered devices, the body harness is worn over the body both
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Figure 1.3: Body-powered systems and their suspension mechanisms: (a) Harness for both suspension and operation [7]. (b) Amputee wearing body-powered prosthesis with suspension harness [8]. (c) Harness for operation only [7]. (d) Example of self-suspended body-powered system with mechanical hand terminal device [9].

for operation and suspension as shown in Figures 1.3(a) and 1.3(b). If the socket is self-suspended, then a more simple body harness can be used with a strap only on the opposing shoulder that is used for operating the prosthesis (Fig. 1.3(c)). The socket can be self-suspended by using a system with a roll-on silicone liner as is shown in Figure 1.3(d).

Terminal devices can also vary depending on the task at hand and range from the most commonly used split hook to specialized tools made for specific activities such as eating, gardening or even golfing (Figure 1.4) [3].
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Figure 1.4: Various terminal devices for a body-powered prosthesis. [10]

Studies have shown that rejection rates tend to depend heavily on the type of terminal device used [4]. Body-powered mechanical hands are poor in terms of grip strength and usability and typically require a lot of energy while achieving little functionality; these devices are associated with rejection rates above 80% [3,4]. Body-powered hooks on the other hand, like the split hook (Figure 1.2), are much more functional and offer superior grip ability, durability, and mechanical strength. These are associated with lower rejection rates [4,11,12].

Restoring elbow function is also another challenge. While below-elbow amputations can still take advantage of their available elbow function, above-elbow amputees face an additional challenge in controlling a prosthetic elbow. To operate an additional body-powered prosthetic elbow, a more complicated body harness and cable mechanism is required. Another option is to use a mechanical lock that is manually controlled by the opposing hand in order to dictate the elbow function [3].

In general, disadvantages of body-powered prostheses are usually centered around harness breakage, discomfort and/or irritation as well as poor aesthetics [4].
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1.1.3 Myoelectric Prostheses

Myoelectric devices are electrically powered prostheses that utilize muscles signals from the amputee’s residual limb for control. These devices comprise a socket that encases the residual limb, electrodes that pick up the muscle signals known as surface electromyographic (sEMG) signals and a terminal device that is controlled using these EMG signals. The myoelectric prosthesis category includes electric hands, wrist rotators and elbows (Figure 1.5). The more known and established companies producing myoelectric devices are Otto Bock (Duderstadt, Germany), Touch Bionics, Inc., RSLSteeper (Leeds, U.K.), Motion Control, Inc. (Salt Lake City, UT, U.S.) and Liberating Technologies, Inc. (Holliston, MA, U.S.).

![Myoelectric devices: (a) Boston Digital Arm™ by Liberating Technologies, Inc. [13] (b) Wrist rotator by Otto Bock [14] and (c) Bebionic3™ hand by RSLSteeper [15].](image)

To use these devices, electrodes are placed in a socket worn over the residual limb, establishing contact between the electrodes and the surface of the residual limb. Electrode locations are chosen based on the control modality being used (described in section 1.3), and the respective signal patterns are used to operate the devices. These
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devices can be more intuitive to use than body-powered prostheses and are usually harness-free, making them more appealing to some users. For some, however, higher costs, more maintenance and control variability may outweigh the benefits [4]. The terminal devices can be as basic as an electrically powered “hook” or as advanced as a multi-dexterous hand capable of individual finger movement. Theses advanced multi-articulated hands on the market such as the i-Limb Ultra™ by Touch Bionics, Inc. or the Bebionic3™ by RSLSteep (Figure 1.6) can provide amputees with more dexterity, but also require more training on the part of the amputee. Despite these apparent drawbacks, myoelectrically operated hands tend to have lower rejection rates in comparison to body-powered mechanical hands [12, 16].

![Figure 1.6: Multi-articulating hands: (a) Bebionic3™ by RSLSteep [17] (b) i-Limb Ultra Revolution™ by Touch Bionics, Inc. [18].](image)

Figure 1.6: Multi-articulating hands: (a) Bebionic3™ by RSLSteep [17] (b) i-Limb Ultra Revolution™ by Touch Bionics, Inc. [18].
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1.1.4 Hybrid Prostheses

Hybrid devices are true to their name, combining two or more of the above categories of devices into one prosthesis. Sometimes the needed functionality can be better achieved by using more than one form of control platform [19]. An example of this would be a myoelectric hand coupled with a cable-driven mechanical elbow [20]. Myoelectric elbows can be difficult to operate if there are not enough sites for control or if the residual limb is short. As such, a cable-driven elbow with an internal locking mechanism in conjunction with a myoelectric hand may be a more functional option than a fully-myoelectric prostheses (hand and elbow).

Another example of a more complex hybrid system developed by Mark Lesek from Dynamic Welding and Engineering Pty Ltd (Tasmania, Australia) is an electrically operated prosthesis controlled using a body-powered harness. A harness housing two cables is worn by the amputee, and each shoulder controls the pull of one cable. Rather than the cables directly driving the movement of the prosthesis, however, the cables are instead connected to linear actuators which are then able to translate that movement into an electrical signal fed to the prosthesis. One shoulder movement is in charge of controlling the electric elbow, and the other is in charge of controlling the electric hand. In this setup, the body-powered harness is used to control an electric system.
1.2 Surface Electromyography (sEMG)

1.2.1 The Nature of the Signal

This work focuses on the development of an alternative textile electrode interface for control of myoelectric prostheses. As such, myoelectric devices and the sEMG signals needed to drive them will be the focus of this thesis. Surface EMG signals have become an important tool in the field of biomechanics due to their non-invasive nature of acquisition. The signal can be used for a variety of things from device control to rehabilitation diagnostics [21]. Myoelectrically operated devices or stroke assistive devices use sEMG to activate motors on the device. From a rehabilitation standpoint, sEMG can be used to observe characteristics such as the onset of muscle activation, qualitative muscle force and muscle fatigue [22,23]. While relatively easy to acquire, the sEMG signal is a complex one in nature, and is influenced by a variety of electrophysiological and environmental factors [21,24].

Essentially, the signal that is observed on the surface of the limb during a voluntary contraction is the manifestation of the contributions from many active motor units and their motor unit action potential (MUAP) trains intermingled in an interference pattern [25]. A pair of electrodes are placed on the skin to record these MUAPs, and the differential signal observed between these two electrodes result in the sEMG signal (Fig. 1.7). In a voluntary contraction, the MUAPs are relatively asynchronous resulting in a stochastic signal, whereas in an electrically-stimulated contraction, the
Figure 1.7: How the sEMG signal is generated: recording electrodes (red) pick up differential potentials observed on the surface of the skin with respect to reference electrode (blue) that are generated as a muscle is flexed.

MUAPs tend to be more synchronous. [25,26]. An sEMG signal can be thought of as the superposition of a broad survey of multiple MUAPs created by a group of muscle fibers during a muscle contraction. As such, stronger contractions will typically elicit larger amplitude EMG signals [24, 25]. The sEMG waveform is complex, however, and many different factors in addition to contraction strength affect the shape and amplitude of the sEMG signal acquired, especially between individuals.

Some deterministic characteristics that affect the sEMG signal directly are the number of active motor units, the motor unit firing rate, duration and amplitude of recorded MUAPs and recruitment stability of the motor units [22]. While the muscle acts as the source of electrical activity, the surrounding low-resistance conducting
medium composed of interstitial fluid, blood, and tissue act as a volume conductor. Due to the nature of this volume conductor, signals generated by deeper muscle tissue are attenuated by the time they reach the skin surface [24,25,27]. In addition, signals from other muscle groups that aren’t directly under the recording site can travel through the volume conductor and still be manifested at the recording site. This is known as “cross-talk”, and if it is very prominent, can sometimes be an issue when trying to use antagonistic muscles to control a myoelectric prosthesis [22,28]. The degree of attenuation and spatial distribution of the sEMG signal depends partly on the presence of subcutaneous fatty tissue. A thicker fat layer between the muscle of interest and the electrode site will result in a greater attenuation of the sEMG amplitude [29–32]. Furthermore, if the subcutaneous fat layer covers the muscle of interest and other surrounding muscles, it is found that the fat layer also contributes to an increased degree of cross-talk. This is because the sEMG signal will “spread” throughout the fat layer and appear at different areas on the skin surface [29,31]. As such, muscle depth, orientation and location with respect to the recording site of both the muscles of interest and those not of interest all play a role in the observed sEMG signal.

Extrinsic factors can also affect the signal. For example, electrode location, spatial orientation and placement will influence which signals are observed [22]. For example, muscle tissue has higher conductivity in the axial direction than in the radial direction so the orientation of the recording electrodes with respect to the muscle will affect the
shape of the resulting sEMG signal [22, 25]. Other factors such as electrode size and material can also affect the recording; these issues will be discussed in more detail in Chapter 2.

1.2.2 Signal Processing

Typically, a completely unfiltered raw sEMG signal is not used. The complex signal must be processed to make it “useful” for its intended application. The first processing step is usually band-pass filtering. The band-pass filter helps isolate the signals of interest and reduce the effects of other interfering physiological signals and artifacts. On the high-pass side, the cutoff frequency of choice is somewhere between 10-20 Hz [23, 33]. Most baseline drift and motion artifact manifests itself as a low-frequency signal below 20 Hz, and so this cutoff is chosen to remove those artifacts without also removing important sEMG frequencies. The low-pass cutoff frequency ranges from 500 Hz to 2 kHz, with 500 Hz being more common [33], since most sEMG information is within the 20-500 Hz band. The signal is also usually amplified at this point. These first stages of filtering will result in the “Original EMG signal” shown in Figure 1.8(a). This type of signal can then be used for pattern recognition, where the classification algorithm then extracts features such as sEMG variance, mean absolute value (MAV) and waveform length (WL) off-line to discriminate between classes [33].

For myoelectric prostheses using amplitude-based control and for muscle-force observations/diagnostics, the sEMG amplitude is usually the processed signal of in-
terest. There are several techniques that can be used to obtain an sEMG envelope to represent the sEMG amplitude. In Figure 1.8, the rectification + low-pass filtering technique is used. Rectification first takes the absolute value of the signal, shown in Figure 1.8(b), and low-pass filtering then takes a moving average of the signal, resulting in the EMG envelope shown in Figure 1.8(c). When obtaining the envelope using digital techniques, this envelope can be representative of the mean absolute value (MAV) or average rectified value (ARV), or alternatively the root mean square (RMS). When dealing with muscle-force observations, the RMS value is thought to

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**Figure 1.8:** Filtering steps for acquiring the sEMG envelope. [33]
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be more appropriate since the RMS is representative of signal power where the ARV is simply the area under the sEMG signal [22, 23].

Noise filtering is another aspect of signal processing. By using amplifiers with a high common mode rejection ratio (CMRR), ambient noise that is common to both electrode contacts are subtracted from of the signal [23]. Even with high-CMRR amplifiers, however, a common issue is 50/60 Hz electrical line noise that disrupts the signal since it is at a frequency within the band-pass of sEMG. A specialized filter called a notch filter is sometimes used to remove that specific frequency signal [33]. Other factors related to the impedance of the electrodes and skin can also be manipulated to reduce this line noise (discussed more in detail in Chapter 2). Another form of “noise” in terms of unwanted muscle signal is cross-talk, as discussed in section 1.2.1. While ambient noise common to both electrodes can usually be removed, an sEMG muscle signal from further away will manifest itself as a low-frequency signal and present itself on both electrodes as a common differential sEMG signal. One electrode configuration known as double-differential configuration can filter out this cross-talk by using three electrodes rather than two to eliminate this common differential signal [26].
1.3 Myoelectric Device Control

Modalities

As mentioned before, muscle signals present on the surface of the skin (sEMG) are used to control myoelectric devices. These signals are amplified, processed and interpreted to control which and when different motors of a prosthesis should move. The first instance of myoelectric control was seen in a May 1945 patent application by Reinhold Reiter, a physics student at Munich University [34]. In this first prototype, the sEMG signal from a contracting muscle was used to control a wooden hand actuated by an electric solenoid. Different “rhythms of contraction” and built-in delay triggers would dictate when the hand opened or closed [34]. In the 1970s, improved signal processing techniques and methods began to bring R&D attention to more advanced control modalities for myoelectric devices [35]. These methods of control for myoelectric prostheses can be tuned to each individual’s own ability. There are always new ideas and/or combinations of control strategies that are being developed in an attempt to better control myoelectric devices on the market. Several of the more common control modalities are discussed in the following sections.

1.3.1 Direct Control

With direct or “conventional” control, the sEMG signals that an amputee generates directly drive specific motors to activate states of a prosthesis. The most common
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form is a three-state controlled system; a prosthetic hand, for example, would have the three states of “open”, “close” and “rest” [36]. This form of direct control can be either amplitude-dependent or rate-dependent [21, 35]. The number of available sEMG sites can also influence how these states are reached and the number of states or degrees of freedom (DOF) that can be achieved. The number of available muscle sites can be limited depending on the size and shape of the residual limb. Typically direct control systems will use one to two muscle sites that are available on the residual limb [37]. Furthermore, while a below-elbow amputee may have sites on the forearm available that can still generate strong sEMG signals, the residual limb or absence of one for above-elbow amputees and shoulder disarticulates makes the issue of finding a suitable muscle site more challenging. An option for these amputees is targeted muscle reinnervation (TMR) surgery. With TMR, typically five or six residual nerve-muscle units that can be independently controlled are reinnervated by surgically anastomosing them to divided areas of a residual muscle such as the pectoralis major (for shoulder disarticulates). Subcutaneous fat above the muscle sites is removed to minimize sEMG attenuation and cross-talk, further isolating each muscle site of interest based on the specific site of reinnervation. This allows for independent control of each separated area on the residual muscle, creating multiple sites now available and suitable for direct control [38,39]. With conventional direct control, only one DOF can be operated at a time, and a “mode-switch” is used to switch between hand and wrist control, for example. With TMR, however, simultaneous direct
control of multiple devices becomes possible, and an amputee can operate up to 4 DOF simultaneously, allowing them to more intuitively control multiple myoelectric devices [39].

1.3.1.1 Amplitude-Dependent

Amplitude-based control is typically what is used by direct controlled myoelectric devices on the market. States of the device are activated by using a threshold-based approach. When the amplitude of an sEMG signal passes a set threshold, the state is then activated [35, 40]. These thresholds can be set using each device’s own software in order to choose a level that is most comfortable for each individual amputee. High thresholds can generate fatigue, whereas low thresholds will pick up too much noise; as such, a happy medium must be found. One or multiple sites can be used with this amplitude-based approach.

Single-site control is not the most straightforward, and is most likely only used when there is a limited number of suitable sEMG sites on the residual limb and multiple states are desired. In this method, three states can be activated with one contracting muscle by using three-level control. With three-level control there are three different thresholds involved: one threshold per state [41]. Dual-site control using antagonistic muscles is the more commonly accepted approach. With this method, two different muscle sites control two states of the device [40]. For example dual-site control with one site on the forearm flexors and the other on the forearm extensors.
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is typically used to control the closing and opening of a myoelectric hand.

Combinations of these techniques can be used to access more complex control systems. For example, a five-state controller that can operate two different devices (for example: open hand, close hand, flex elbow, extend elbow, rest) can be achieved with only two muscles by using dual-site, three-level control [40]. Another optional feature is proportional control. With proportional control, the speed of the state can also be manipulated in a proportional manner. Passing a threshold will activate the state such as the opening of a hand, but as the sEMG signal amplitude increases past the threshold, the speed of the opening will also increase [33].

1.3.1.2 Rate-Dependent

Rate-based control is an alternative to the amplitude-based approach. Rather than only relying on the sEMG signal to pass a set threshold, the rate at which the muscle contracts will control which state is activated. For example, with single-site control, no contraction will keep a device at rest, a slow contraction will activate the second state (open hand) and a fast contraction will activate the third state (close hand). Dual-site rate-dependent control can be used to activate up to five states and consequently two different myoelectric devices [40,41].
1.3.1.3 Grip Switching

Up to this point, discussion has been focused on three states per device. For an electric elbow, this would make sense, since the functionality that can be achieved with the elbows on the market is only rest, elbow flexion and elbow extension. For multi-dexterous myoelectric hands, however, many different grips are possible. With the conventional amplitude-based, dual-site control, only two sites are available for accessing those grips, making the use of those added degrees of freedom more challenging [42]. As such, grip switching is typically conquered by generating sEMG “triggers” that the hand is designed to recognize. These patterns include: co-contraction, where a threshold must be hit at the same time by two muscles, holding open, where a signal must be held above a threshold for a set period of time, and double-pulses and triple-pulses, where a set threshold must be reached two or three times, respectively, within a certain time frame. When a trigger is recognized, the hand switches modes to activate that grip. The two sites are then still used to open and close the hand in that specific grip mode. A novel RFID-enhanced approach developed by Infinite Biomedical Technologies (Baltimore, MD) presents an alternative to this grip switching methodology. With this novel method, an external circuit called morph™ is plugged into to the hand, giving the myoelectric hand wireless RFID capability. RFID tags can then be programmed to represent specific grips, and if the hand is brought near a tag, the grip is automatically switched. This reduces the overall cognitive burden and physical effort the subject must exert to control his or her prosthesis.
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Alternative options like iPad applications and other manual button switches can be used to change grips as well.

1.3.2 Pattern Recognition

Another form of control that is more in the research phase as an alternative to direct control is pattern recognition [21, 33, 43]. With pattern recognition, multiple sites on the limb are used to record sEMG signals. Rather than using specific signal sites to directly drive specific motors on the device, naturally-generated contraction patterns are instead decoded and used to more intuitively operate the prosthesis.

Initially, sEMG signals are recorded from multiple sites surrounding the residual limb during a calibration training period where the amputee performs different “grips” or “actions” that he or she would like the prosthesis to perform. To record sEMG from multiple sites, the electrodes are placed around the residual limb, or for TMR patients, on and surrounding the area of reinnervation [37, 44]. Even though the amputee cannot physically perform the action itself with a live appendage, the residual muscles and nerves can still generate the contraction that would normally perform that action. After calibration, specific features are then extracted from the signals and processed using an LDA algorithm to best distinguish each pattern from one another and classify each pattern as a desired action [21, 45].

Theoretically with patterns now internally recorded, every time the user naturally generates a specific contraction in an attempt to perform an action, the prosthesis
can then decode the pattern based on the calibration data and perform that action in an intuitive manner, reducing the cognitive burden for the prosthesis user. Pattern recognition has the potential to provide an amputee with many more degrees of freedom and the ability to more easily operate multiple myoelectric devices [43]. This mode of control, while possibly more natural for the user, is challenging for the engineer. Development is still needed in this field to get a reliable device that uses pattern recognition on the market.

1.4 Current Myoelectric Prosthesis

Interface

1.4.1 Fitting the Patient

There are many steps involved with fitting an amputee for a myoelectric prosthesis. After an amputation, the residual limb experiences swelling, and the prosthetist must apply wraps and/or bandages to provide compression over the limb. The initial aim is to reduce the overall size of the limb eventually down to a stable size and shape. After wound closure, the amputee’s residual limb must constantly be evaluated to determine if it has shrunk to a stable size, is properly healed and is ready for prosthesis wear [46, 47]. The time needed for this preparatory phase varies from individual to individual, however, and throughout this process, multiple casting iterations are made
to fit the amputee with diagnostic test sockets. By wearing these test sockets, the rehabilitation process is expedited and the residual limb is better stabilized in volume and shape [46, 47]. Once the residual limb is determined to be ready for prosthesis wear and the amputee is in good physiological and psychological health, a “final” socket is made for the amputee [10, 46, 47].

Electrodes are then placed in a socket to be worn over the limb, slightly protruding from the inner surface of the socket to make contact with the surface of the residual limb. They cannot be placed blindly within the socket, however; the electrodes need to be positioned in the socket in a specific manner in order to target certain areas on the limb. To do this, a prosthetist surveys the strength and clarity of sEMG signals that can be generated from certain muscles on the limb, and mounts the electrodes in the socket accordingly [33]. As discussed earlier, the strength and shape of sEMG signals will vary from location to location, and from patient to patient. This makes the fitting process custom to each individual, with specific electrode sites chosen based on the individual’s own signal characteristics.

1.4.2 Suspension with a Rigid Socket

To maintain good electrode-skin contact with rigid electrodes, typically a rigid to semi-rigid socket is needed to secure the electrodes down onto the residual limb. The socket is worn in a “push-in” fashion where the amputee pushes his or her arm into the socket. Self-suspension for below-elbow prostheses is typically achieved using
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epicondylar suspension by essentially “wedging” the arm into the socket and applying pressure to the epicondyle area of the residual limb. For above-elbow prostheses, this tight fit is achieved closer to the shoulder. A body harness can also be worn to help suspend and distribute the load of the prosthesis.

Over time, these tight fitting sockets can be uncomfortable. High pressures experienced by the limb due to loading from the hard socket can greatly compromise the health of the limb, and can lead to issues such as skin irritation, epidermoid cysts, verrucous hyperplasia or other skin conditions [48]. High pressures, while seen more commonly with lower-limb sockets, can still affect the limb at a deeper, subcutaneous level, facilitating the development of painful pressure ulcers [49, 50]. Furthermore, to achieve higher suspension for below-elbow prostheses, the shape of the socket will typically extend over the epicondyles and reduce the amputee’s overall range of motion. Adjusting the shape of the socket to increase the user’s range of motion and overall comfort, however, will typically result in a decrease in suspension quality [51].

1.4.3 Changes in Limb Volume

Immediately after amputation, the residual limb volume can fluctuate drastically, experiencing initial swelling then overall shrinkage over time [47, 52]. In a long-term setting, the residual limb can change size and shape as the amputee experiences weight gain or weight loss. Depending on the health of the amputee and his or her limb, the volume of the limb can also fluctuate [53]. The forces exerted on the residual
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limb by the hard socket can also force fluid out of the interstitial space of the limb tissue, causing the overall size of the limb to change [54]. Throughout the day, natural diurnal movement may cause the volume to fluctuate. These changes in volume pose a problem for rigid sockets, because they cannot adapt to this change. As such, if an amputee experiences a more long-term volume change in the residual limb such as is experienced with weight gain or weight loss, a new socket must then be fabricated to maintain a good fit [46, 47, 55].

In a short-term setting, shape or volume changes of the limb can also occur. Depending on the load bearing of the prosthesis or the position of the arm in space, certain areas of the limb will contract, causing the limb to change shape. Areas of the residual limb will rise due the contracted muscles whereas other areas on the limb will become recessed, and can lift away from the inner side of the socket.

1.4.4 Resulting Electrode-Skin Interface Issues

As discussed above, fluctuations in limb volume in both a short-term and long-term setting are common. When these limb volume changes occur, the electrode-skin interface can lose its integrity and stability. If the arm is no longer filling the entire socket, it becomes free to move within it. As mentioned before, electrode location is crucial when trying to obtain the best sEMG signals for precise myoelectric control [28, 33]. If a limb can move within the socket, however, this movement leads to changes in electrode locations with respect to the limb, resulting in electrode shift,
and possible sub-optimal electrode locations that were not originally chosen during
the initial fitting process (Fig. 1.9(b)). Volume loss in the limb can also lead to
electrode lift-off, where the actual electrode contact is now physically compromised.
If an amputee is bearing a load on the prosthesis or even during normal muscle
contractions, areas on the residual limb can recess away from the inner surface of the
socket, and can also lead to electrode lift-off (Fig. 1.9(c)). If the contacts are not
secured onto the skin, motion artifact and unwanted noise is introduced, disrupting
the sEMG signal and leading to a loss of control of the prosthesis.

Figure 1.9: Model of the electrode-skin contact under different conditions: (a) ideal
contact (b) electrode shift and (c) electrode lift-off.

1.5 Addressing the Interface Problems

As mentioned before, the use of hard electrodes in a rigid or semi-rigid socket
for myoelectric prosthesis can lead to a sub-optimal electrode-skin interface. Volume
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changes in the limb can reduce the contact between the electrode and limb, causing the user to have less control of his or her prosthesis. Suspension is also limited by this interface, as the hard socket sometimes in conjunction with a harness will grip areas of the residual limb to hold the entire prosthesis in place, reducing comfort and the range of motion for the amputee. Therefore, a solution is needed for myoelectric prostheses that can maintain the needed electrode-skin contact during volume fluctuations and also provide more comfortable mechanical suspension.

1.5.1 The Lower-Limb Solution

In lower limb prostheses, one way that amputees try to cope with local stresses and volume fluctuations is with the use of socks and liners that can be worn over the residual limb underneath the rigid socket. Prosthetic socks are available in a variety of thicknesses and materials and provide some padding to protect the limb’s bony prominences [56]. Roll-on liners are also used to try and increase comfort and suspension. These liners are first inverted, then rolled onto the residual limb, re-inverting the liner as it is rolled on. As the liner is rolled onto the limb, a vacuum is essentially formed around the residual limb and suction suspension is achieved. Originally made using Pelite, a polyethylene foam, these liners are now generally made from elastomers such as polyurethane or silicone. These gel-like materials provide protection against friction and help distribute load across the entire surface of the residual limb [48]. Because of their more tissue-like characteristics, these liners can
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adapt with varying load bearing and can flow away from regions of high pressure, distributing the pressure more evenly across the limb.

In addition to providing additional comfort, these socks and liners also help amputees maintain proper socket fit after the residual limb has experienced volume loss. Wearing additional socks and/or a silicone liner can help fill the resulting void that is created between the limb and socket walls after volume loss. Silicone, specifically, also has adhesive properties that help keep the liner and socket in place throughout the day, offering the amputee an added degree of suspension.

1.5.2 Existing Solutions for Myoelectric Devices

While silicone liners and socks used for lower-limb prostheses can also be used for upper-limb, body-powered devices, they cannot typically be worn, at least in the traditional sense, over the residual limb with myoelectric prostheses. This is due to the fact that the sock or liner would cover the surface of the residual limb, making sEMG signal pick-up impossible. As such, there have been attempts in the clinical space to address the interface issues laid out in section 1.4.4, however the solutions are still sub-optimal.

To create an interface that is more flexible, rolled silicone sockets are sometimes used with conventional electrodes (Fig. 1.10). Rolled silicone is thinner and more flexible than a hard socket, but is still very different from a roll-on silicone liner in the sense that a rolled silicone socket does not offer added suspension. The amputee
Figure 1.10: Rolled silicone socket embedded with conventional Touch Bionics electrodes.

still pushes his or her arm into this silicone socket as he or she would with a conventional rigid socket, and an anatomical socket is worn over it to maintain suspension and electrode-skin contact. While the socket itself may be more comfortable and/or flexible, overall suspension is not improved and the rigid electrode nature can still lead to the same electrode-skin contact issues.

The use of a modified roll-on silicone liner with a myoelectric device to increase suspension and overall range of motion has also been investigated in the clinical space. One attempted solution is to use a roll-on silicone liner underneath the hard socket with holes cut out of the liner that correspond to the locations of the electrodes. In this case, the electrodes are still embedded in the hard socket, and the amputee must align the locations of the electrodes to make contact with the residual limb through the cutouts (Fig. 1.11) [57,58].

The holes in the cut-out liner however, pose a problem, since suction and suspension are lost in those areas. Furthermore, electrode alignment with the cutouts is a difficult a process and can make donning and doffing the prosthesis a challenge for
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Figure 1.11: Silicone liners with cutouts for electrode placement. [58]

the user. Edema at the cutout locations has also been observed [58, 59].

In another clinical application, remote metal dome electrodes are passed through a roll-on liner with protruding connectors on the external side of the liner. The electrodes are then connected to pre-amplifiers downstream of the liner [60]. With this approach, overall range of motion and suspension has been shown to be improved [60, 61]. A modified example of this application is shown in Figure 1.12. This figure shows two roll-on silicone liners that were cut into a cuffs for in-house use.

As can be observed, this solution requires each electrode to be placed in the

Figure 1.12: (a) Roll-on silicone cuff embedded with remote metal dome electrodes and (b) failed attachment points.
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liner individually which can be tedious for the prosthetist, especially for a multiple-electrode interface. Cable management is also an issue, since the pre-amplifiers must be connected to each pair of electrodes downstream from the contacts themselves (Fig. 1.12(a)). Furthermore, after rolling the liner on and off multiple times, the attachment points at the electrode-silicone interface begin to fail. The holes begin to tear, compromising the attachment, and the electrodes become loose within the liner. An example of this tearing that was observed on another in-house cuff is pictured in Figure 1.12(b).

All these current solutions for improving the myoelectric interface approach the problem from a different angle and are able to address some but not all the interface issues discussed in section 1.4. As such, we propose an alternative textile-electrode interface that aims to address both interface issues of suspension and maintained electrode-skin contact. This alternative solution is explained in the following section.

1.5.3 Proposed Textile-Electrode Interface

The alternative textile-electrode interface addressed in this thesis comprises our newly developed flexible textile electrodes integrated into a roll-on silicone liner. By using textile for this application instead of a hard electrode, the electrodes can now be fully embedded directly into the silicone liner. With this design both the silicone liner and the electrodes themselves are flexible, creating a completely conformal interface that can maintain the electrode-skin contact during volume fluctuations (Fig 1.13).
Figure 1.13: (a) Model for the electrode-skin interface with a textile-electrode interface embedded in a silicone liner and (b) proof-of-concept manufactured in-house.

Furthermore, with hard metal electrodes passed through the liner, repetitive donning and doffing of the liner can lead to tears at the attachment points, whereas a fully embedded solution will not. To address this, we’ve designed the electrodes to comply with a simple embedding procedure that generates minimal openings that need to be sealed throughout the process. This embedding procedure is described in more detail in Chapter 5. By having minimal openings, we increase the chance of a tight seal and a more durable interface that can maintain suction and consequently, suspension.

The textile electrodes are interfaced with textile data cables as well. With the use of these textile data cables, the cables can also be easily embedded in a liner (Fig. 1.14), allowing for convenient cable management. This design leads to a fully-packaged system that simply needs to be “plugged” into the prosthesis. The overall integration of the liner into a prosthesis is described in more detail in Chapter 5.
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1.6 Thesis Objectives

Although myoelectric control can be more natural than body-powered control, there are some pitfalls that prevent amputees from using this form of prosthesis control. As discussed, one major pitfall is that due to the rigid electrode interface, the reliability of sEMG signals to control the hand as desired is not always optimal. When considering issues such as motion artifact and noise interference, it is clear that a secure electrode-skin interface in any setting is the most important. To reiterate what was covered in the previous section, with hard sockets and rigid electrodes that are being used today, the electrode-skin interface is not optimal. Over time and even throughout the day, limb volumes change, and what may have been good contact initially between the skin and electrodes, begins to suffer. Electrode lift-
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off and electrode shift can occur, compromising the integrity of the sEMG signals. Furthermore, because the rigid interface cannot adapt to changes in limb volume, even if the residual limb wounds have healed soon after amputation, the amputee must wait until the residual limb has achieved a stable volume before being fitted with the prosthesis. An adaptable interface, on the other hand, would allow for the amputee to be fitted sooner and can expedite the rehabilitation process [52]. Lack of optimal suspension is also an issue. With the use of a conventional myoelectric hard socket, comfort and range of motion are both compromised, which also discourages long-term use of the prosthesis.

In this thesis, we propose an alternative interface that will be referred to throughout this thesis as the “MyoLiner”. The MyoLiner design focuses on providing a solution that addresses the above-discussed issues. This design comprises textile-based electrodes that are fully embedded in a roll-on silicone liner. Conceptually, this design provides a conformal electrode-skin interface that is secure even when the residual limb experiences changes in shape or volume. The stability of the liner can also maintain the electrode-skin contact, reducing the negative effects of motion artifact. Furthermore, the porous nature of the fabric used to fabricate the textile electrodes make them easy to embed into a silicone liner without creating large holes that could compromise the suspension of the liner. The padding and adhesive properties of the silicone liner improve overall comfort level and suspension of the prosthesis as well.

The primary objectives of this thesis are to characterize the electrode-skin inter-
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face, model and discuss sEMG electrodes and their design, report preliminary results obtained by various experiments conducted using newly developed textile electrodes and establish the groundwork for a fully-integrated version of the MyoLiner.

1.7 Thesis Overview

Chapter 1 was an introduction to the field of prosthetics and the different prostheses types on the market today. The chapter also gave a brief overview on the nature of the surface EMG signal and the signal processing steps needed to make the signal useful for its application. The chapter further focused on myoelectric prostheses and their different control modalities, and finally, it outlined the issues with the current electrode-skin interface for myoelectric prosthesis and the MyoLiner solution being addressed in this thesis.

Ch. 2 focuses on characterizing and modeling important characteristics and limitations of surface EMG electrodes as well as the electrode-skin interface. Noise analysis also describes the different sources of interference that are encountered when trying to obtain an sEMG signal. These sources of interference are modeled mathematically and related to their real-world counterparts and observations.

Ch. 3 briefly overviews current uses of textile electrodes today. The challenges faced during the development of the textile electrodes and the electrode design itself are also discussed.
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In Ch. 4, the preliminary experimental results regarding signal quality and performance of the electrode in both a research and functional setting are reviewed. Impedance and noise analysis focus on the electrodes from a signal quality standpoint. Three other experiments (Envelope SNR, Cue Test, and Pattern Recognition) focus on the quality of the electrode signal from a functional standpoint.

Ch. 5 relays the big-picture overview of how the textile electrodes are to be interfaced into a roll-on silicone liner, and how this electrode-embedded liner is eventually to be interfaced with a prosthesis. The overall vision as well as how we imagine the final MyoLiner will be used from both the prosthetist and amputee standpoint are modeled and described.

Finally, in Ch. 6 we conclude the thesis by reviewing once more the thesis objectives and how they were met. The future steps needed to move forward towards a finalized device and to improve the overall experimental process are also laid out in detail.
Chapter 2

Characterizing and Modeling the Electrode-Skin Interface

2.1 Interface Circuit Models

The electrode-skin interface is affected by a variety of factors both related to the skin and the electrode of use. To understand why these factors have an effect, a good approach is to electrically model the elements involved. The skin model will first be discussed, then the combined electrode-skin model will be covered.

2.1.1 Skin Circuit Model

Historically, the equivalent circuit model for tissue or skin has been investigated by measuring tissue impedance. Typically, the impedance is obtained by applying a
controlled sinusoidal or other known current to the skin and measuring the resulting drop in voltage. The resulting complex impedance can be computed based on the RMS amplitude of the signal and the phase shift in the input-output current-voltage signal [62, 63]. In a four-terminal or tetrapolar setup, a total of four electrodes are used. Two electrodes are used for applying the current to the skin and a different two are used to detect the voltage drop. By measuring the interface this way, the electrode impedance is not included in the measurement and the resulting values represent only the skin or tissue impedance.

The impedance is dependent on frequency, and as such, typically a bioimpedance analyzer is used to perform an automatic frequency sweep to observe the frequency-based impedance. Bioimpedance analyzers are advanced and costly devices however, and some have focused on developing smaller, cheaper alternatives to perform multi-frequency impedance analysis [62, 64]. Another alternative to the bioimpedance analyzer is an LCR meter. The LCR meter can be used to manually sweep the frequencies of interest [65]. This is much more tedious, however, and is more useful most likely when interested in the impedance at set one or few frequencies.

A common way to display the results from a frequency sweep is a locus diagram [63, 66, 67]. The locus diagram model in Figure 2.1 represents the complex impedance observed when analyzing biological skin or tissue and plots the negative reactance versus the resistance.

The impedance modulus ($|Z|\omega$) and the phase angle of the observed phase-shift
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Figure 2.1: Complex impedance locus diagram representing the impedance of tissue. [63]

During impedance measurements (θ) are to calculate the following:

\[ R(\omega) = |Z|\cos(\theta) \] \hspace{1cm} (2.1)
\[ -X(\omega) = |Z|\sin(\theta) \] \hspace{1cm} (2.2)

Equation 2.1 represents the “real” part of the complex impedance, and is representative of the resistive elements. Equation 2.2 represents the “imaginary” part of the complex impedance, and is representative of the capacitive elements. The imaginary part is equivalent to the negative reactance \(-X\), where \(X = \frac{1}{\omega C}\).

By observing the results of this impedance analysis, researchers found that the tissue exhibited a constant-phase angle impedance \(Z_{CPA}\), implying that the tissue could be modeled as a resistor and capacitor in series, \(R_{pol}\) and \(C_{pol}\). Together the series resistor-capacitor element is sometimes referred to as the “polarization impedance.” Deviations in observed impedance from this constant-phase relationship were observed at lower frequencies. This lead to the belief that the correct model
was in fact a constant-phase angle impedance in parallel with a shunt resistor ($R_2$) attributed to Faradaic charge transfer processes that take place at lower frequencies (Fig. 2.2) [67,68].

![Tissue model including polarization characteristics.](image)

**Figure 2.2:** Tissue model including polarization characteristics. [67]

Relating these equivalent circuits back to physiological parameters, it must be reiterated that they are modeled based on the skin’s response to current during impedance measurement. As such, the circuit models represent the interface at the tissue membrane through which the current passes. Referring to the models in Figure 2.2, $R_1$ represents the resistance of the tissue cell interior fluid and $R_2$ and $Z_{CPA}$ (or $R_{pol}$ and $C_{pol}$) represent the resistance and parallel constant-phase angle impedance of the tissue cell membrane. By revisiting Figure 2.1, it becomes apparent that at lower frequencies, the resistive components ($R_1 + R_2$) of the interface play a larger role in the overall impedance value than at higher frequencies ($R_1$).

A simplified model of the skin tissue impedance known as the three-component model (TCM) replaces the constant-phase angle impedance with a simple capacitor $C_1$ [63,67] (Fig. 2.3). While this circuit model does not entirely explain the skin tissue
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characteristics, it is a common model used to roughly represent the impedance of skin and estimate the frequency behavior of tissue. In this model, $C_1$ now represents the capacitance of the skin tissue.

![Figure 2.3: Simplified tissue model: three component model (TCM). [63,67]](image)

**2.1.2 Electrode-Skin Circuit Model**

Both of the above-discussed models exhibit impedance properties that suggest *two* cut-off frequencies. A more recent model proposed by Hewson et. al, however, is a simple parallel RC circuit shown in Fig. 2.4. In this case, the resistor and capacitor elements represent the interface resistance and interface capacitance between the electrode and the skin.

![Figure 2.4: Simplified model of electrode-skin interface proposed by Hewson et al. [69]](image)
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This model is based on the results of a study by Hewson et. al that suggested, with respect to sEMG applications and the electrode-skin interface, only one cut-off frequency is typically found within the frequency band of interest [69]. In an impedance study by Searle and Kirkup [63], the impedance locus was plotted for a pair of silver/silver-chloride (Ag/AgCl) electrodes on the skin. As can be observed, the $R_1$ value from the TCM skin model that is typically represented by the first x-intercept on the locus graph can be approximated to be zero (Fig. 2.5). With a zero resistance $R_1$, the resulting model is the simplified RC model comprising only $R_2$ and $C_1$ as shown in Figure 2.4.

Figure 2.5: Electrode-skin Impedance vs. frequency over time with pre-gelled Ag/AgCl electrodes. [63]

This is the simplest electrical model that is widely used to represent the electrode-skin interface. Together, the interface’s resistance and capacitance influence the overall electrode-skin interface impedance, and consequently, the signal and noise char-
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acteristics of the interface [69, 70]. More detailed and specific electrode models are discussed in section 2.3.5.

2.1.2.1 Half-Cell Potential

Another important element in the electrode-skin model is the half-cell potential as represented on the model in Figure 2.6 by $E_{hc}$. The half-cell potential stems from the method of charge transfer from the tissue to the electrode.

In order to record biopotentials from the surface of the skin, the electrode must transduce ionic body current into electrical current. There is no direct current passed from the body to the electrode. This charge transfer instead occurs as a result of reduction-oxidation (redox) reactions that take place at the interface between the electrode and an electrolyte. With dry electrodes, the interaction takes place between the electrode and the subject’s sweat. With wet electrodes using gel, this interaction takes place at the electrode-gel interface. [71–74]. A model of the current transfer at the interface is shown in Figure 2.7.

In this diagram (Fig. 2.7), the metallic atoms inherent to the electrode are denoted
Figure 2.7: Electrode-electrolyte interface: current crosses from left to right where C represents a metallic atom, and the electrolyte contains cations $C^+$ and anions $A^-$ [74].

by a C. When the electrode makes contact with the electrolyte, dissociation of the metallic atoms occurs at the surface of the electrode, releasing metallic cations $C^+$ in the electrolyte immediately adjacent to the electrode (Eq. 2.3). The electrons $e^-$ produced as a result of this reaction remain as the charge carriers in the electrode metal. Anions that are present in the solution due to ionic current can also be neutralized as they meet the cations at the interface, giving off free electrons to the electrode metal (Eq. 2.4).

$$C \leftrightarrow C^{n+} + ne^- \quad (2.3)$$

$$A^{m-} \leftrightarrow A + me^- \quad (2.4)$$

This process continues until equilibrium is reached, where then a double charge layer is formed between the metallic surface and the electrolyte. At equilibrium, this double layer affects the local concentration of the cations and anions in the electrolyte...
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and a net potential at the electrode surface is observed. This potential is known as the half-cell potential and is characteristic to the electrode material. As a standard convention, the half-cell potential of a particular electrode is measured against the hydrogen electrode under standard conditions [71, 72]. The hydrogen electrode half-cell potential is defined as being zero under these standard conditions. The half-cell potentials of common electrode materials are shown in Figure 2.8.

![Half-cell potentials for common electrode materials at 25°C](image)

**Figure 2.8**: Half-cell potentials for common electrode materials at 25°C [74].

The double charge layer interface can be modeled as a semi-permeable membrane where the half-cell potential is represented by the Nernst equation:

\[
E_0 = -\frac{RT}{nF} \ln \frac{a_1}{a_2}
\]

where \( R \) is the universal gas constant, \( T \) is the temperature in Kelvin, \( n \) is the valence electron number, \( F \) is Faraday’s constant and \( a_1 \) and \( a_2 \) are the activities of the ions.
2.1.3 Measuring Electrode-Skin Interface Impedance

Measuring the electrode-skin interface impedance is of interest when trying to compare across different designs or materials to achieve a desired interface impedance. This type of impedance analysis is typically done using a two-terminal or two-electrode setup. With this setup, the same electrodes are used for applying the current and sensing the voltage drop, so both the electrode and skin impedance are included in the results, effectively measuring the entire electrode-skin impedance.

For points of comparison, the effective impedance is sometimes reported at a set frequency within the sEMG frequency band (e.g. 150 Hz), the power line frequency, or a frequency that is much higher than the sEMG frequency upper-limit (e.g. 20 kHz). A frequency sweep resulting in the impedance locus graphs shown earlier in this section is also useful for comparison. A frequency sweep is not instantaneous, however and does take time depending on the number of frequencies, number of data data points, and the range of frequencies in the sweep. If a settling period is not allowed for the electrode-skin interface impedance to stabilize and/or instantaneous impedance values are of interest, then the time needed to complete the frequency sweep may skew the data [63]. Figure 2.9 shows how after only 90 seconds, the impedance at
an Ag/AgCl electrode interface actually shifts. Methods have been developed that instead use *one* compound signal made up of many signals of different frequencies that are related to each other in a way that is describable by the Fourier series. Using Fourier analysis on the resulting signal, the individual components can then be broken up, effectively performing multi-frequency analysis using one signal [63, 66]. An example of this compound signal is a square wave, which is effectively a wave composed of a multitude of sine waves of different frequencies [66].

### 2.2 Interface Characteristics

To obtain good quality sEMG with contact-based electrodes, it is often favorable to lower the overall impedance of the electrode-skin interface by lowering the interface
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resistance, increasing the interface capacitance or both. By lowering the interface impedance, more of the signal can be detected by the electrode while reducing noise, and the fidelity of the signal will be preserved. More details on the resistance and capacitance characteristics of the interface and methods to reduce their impedance are discussed in the following sections.

2.2.1 Interface Resistance

Overall, the electrode-skin interface is dominated by the properties of the outermost layer of skin, known as the stratum corneum [75]. The stratum corneum is highly keratinized and composed of dry, dead cells making it relatively non-conductive. The stratum corneum, while mostly non-conductive, can still allow some current to flow through. While the dry layer serves as an insulative boundary to the sEMG signal, there are still several direct “routes” in the layer through which the signal can travel. These “routes” are called appendageal pathways and are essentially ducts traversing the skin that allow for ionic current, and consequently EMG signal, to reach the skin’s surface (Fig. 2.10). Sweat glands are a common example of these appendageal pathways that, when present, will lower the resistance of the skin [76]. Hair follicles also create direct “routes” for ion transfer, however the hair itself serves as a hindrance to obtaining good quality sEMG because of the difficulty it creates in maintaining good mechanical contact with the skin. This bad contact decreases the capacitance of the interface by reducing the effective surface area of the contact (explained in
more detail in the following section 2.2.2) and contributes an increase to the interface impedance. These negative effects on the interface capacitance counterbalance the presence of an added pathway and instead actually increase the overall interface impedance. As such, hair is sometimes shaved off on the electrode site [75]. The density of sweat glands and other appendageal pathways highly influences the resistance of the layer. Since densities will vary from body site to body site as well as person to person, the resistance of the skin will undoubtedly vary depending on the individual and electrode location. Time and environment also have their influence. Sweat glands/ducts will activate or close in response to changes in the body or surrounding environment. Stimuli as a result of an increase in temperature or psychophysiological events such as stress or excitement can cause the activation of sweat glands and open...
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up more appendageal pathways, decreasing the skin resistance [75, 78]. This resistance is frequency-dependent as well. Higher frequency signals can travel through the stratum corneum layer more easily than lower frequency signals, where the signal transmission becomes more resistance-dominant.

Multiple issues will vary the skin resistance across different body sites and different people. The common goal for contact-based electrodes, however, is to lower this overall resistance. A variety of skin-preparation techniques have been developed for this purpose. One method of lowering resistance is the use of conductive gels between the skin site and electrode contact. Conductive gels are typically made using biocompatible salts such as sodium chloride (NaCl) and potassium chloride (KCl). The use of these ions which are already present in the conductive tissue beneath the outer skin layer allow for diffusion into the skin due to the induced concentration gradient. This renders the layer more conductive, opening additional pathways for current transfer [75]. The gels tend to be aggressive in the sense that in a long-term application, they may result in skin irritation. Biological tissue typically cannot tolerate ionic concentrations that are much higher than physiological levels and so these gels are usually used in short-term monitoring applications.

An alternative to the use of less viscous or “wet” gels is the use of solid hydrogels. These hydrogels incorporate conductive particles and serve as a conductive pad to ensure electrical contact between the skin and electrode. These solid gels are more resistive than the typical “wet” gels, however this increase in resistance can be coun-
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Figure 2.11: Skintact(R) pre-gelled disposable electrodes: “Aqua-Wet” electrode with liquid electrode gel (left) and “Aqua-Tac” electrode with solid electrode gel (right). [79]

...terbalanced by decreasing the thickness of the hydrogel pad used and increasing the overall surface area of the pad. These changes can help decrease hydrogel resistance as well as increase it’s capacitance, both factors which drive down the overall interface impedance. Most disposable electrodes will come with pre-gelled pads to make application of the wet electrode convenient (Fig. 2.11).

The use of gels is not always desired or practical, however, and depending on the application, the use of a dry electrode might be preferred. A short-term solution to lower the interface resistance for these dry electrodes is to first rub the electrode site with an aggressive gel such as those used by wet electrodes. Dry electrodes, however, are typically meant for more long-term applications, and using aggressive gels in a long-term setting will lead to skin irritation. Different skin-preparation techniques have been adopted for these electrodes. Since the outermost layers of the stratum corneum are the most resistive, oftentimes the skin-preparation process will involve scraping off a few of the top skin layers. This can be achieved by rubbing the skin using a mildly abrasive sandpaper-like material or even a fingernail. Sometimes
application and removal of a tape adhesive is used to strip off a few top layers of the stratum corneum. A rub down with alcohol wipes on the electrode site can also remove loose, dry cells and any poorly conducting lipid substances from the skin surface. Although the electrode-skin contact is technically “dry”, a thin layer of sweat will eventually form under the electrode contacts, naturally lowering the skin resistance at the interface [69, 80]. The time it takes to achieve this lowered, more consistent resistance is referred to as the settling period. Sometimes spraying the skin down with a saline spray or even tap water before electrode application can lower the initial resistance to a value that closely mimics the resistance achieved at the end of the settling period [81].

\subsection{2.2.2 Interface Capacitance}

The dry top layer of the skin also leads the skin to behave as a capacitor. One “plate” of the capacitor is the conductive tissue, the other “plate” is the electrode contact, and the dielectric is the stratum corneum. As such, the skin capacitance, unlike skin resistance, is not as affected by sweat and time-varying properties of the skin, but rather the electrode and the composition of the skin itself. By referring to the equation for capacitance (Eq. 2.6), it can be understood why this is so.

\[ C = \varepsilon_0 \varepsilon_r \frac{A}{d} \]  

The $\varepsilon_r$ value is based on the dielectric properties of the skin and may vary based
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on composition. The $d$ variable is related to the distance between the “plates”. In this sense, it is representative of the skin thickness. Consequently, areas on the body or individuals with thicker skin will exhibit lower skin capacitance than areas on the body or those with thinner skin. The $A$ variable is representative of the surface area of the “plates”. In this case, the surface area of the volume conductor side of the capacitor cannot be varied, however, the surface area of the electrode contact can. Electrodes with a larger overall surface area will exhibit larger capacitance at the electrode-skin interface [74, 81].

There are several ways to increase the overall electrode-skin interface capacitance. In terms of skin preparation, removing the outermost layers as described in section 2.2.1 can lead to a decrease in $d$ and also remove any dry, loose cells which will lead to better and larger overall contact area $A$. Gels or sprays can reduce the skin resistance, however they also work to increase the contact area by filling any gaps or crevices on the skin surface and facilitating better electrical contact between the skin and electrode. A rough or protruding electrode surface such as is seen with the conventional “metal dome” electrodes (Fig. 2.12) can also lead to an increase in capacitance by increasing the electrode contact area (increasing $A$) and pushing deeper into the residual limb (decreasing $d$) [74]. By increasing interface capacitance, the overall interface impedance favorably decreases.
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Figure 2.12: Liberating Technologies’ metal dome electrodes in three sizes: standard, small and high-dome. Small size is used typically only when there are limb size constraints and high-dome size is typically used for patients with thicker/tougher skin to reduce the interface impedance. [82]

2.3 sEMG Electrode Types

Surface EMG electrodes can be categorized based on their contact materials, interface characteristics, instrumentation properties and more. For the purpose of this section, the electrodes are categorized based on their contact interface to the skin. These categories are separated into: wet, dry and capacitive.

2.3.1 Wet

Surface EMG electrodes that are classified as “wet” electrodes use conductive gels for biopotential recordings. As discussed earlier, these conductive gels typically contain ions in order to lower the skin impedance and facilitate the signal transfer at the electrode-skin interface. The texture of the gel can range from a wet liquid-like material to a more solid hydrogel. Typically this type of electrode is used for short-term recordings, since the conductive gels can cause irritation at the skin site,
especially if they have a higher ionic concentration. The common wet electrode design is an adhesive-backed Ag/AgCl electrode contact that is pre-gelled. This gel can be soaked into a soft foam that sits over the Ag/AgCl contact, or it can be a more solid-type gel that is formed around the electrode contact [69,70,83]. “Snap” attachments are located on the back of these disposable electrodes to allow for mating cables to snap onto the electrodes for recording sEMG (or ECG). For sEMG purposes, the electrodes are sometimes designed to have a pair of contacts on the same adhesive patch to maintain inter-electrode distance and for easy application (Fig. 2.14).

Figure 2.13: Disposable pre-gelled surface biopotential electrodes with “wet” gel (left) and “solid” gel (right).

Figure 2.14: Examples of paired disposable EMG electrodes.
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2.3.2 Dry-Contact

Dry-contact or dry electrodes are termed so because of the fact that the contact interface is “dry”, meaning no gel is used and the electrode contact makes direct contact with the skin. Because there is no gel to help lower the impedance of the interface, these types of electrodes are more challenging in terms of acquiring good sEMG signals with low noise [70,80,84]. The implications of this high impedance interface are discussed in more detail later in this chapter. Stainless steel is a common metal used for dry electrodes due to its non-corrosive properties and stability, although theoretically, any conductive material can be used. Depending on the electrode design, the interface with the circuitry will be different. The dry electrodes can be used in a remote setting and an active setting. Figure 2.15 shows both. An active electrode such as is shown in Figures 2.15(a) and 2.15(b) houses the processing electronics and shielding directly on the back of the electrode. In Figure 2.15(c), remote metal dome electrodes are shown. These electrodes comprise just the electrode

Figure 2.15: Examples of dry-contact EMG electrodes: (a) Otto Bock active electrode (b) Touch Bionics, Inc. active electrode (c) MagnaSnap™ metal dome remote electrodes by LTI, Inc.
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contacts themselves. The particular electrodes shown have a magnetic backing, and couple to the processing electronics downstream from the contacts themselves.

Novel dry-type electrodes investigate the use of other conductive materials besides metal. Due to the fact that any conductive material is theoretically capable of acquiring sEMG (although maybe not very good quality sEMG), some have tried constructing wearable-friendly electrodes out of conductive textile or conductive foam [81, 85–88].

2.3.3 Capacitive

Capacitive electrodes record biosignals using capacitive coupling rather than the resistive pathway at the electrode-skin interface. With these electrodes, there is no interaction between the metal and the skin or sweat. As such, no half-cell potential can develop and the recording is purely capacitive [70, 89, 90]. These types of electrodes can be grouped into two main categories: insulated and non-contact. In the case of the insulated electrode design, there is an insulating dielectric layer between the electrode and the skin. It may seem that this would make the electrode a non-contact electrode, however in this case the electrode here is purposefully insulated with a known dielectric material, and this material still needs to make contact with the skin. In the case of a non-contact electrode, the electrode is designed to work over clothing or some other unknown material, and the electrode overall does not need any direct skin contact. The insulated electrode design attempts to remove the
variable effects of the skin on the overall signal acquisition, since now the electrode is capacitively coupled. With a suitable dielectric material and secure skin contact, the coupling capacitance can be large, allowing for good signal acquisition potentially without the negative effects of low-frequency noise [70].

Non-contact electrodes are meant to work without the need for any direct physical contact between the electrode and the skin. This is especially useful for situations where a measurement must be taken quickly over clothing [70]. These electrodes typically have larger electrode contacts for better capacitive coupling (Fig. 2.16).

Capacitive electrodes are different from the wet and dry contact-based electrodes in the sense that they usually need special instrumentation. With insulating electrodes where the dielectric touches the skin, standard amplifiers may be able to acquire good signals if the insulating layer is very thin. However for capacitive electrodes and especially non-contact types, an instrumentation amplifier with ultra-high input impedance (e.g. $10^{16}\Omega$, $10^{-17}F$) is required. Due to the high coupling capacitance (150-300 pF) typically required to dominate skin impedance, proper shielding and guarding as well as manual calibration of a stable bias network are needed to prevent any parasitic capacitance and to maintain good quality sEMG [89, 91, 92].
2.3.4 Other

In the research space, alternative and novel electrode designs have also been developed that may not fit very well into these categories. One example is the electrode design based on conductive polymer. In this case, a polymer such as silicone is permeated with conductive particles [74,93,94]. This electrode type is somewhat of a hybrid between wet and dry-contact. It is dry in the sense where there is no additional gel added over the electrode contact at the interface; however, it is wet in the sense that it is essentially employing a hydrogel as the electrode contact itself.

Another novel electrode design is the “tattoo” electrode developed by the J. Rogers group at University of Illinois [95]. This design employs silicon MOSFETs and other circuit elements made of silicon and gallium arsenide integrated onto a elastomeric sheet based on modified polyester. This sheet can then be laminated onto the skin much like a temporary tattoo (Fig. 2.17). While there is still work to be done to allow for this type of electrode tattoo to work in a long-term setting, an interface like this

![Figure 2.17: Novel multi-functional sensor “tattoo” interface (left) with close up of the EMG sensor (right).][95]
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could potentially be powered wirelessly and essentially remove the effects of motion artifact and unwanted cable noise. Furthermore, locations would remain secure and the electrode would be fully adaptable to any changes in limb volume or shape.

2.3.5 Electrode-Based Circuit Models

The simple parallel RC circuit model is a simplified version of the actual electrode interface, however it provides a basis for understanding the various factors that affect the interface [70, 74]. In Figure 2.18, the different types of electrodes just discussed are modeled. A general model is shown on the left with more specific models to the right that are based on electrode type. Depending on the materials used and overall electrode design, there may be one or multiple interfaces, termed “coupling layers”, that make up the entire model. Each coupling layer is represented by an additional RC element placed in series in the overall circuit model. The first element $R_{\text{body}}$ represents the resistance of the subcutaneous layer and deeper tissue underneath the skin surface. This resistance is typically very low ($< 500\Omega$) in comparison to the impedance of the stratum corneum [96]. The next element is an RC parallel circuit.

This first RC circuit symbolized by $R_1$ and $C_1$ represents the first coupling layer of the electrode interface at the skin surface.

For wet Ag/AgCl electrodes, this RC element is the coupling layer between the volume conductor and the electrolyte gel, with the skin sandwiched in the middle as the dielectric. The second RC element in the wet electrode model is the gel-electrode
interface, represented by the coupling layer between the skin surface and the “top” of the electrolyte gel layer right beneath Ag/AgCl electrode itself. Finally, the last element is the voltage source $V_{hc}$, which represents the half-cell potential resulting from the contact between the gel electrolyte and the Ag/AgCl metal. For dry-contact electrodes, there is only one RC parallel circuit coupling layer in the model. This layer is representative of the electrode-skin interface. Since there is no gel between the skin and the electrode contact, the only other element in series with the RC parallel circuit is the half-cell potential resulting from the contact between the skin and the metal.

The third model characterizes an insulated electrode. The interface between the volume conductor and the insulating material is represented by the first RC parallel circuit. Since the electrode itself is insulated and there is no direct conductive pathway between the electrode and the skin, the following (and last) element modeled for this electrode type is a capacitor. The capacitor is representative of the gap created by
the thickness of the insulating material between the skin surface and the electrode contact. The last electrode model illustrates the more complex non-contact electrode interface. This particular model is assuming that the non-contact electrode is being used through cotton fabric. The first RC element is the coupling layer between the volume conductor and the cotton and the second RC element illustrates the coupling layer between the skin and the “top” layer of the cotton. The third and last element in this model is a capacitor which represents any air gap that may be present between the cotton fabric and the electrode contact.

It must be noted, however, that these are simplified models of the actual interface, and while they can help with predicting the characteristics of each electrode-skin interface type, they might not be entirely representative of the interface.

2.4 Noise Analysis

2.4.1 Noise Sources

Noise is always a challenge for surface biopotential recordings. There are many different noise sources that can contribute to the overall noise of the system. Electrical components have their own inherent noise, while environmental and situational factors can also inject noise into the system. This makes obtaining a clear sEMG signal a challenge.
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2.4.1.1 Electrode Material

A possible noise source for sEMG can be the material chosen for the electrode. An example of a noise source is the use of mismatched material types to record a signal. Because the half-cell potentials of different materials are not equivalent, it is imperative that the electrode material is used when taking a differential measurement using a pair of electrodes. For example, if a silver electrode and gold electrode were used to take an sEMG recording, the difference in half-cell potentials would create a voltage offset that might saturate the amplifier. As such, any metals or materials that come into contact with the electrolyte on a pair of electrodes should be the same.

The electrode material chosen can also have inherent properties that affect the overall quality of an sEMG recording, namely the material’s impedance and polarizability. A lower impedance is typically favored when trying to record good quality signals. A lower impedance electrode interface will be less susceptible to noise and interference, helping maintain the integrity of the original sEMG signal. More on the implications of a lower versus higher electrode impedance and how it affects noise rejection are discussed throughout following sections.

With respect to the matter of polarizability, however, the implications are different depending on the measurement application. To understand how polarizability can affect the signal, the reactions at the electrode-electrolyte interface must be revisited. In the case of a “wet” electrode, the electrolyte is represented by the conductive gel at the interface. In the case of a “dry” electrode, bodily fluids (sweat) at the in-
interface surface represent the electrolyte. Regardless of the electrolyte, each material will have its own half-cell potential value as discussed earlier. This half-cell potential is representative of the condition where no electric current exists between the electrode and electrolyte. If current is present, however, the half-cell potential $E_0$ is altered; the difference between the equilibrium half-cell potential and the altered potential is termed the overpotential. Overpotentials can be classified as three types: ohmic overpotential, concentration overpotential and activation overpotential. Ohmic overpotentials ($V_r$) are the direct result of a change in resistance of the electrolyte. Concentration overpotentials ($V_c$) are a result of a change in concentration distribution in the vicinity of the electrode. Activation overpotentials ($V_a$) are attributed to the energy barrier of the redox reactions related to the direction of the current flow. These three mechanisms are additive as shown in Equation 2.7 and alter the half-cell potential as shown in Equation 2.8, where the $a$’s represent the activities of all the ions involved in the reaction.

$$E = E_0 + V_r + V_c + V_a$$  \hspace{1cm} (2.7)

$$E = E_0 + \frac{RT}{nF} \ln \left( \frac{a_C^{\delta} a_D^{\gamma}}{a_A^{\alpha} a_B^{\beta}} \right)$$  \hspace{1cm} (2.8)

An electrode is more polarizable if it is more susceptible to the development of overpotentials. Theoretically, electrodes can be classified as perfectly polarizable or perfectly nonpolarizable. Electrodes that are perfectly polarizable transfer current by changing the charge distribution within an electrolyte solution near the electrode surface. As such, no products are formed at the electrode-electrolyte interface and
the electrode is capacitive in nature. Noble metals such as platinum, silver, gold and palladium and inert metals such as stainless steel are examples of highly polarizable metals. For stimulating electrodes, these are typically the metal types of interest, since the chance of producing a chemical reaction at the interface is minimal.

Perfectly nonpolarizable electrodes allow current to travel across the interface freely with no change in the charge distribution in the electrolyte solution near the electrode. No energy is required and overpotentials are not seen; these electrodes are more ohmic in nature. Examples of electrodes that exhibit this nonpolarizable behavior are silver/silver-chloride (Ag/AgCl) and calomel (Hg₂Cl₂) electrodes.

Referring back to our simplified electrode-skin RC model, a polarizable electrode material will create an interface that is more capacitively dominant whereas a more nonpolarizable electrode material will create an interface that is more resistively dominant (Fig. 2.19).

![Fig. 2.19: Dominant elements in terms of polarizability.](image)

In reality however, electrodes cannot be completely perfect in either direction and will exhibit some degree of polarizability. Electrodes that are more polarizable are more susceptible to motion artifacts and as such, electrodes that exhibit nonpolariz-
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able characteristics are preferred for biopotential recordings. This is because with a highly polarizable electrode, when the electrode moves with respect to the electrolyte, the change in charge distribution at the surface will easily induce a concentration over-potential and the change in potential will show up on the sEMG recording as motion artifact. Nonpolarizable electrodes are more ohmic in nature and because the current can freely move through, these overpotentials are not observed.

The silver/silver-chloride (Ag/AgCl) electrode with conductive gel is commonly used for biopotential recordings since it can achieve its half-cell potential equilibrium quickly and in a stable fashion. Ag/AgCl electrodes are silver (Ag) electrodes quasi-coated and infiltrated with silver chloride (AgCl). Due to the fact that Cl$^-$ ions are used in conductive gel, the reduction of Ag$^+$ cations present at the electrode-electrolyte interface will neutralize back into AgCl. Since AgCl is highly insoluble, the AgCl deposits back onto the electrode keeping the interface stable. Due to the wide availability of the chloride anions, charge transfer at the surface can continuously occur, and the electrode approaches a nonpolarizable electrode.

In a long-term recording application, however, as such is needed with the use of myoelectric prostheses, the use of conductive gel after a few hours can result in skin irritation. The use of these gels is not recommended for over 24 hours, and the gel itself tends to dry out by then anyways. Furthermore, if the electrode is to make contact with the skin in a long-term setting, materials are typically chosen that will minimize the chance of any chemical reactions with perspiration. As such, more inert
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materials such as stainless steel, silver or gold-plated disks are used. As mentioned
before, these inert materials are more polarizable, making them more susceptible to
motion artifact. These materials also exhibit higher impedance at lower frequencies
in comparison to the Ag/AgCl electrode, making them more susceptible to noise in
the motion artifact-low frequency range [71,74,84]. Unfortunately, this trade-off is an
issue that cannot be entirely avoided when dealing with measurements in a long-term
setting. As such, measures must be taken to secure the electrode onto the skin and
prevent any movement of the electrode with respect to the skin. More on motion
artifact, how it is generated and its implications are discussed later in this chapter.

2.4.1.2 The Amplifier

Fluctuations in electric charge exist in all electrical components, creating random
variations in the potential between the ends of any conductor. These fluctuations are
what we refer to as the “noise” of a component. The common noise types inherent
in electrical components are shot noise, thermal noise, flicker noise and burst noise.
Shot noise is associated with current flow. It is manifested as a large number of
random independent current pulses that create a noise profile which is spectrally flat.
This is classified as “white noise” and means that the noise is frequency-independent.
Thermal noise is caused by agitation of charge carriers (electrons and holes) in a
conductor and is due to changes in temperature. This noise is also spectrally flat
and is inherent to any passive resistive elements. Flicker noise is also known as 1/f
noise and is proportional to the DC current running through a component. As such, in a very low-current device, flicker noise is negligible in comparison to the device’s inherent thermal noise. Unlike shot and thermal noise, flicker noise is frequency-dependent. Burst noise is also known as popcorn noise and is related to imperfections in the semiconductor material or heavy ion implants [97].

In a biopotential recording system, these inherent noise sources must be minimized. A key component in the circuitry attached to the electrodes is the amplifier. For the purposes of detecting sEMG, a low-noise amplifier is crucial [70, 84, 97]. Since the sEMG signal is already very small, any disturbances in the signal introduced by the circuitry noise could distort it. Inherent noise levels of an amplifier are typically expressed by their spectral or power density, reported as voltages or currents per root Hertz ($V/\sqrt{Hz}$ or $A/\sqrt{Hz}$). For white noise sources, the voltage RMS value is a parameter of interest. These noise sources have a Gaussian distribution where the average mean-square variation is the variance $\delta^2$ and the $V_{RMS}$ value is the standard deviation $\delta$. As such, the $V_{RMS}/\sqrt{Hz}$ frequency spectrum is also of interest. The majority of the signal is within $\pm 3\delta$, so the RMS value ($\delta$) x 6 is reported as the peak-to-peak ($V_{p-p}$) noise level. Flicker or 1/f noise, as suggested by its name, is a straight line with a constant, negative slope when plotted against frequency. The power density of flicker noise is in terms of $V^2/\sqrt{Hz}$ with the square root of that equivalent to $V_{RMS}/\sqrt{Hz}$. “Input-referred noise” is the term used to describe the internal noise sources of an operational amplifier, and comprises both an input-referred
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voltage noise component and an input-referred current noise component; these internal sources are treated as uncorrelated, independent random noise generators in series or parallel with the inputs of an ideal, noiseless amplifier [98]. Input-referred noise is typically modeled as a voltage noise source in series with the positive input of an amplifier in conjunction with two current noise sources at both inputs of the amplifier (Figure 2.20).

Figure 2.20: Amplifier input referred noise model; en: input referred voltage noise, inn/inp: input referred current noise. [97]

The total input-referred noise contains both white noise and 1/f noise [97–99]. Because these are treated as independent noise sources \( n \), they can be added where:

\[
\delta_{Total}^2 = \delta_{n1}^2 + \delta_{n2}^2
\]

(2.9)

and

\[
\delta_{TotalRMS} = \sqrt{\delta_{n1}^2 + \delta_{n2}^2}.
\]

(2.10)

Due to the nature of this relationship, the worst noise source will tend to dominate.
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At lower frequencies, 1/f noise dominates, whereas in higher frequencies, white noise dominates. The noise corner frequency $f_{nc}$ is the point on the $V_{RMS}/\sqrt{Hz}$ frequency spectrum where the 1/f noise level and white noise level are equivalent. At the noise corner frequency, the actual sum of the noise sources is 3dB higher than the 1/f noise alone [99]. A typical graph showing the separate noise sources and their combined effect is shown in Figure 2.21. Typically on amplifier datasheets, the inherent noise level at $f_{nc}$ is reported. Both the input-referred voltage noise level and input-referred current noise level are usually reported, however for a high-impedance electrode interface, it is the input-referred current noise which plays a larger role in the overall noise interference. A high-impedance electrode interface is more sensitive to input-referred current noise, since larger voltage errors result from the noise current passing through the higher impedance. As such, an amplifier with very low input-referred

Figure 2.21: Typical noise graph from an amplifier datasheet; white noise, $1/f$ noise and their combined noise voltage effect are plotted. [97]
current noise is important for dry contact electrodes.

Another source of noise in terms of signal distortion is the amplifier input bias current. While in an ideal system no current flows into the inputs of an amplifier, in reality there are input bias currents which will flow through the external impedances into the amplifier input [84,100]. If the source impedances are very high, which can be the case for dry electrodes, the input bias current will result in an added voltage error. Dry electrodes can have an initial electrode-skin impedance of 1MΩ or greater. With a source impedance of 1MΩ, an amplifier with an input bias current of 10nA would introduce 10mV of error into the signal. As such, an amplifier with low input bias current is also desired.

A very important characteristic in terms of amplifier choice is also the amplifier’s common mode rejection ratio (CMRR). Due to the fact that the sEMG signal is a time-varying signal, it will show up on the amplifier inputs as a differential signal and will be amplified. The common mode voltage is the potential of the skin or body that is common to both amplifier inputs. As such, unwanted noise that affects both amplifier inputs such as power line interference (more in section 2.4.1.5) would theoretically show up as common mode voltage and would be removed. The degree to which the amplifier rejects this common mode voltage is known as CMRR. For biopotential recordings, a high CMRR is important [83,101].

A high input impedance at the amplifier input is also crucial for rejecting noise. As discussed in the previous sections, the electrode-skin interface has its own impedance.
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As such, the electrical representation at the amplifier input can be modeled as a voltage divider created by the electrode-skin interface impedance and the amplifier’s input impedance (Fig. 2.22) [101]. The amplifier input impedance thus must be significantly higher than the interface impedance to avoid a voltage drop across the inputs that would lead to signal attenuation. A high input impedance amplifier is especially important for high-impedance type electrodes, such as dry electrodes [83]. Additionally, in an ideal setting, the electrode-skin impedance of the electrodes used in the differential sEMG measurement would be identical. In a more realistic, non-ideal situation however, these impedances are mismatched. This can be due to the electrodes, the electrode cables, the surroundings and/or the skin. For example, faulty electrode design where the size or shape of the recording electrodes aren’t the same can result in impedance mismatch. Differences in the skin properties (moisture, thickness) at the

![Figure 2.22: Voltage divider effect at amplifier inputs; where $v_{in(+)} = Z_{ia}/(Z_{ia} + Z_{ea})$ and $v_{in(-)} = Z_{ib}/(Z_{ib} + Z_{eb})$. If $Z_{ia}, Z_{ib} \gg Z_{ea}, Z_{eb}$, the voltage divider effect becomes negligible.](image)
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electrode locations can result in it as well. Impedance mismatch due to cables and electrical interference is explained in more detail in section 2.4.1.4. If the impedance mismatch is large enough, the voltage divider effect at the amplifier’s inputs would now also be mismatched, transforming a common mode signal into a differential signal that can no longer be rejected by the amplifier [80]. The mismatch effectively reduces the overall CMRR of the amplifier [102]. If the amplifier input impedance is significantly larger than the electrode-skin interface impedance, however, the potential differences at the amplifier inputs arising from the mismatched impedance will be minimal, and the common mode signal will still be attenuated [83].

The analog gain of the amplifier is also important. Surface EMG signals are very small and on the mV scale, and as such, need to be amplified above the noise floor. Amplification can be done in an analog or digital fashion; in both cases, the resolution is of concern. On the analog side, the amplification must be sufficient enough or adjustable, such that minimal digital amplification is required. The output voltage resulting from a fully-analog filtered envelope signal can be used directly to activate amplitude-based hands. Often enough, however, the signals are first digitally processed before driving a device. As such, a front-end amplifier will typically have an analog-to-digital converter (ADC) on board. Individual ADCs are also possible based on the electrode design. In either case, when converting the signal, the resolution of the built-in or stand-alone ADC must be high enough (12-24 bit) such that the resulting digital signal is as smooth and continuous as possible, closely resembling the
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Figure 2.23: Low magnitude sEMG envelope signal appears discrete due to resolution limitations of the ADC being used; if amplified, overall magnitude would increase, but discrete-like appearance would remain.

original analog signal. With a low-resolution ADC, the resulting signal may look very “step-like” and discrete (Fig. 2.23). Thus on the digital side, when implementing a digital gain to a signal, the resolution of the signal remains the same while individual values of each data point are multiplied. With a high-resolution ADC, the digitally-amplified signal may still somewhat look continuous. Again, however, if the digital gain is too high or the ADC resolution is too low, the signal will look more discrete. The optimal solution is to implement a sufficient analog gain at the source/amplifier while also using a high-resolution ADC prior to any digital processing.

2.4.1.3 Motion Artifact

Another major issue with surface biopotential recordings is the presence of motion artifact. This type of noise can be generated in a variety of different ways. At the electrode interface level, the motion artifact that results is sometimes also referred to as movement artifact. When the electrode moves with respect to the skin, unwanted
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signals are created. This movement can occur due to force impulses delivered to the interface as a result of hitting an object or even due to something as ordinary as walking [103, 104]. Transversal motion of the electrode occurs when the degree of contact between the electrode and skin is disturbed or altered, leading to instantaneous changes in the electrode-skin interface impedance that distort the signal. An example of this case is skin stretch beneath the electrode. Stretching is thought to spread the outer layer of the skin, exposing some of the inner layer skin cells and ultimately altering the conductivity of the skin interface by changing the availability of extracellular channels. This effect results in a change in interface impedance and can induce offset potentials of up to 10mV, adding noise to the sEMG signal [74,105]. An extreme case of transversal motion is electrode lift-off, where contact between the electrode and skin is significantly compromised. This results in the inability to record accurate sEMG signals. Lateral motion artifact occurs when the electrode maintains contact with the skin, but shifts or slides across it. This can possibly generate electric charges due to friction or a net current due to the polarization properties of the electrode material [70,74].

The effects of motion artifact are enhanced when using highly polarizable electrodes. As mentioned earlier, in reality a perfectly polarizable or nonpolarizable electrode is not possible, and an electrode depending on its material properties will exhibit a degree of polarizability. When considering taking biopotential measurements of an active subject, an electrode material that is more nonpolarizable is preferred,
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since it is less susceptible to motion artifact [74].

Another source of motion artifact is more intrinsic to the body and is induced by the actual movement of the muscle underneath the skin. During the flexion or extension of a muscle, the muscle spindles stretch and relax to contract the muscle. The motor unit action potentials (MUAPs) are responsible for creating the sEMG signal whereas the movement of the muscle can result in signals not related to the actual firing of the MUAPs. This movement is usually manifested on the resulting sEMG recording as an underlying low-frequency signal [103,104].

In general, motion artifact interference signals are typically in the low-frequency range up to 20 Hz [103]. This poses a problem more specifically for low-frequency biopotential signals such as ECG. Since ECG signals are also low-frequency, filtering techniques cannot be used to high-pass filter out the artifact noise. For sEMG, however, a high-pass filter can be used to minimize the effects, and depending on the filter type and order, this filtering can be very effective. Effects of this filtering and increasing cut-off frequencies $f_c$ can be seen in Fig. 2.24. The high-pass filter is typically implemented with a cutoff frequency $f_c$ between 10-20 Hz.

In a practical sense, however, motion artifact still remains a common source of interference, especially if the electrodes are used in an active setting. To help minimize this artifact, electrodes of different types will use different techniques. Disposable electrodes are adhesive-backed for the purpose of adhering them to the skin and keeping them in place (Fig. 2.25(a)). Active electrodes developed by Otto Bock and

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Figure 2.24: Effects of an implemented digital high-pass filter; a low-frequency motion artifact can be observed during contraction of the arm flexors in the first and second graph. As the $f_c$ increases, this low-frequency artifact is ultimately rejected.

Touch Bionsics, Inc. have relief areas built in around electrode contact protrusions to essentially “bury” the electrodes in the residual limb and minimize movement with respect to the skin (Fig. 2.25(b)). For electrodes embedded in wearable electronics, compression is used to attempt to hold the sensors in place. This is achieved by using tight adjustable straps or compression shirts and sleeves such as those typically seen in sportswear to maintain the electrode-skin contact (Fig. 2.25(c)). Inevitably, however, motion will appear on the sEMG signal if the electrode isn’t held tightly enough against the skin and/or if proper high-pass filtering parameters are not in
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place.

Figure 2.25: Examples of design considerations taken to minimize motion artifact: (a) adhesive backing on disposable electrode (b) relief areas and protruding contacts in active Touch Bionics electrode and (c) compression shirt with wearable sensors by Under Armour, Inc (Baltimore, MD).

2.4.1.4 Cable Artifact

Cable artifact is also another source of noise in sEMG recordings. When modeling the electrical representation of an sEMG recording setup, it becomes apparent that every aspect of the system is capacitively coupled to its surroundings. The body, the electrode cables, and the amplifier are all coupled capacitively to surrounding environmental noise and/or the earth ground. With the air serving as a dielectric, noise injects itself into the system via this capacitive coupling and is a large source of interference [83,101]. As unshielded cables move through changing electric fields, displacement currents are generated in the cables that show up as voltage differences at the amplifier inputs. Since the effective voltage difference is the product of the displacement current and the source input impedance, high-impedance electrodes are
Figure 2.26: Interference caused by an induced current on the electrode cable manifests itself as an artificial sEMG signal while recording with a high impedance dry electrode.

more susceptible to this noise and create larger voltage errors at the amplifier inputs [23]. Figure 2.26 shows the result of interference induced on the sEMG recording by bringing near and touching an unshielded cable to a piece of silicone. Since the silicone is in insulator, it possesses a static electric field that in-turn induces a displacement current on the cable and manifests itself as an “artificial sEMG” signal. The particular electrode used in this example was a dry-type high impedance electrode, and as such, the resulting voltage error was large.

In an attempt to reduce the effects of cable artifact, some may use passively or actively shielded cable (more in section 2.4.3.1). Shielded cables can contribute their own artifacts to the system, however. Movement of the cables can lead to deformation in the shield and cable insulation. This deformation can lead to friction between the layers of cable insulation, generating small static charges termed tribo-electrical
discharges that dissipate throughout the measurement system [23]. This is especially an issue when using loosely braided wire for shielding, since the braid can deform to allow for changes in contact between the cable insulation.

Magnetic coupling interference is also manifested as voltage differences at the inputs and originates from fluorescent lights or electric appliances [83, 101]. If unshielded or grounded cable loops are formed between the subject, the cables and the amplifier, this effectively creates a single-turn coil. As described by Faraday’s Law, if a magnetic field passes through a coil, a voltage is induced in this coil, which in this case would be the recording cables. According to Faraday’s Law that induced voltage $V$ is equivalent to:

$$V = -N \frac{\Delta (BA)}{\Delta (t)}$$  \hspace{1cm} (2.11)

where $N$ is the number of turns on the coil, $B$ represents the magnetic field and $A$ is the area of this loop. As such, in the recording setup this induced voltage is dependent on the magnetic field strength and the area of the loop. To reduce this interference, the differential pair of cables are typically twisted together, reducing the overall area of the loop, and consequently, the degree of magnetic interference. With buffered signal outputs or cables, magnetic interference becomes much less of an issue [23, 101, 106].
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2.4.1.5 Power Line Interference

Power line noise can interfere with surface biopotential recordings and can result by simply being near everyday electrical equipment. Power line interference can be encountered as both magnetic interference and electrical interference. Magnetic interference is the result of loops formed by the cables, subject, and amplifier as discussed in the previous section. Electrical interference from power lines show up on the amplifier as unwanted 50/60 Hz signals. Ideally, these signals would appear on the amplifier inputs as a common mode signal. As discussed before, with a high enough CMRR, this common mode signal would be rejected and not included in the differential signal. Due to impedance mismatch in an actual non-ideal setting, however, the signal does not appear entirely on the amplifier inputs as a common mode voltage and the differential signal is amplified [80]. The power line noise is also a much stronger signal in comparison to sEMG; as such, if the subject is close enough to a piece of electrical equipment, without a high enough CMRR, the interference signal will dominate the recording and mask the sEMG signal. Figure 2.27 shows how the power line noise can infiltrate the sEMG recording. In this case, the subject held the cable of a laptop charger to induce the power liner interference. Methods of reducing this type of noise are discussed in a later section (section 2.4.3).
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Figure 2.27: 60 Hz interference observed on an sEMG recording when the subject touches the cable of a laptop charger.

2.4.1.6 Unwanted Biopotential Signals

When recording sEMG, other biopotential signals may contaminate the recording. The ECG signal can show up on and sEMG recording depending on where the electrodes are placed. For example, when recording sEMG on the trunk or the neck, the ECG signal is usually apparent if a high-pass filter to remove lower-frequency signals is not implemented [23]. Similarly, sEMG signals that are not of interest can also, in a sense, contaminate the recording. As mentioned in the previous chapter, the interstitial fluid and fatty tissue underneath the skin act as a volume conductor. Consequently, signals originating from muscles that are not directly under the recording site can still be obtained at the site. This is termed cross-talk, and the resulting recording is an sEMG signal comprising the muscle of interest as well as other un-targeted muscles that are not of interest [107]. As can be observed in Figure 2.28,
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during contraction of the forearm extensor muscles, the sEMG from these muscles can also be observed on the “flex” channel, which is the electrode placed above the forearm flexors (and vice versa). Muscle tension can also contribute to the overall baseline “noise” level of the recorded signal. If the subject is unable to fully relax his or her muscles, the residual signal due to incomplete muscle relaxation can falsely elevate the apparent noise floor of the sEMG recording [23].

Figure 2.28: Graph showing envelope sEMG from two electrodes placed on the forearm flexors and forearm extensors; during flexion and extension of the forearm muscles, cross-talk is observed on both the “flex” and “extend” channels.

2.4.2 Amplifier Circuit Noise Model

To better understand how different aspects of the measuring system affect the signal, a good method is to analyze an electrical model of the source input stage between the electrode on the skin and the amplifier input. Figure 2.29 represents an amplifier circuit noise model presented by Chi et al. [70] for a dry electrode interface. In this model, each electrode is connected to its own operational amplifier, with the output of the signal used to drive an active shield that guards the input.
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Figure 2.29: Dry amplifier circuit noise model. [70]

This model takes into consideration the signal source voltage, the coupling (interface) impedance, the amplifier’s input impedance, an active shield and the inherent noise of the amplifier. Because capacitive coupling of cables are not taken into account, this model more accurately represents an “active” electrode, where the amplifier is placed directly on the back or in very close proximity of the electrode and consequently the signal source itself. A remote electrode interface where the amplification is done downstream from the signal source is better represented in the next section (2.4.3).

The parameters in the circuit are defined as follows:

\[ v_s(j\omega) \] signal source on the skin surface;

\[ v_o(j\omega) \] output signal of the amplifier;

\[ v_i(j\omega) \] voltage at the amplifier input;

\[ v_{i,n}(j\omega) \] input-referred amplifier voltage noise;

\[ i_{i,n}(j\omega) \] net current noise at amplifier input;

\[ Z_c(j\omega) \] electrode-skin coupling impedance;
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\( Z_i(j\omega) \) amplifier input impedance;

\( C_s \) active shield to electrode capacitance; and

\( A_v \) amplifier voltage gain.

The transfer function can be solved for by performing KCL node analysis at the input of the amplifier shown in Figure 2.29, where:

\[
\frac{v_o - v_i}{1/j\omega C} + \frac{v_s - v_i}{Z_c} - \frac{v_i}{Z_i} + i_{i,n} = 0 \quad (2.12)
\]

and

\[
v_o = A_v(v_i + v_{i,n}). \quad (2.13)
\]

The inverse of impedance \( Z(j\omega) \) is admittance \( Y(j\omega) \) and is represented by both the conductance and capacitance components, \( g \) and \( j\omega C \), respectively. As such, the admittance components are represented by:

\[
Y_c(j\omega) = g_c + j\omega C_c, \text{ electrode-skin coupling admittance;}
\]

\[
Y_i(j\omega) = g_i + j\omega C_i, \text{ amplifier input admittance.}
\]

Using elimination of \( v_i \) in equations 2.12 and 2.13 and substituting admittance \( Y(j\omega) \) in for impedance \( Z(j\omega) \) where appropriate, the equation becomes:

\[
(Y_c + Y_i + j\omega C_s - A_v j\omega C)v_o =
A_v(Y_c v_s + (Y_c + Y_i + j\omega C_s)v_{i,n} + i_{i,n}) \quad (2.14)
\]
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The output signal transfer function for $v_o$ can then be rewritten as a function of the source signal $v_s$ and the source input referred voltage noise $v_{s,n}$:

$$v_o = G(j\omega)(v_s + v_{s,n})$$  \hspace{1cm} (2.15)

with a gain of

$$G(j\omega) = A_v \frac{Y_c(j\omega)}{Y_c(j\omega) + Y_i(j\omega) + j\omega(1 - A_v)C_s}$$  \hspace{1cm} (2.16)

and a source input-referred voltage noise of

$$v_{s,n} = \frac{Y_c(j\omega) + Y_i(j\omega) + j\omega C_s}{Y_c(j\omega)} v_{i,n}$$  \hspace{1cm} (2.17)

$$+ \frac{i_{i,n}}{Y_c(j\omega)}. \hspace{1cm} (2.18)$$

As can be seen by the combined Equation 2.17 and 2.18, the source input-referred noise comprises both a voltage component (Eq. 2.17) and a current component (Eq. 2.18). To interpret what factors affect the source input-referred noise, the noise power density ($v_{s,n,rms}^2$) must be taken into consideration. The total power of the noise signal is equivalent to the sum of the squared RMS of each noise component. Substituting the admittance conductance and capacitance components, the power of the noise signal becomes:

$$v_{s,n,rms}^2 = \frac{(g_c + g_i)^2 + \omega^2(C_c + C_i + C_s)^2}{g_c^2 + \omega^2 C_c^2} v_{i,n,rms}^2$$  \hspace{1cm} (2.19)

$$+ \frac{i_{i,n,rms}^2}{g_c^2 + \omega^2 C_c^2}. \hspace{1cm} (2.20)$$

Using the final power density equations (2.19 + 2.20) derived from the amplifier circuit noise model, it can be observed how the characteristics of the coupling interface as
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well as the amplifier inputs affect the total source input-referred noise of the sEMG measurement. The next section (2.4.2.1) will investigate how to reduce the total system noise based on this model.

2.4.2.1 Model-Based Noise Reduction

As mentioned before, an amplifier with a high input impedance is crucial for noise reduction and improved signal fidelity. In the presented model, this parameter is represented by a high $Z_i$, which translates into low input conductance $g_i$ and small input capacitance $C_i$. The noise components will be analyzed with the assumption that a high input impedance amplifier is being used as part of the sEMG measurement system.

The first noise component (Eq. 2.19) of the power density equation is proportional to the amplifier’s input-referred voltage noise $v_{i,n,rms}^2$. For low-impedance type electrodes, this noise component reduces to the voltage noise floor of the amplifier, since the coupling conductance approaches infinity. For high-impedance electrode interfaces however, such as is the case for dry electrodes, the voltage noise floor is amplified by a factor of

$$1 + \frac{(C_i + C_s)}{C_c}. \quad (2.21)$$

To minimize the noise floor amplification, this coefficient (Eq. 2.21) must be minimized. It becomes apparent that in order to to this, the input capacitance $C_i$ must be minimized and the coupling capacitance $C_c$ must be maximized. The input ca-
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Capacitance can be minimized by careful amplifier choice with a high input impedance as discussed in section 2.4.1.2, and the capacitance can be maximized by optimizing the electrode design as described in section 2.2.2. These statements also agree with the previous discussion in section 1.2 which mentions that lowering the impedance of the electrode-skin interface can help with acquiring clearer sEMG signals. Minimizing the capacitance of the active shield can also help in reducing the overall noise amplification.

The second noise component (Eq. 2.20) is proportional to the net current noise that interferes with the system, and is the more dominant noise component, especially in high-impedance electrode interfaces. This net current noise combines contributions from the thermal noise of the coupling and amplifier interface as well as the amplifier’s own inherent current noise. Assuming that external current noise is not an issue and an ideal “noise-less” infinite input impedance amplifier is used, the power density equation reduces to:

\[ v_{s,n,rms}^2 = \frac{4kT}{g_c} + \frac{\omega^2 \xi^2}{g_c} \cdot (2.22) \]

To minimize this noise component, coupling conductance must be maximized; this observation again agrees with the statement that a lower impedance electrode-skin interface improves the integrity of the acquired sEMG signal. This is true for a contact-based electrode interface, which for the purposes of this thesis, is the interface of interest. In special non-contact electrode cases however, paradoxically maximizing coupling impedance will also reduce this noise component. This is only applicable for
values of $R_c$ larger than $\frac{1}{\omega C_c}$, so for increasing electrode distances, the resistance value needed to meet this constraint becomes exceedingly high. As such, even with a non-contact type electrode, close proximity to the skin is desired and coupling capacitance should be maximized.

The effects of motion artifact can also be somewhat represented by this amplifier circuit noise model. As discussed previously in section 2.4.1.3, there can be both transversal and lateral motion artifact when dealing with the electrode moving with respect to the skin. Transversal motion manifests itself as instantaneous changes in the coupling admittance ($g_c + j\omega C_c$). If the gain equation is revisited (Eq. 2.16), it is observed that these changes can be effectively “nulled” out if these two conditions are met:

$$g_i + C_i = 0 \quad (2.23)$$

$$1 - A_v = 0. \quad (2.24)$$

The first condition (Eq. 2.23) can be implemented by using a high input impedance amplifier, where the input conductance approaches zero and the input capacitance is minimized. The second condition (Eq. 2.24) can be met if the shield is driven by the output of the signal with the amplifier configured as a unity-gain buffer ($A_v = 1$).

Of course, the sEMG signal is very small and needs to eventually be amplified, but this can be done downstream in the measurement system. Lateral motion artifact, where friction is created between the electrode and skin surface, can be best modeled as an added noise current (Eq. 2.25). This added noise source can be minimized by
lowering the coupling impedance (high $g_c$, high $C_c$).

$$v_{s,n} = \frac{i_{i,n}}{g_c + j\omega C_c}. \quad (2.25)$$

Of course, motion artifact is also best reduced by having a secure electrode-skin interface, such that the above-discussed changes in gain and/or added noise currents do not manifest themselves in the measurement system.

In summary, based on this active electrode amplifier circuit noise model, the factors which reduce the overall noise in the measurement system are: a low electrode-skin coupling impedance, a high amplifier input impedance and a low-noise amplifier configured with unity-gain.

### 2.4.3 Interference Model

The previous amplifier circuit noise model (Fig. 2.29) is representative of an active electrode model. As such, external noise interference is not really taken into consideration. As mentioned in section 2.4.1.4, the entire measurement system including the body itself is capacitively coupled to the surrounding power line or mains noise and the earth ground (GND). This can be better represented by the interference model (Fig. 2.30). The model comprises the impedances of the electrodes and amplifier inputs ($Z_{ea}$, $Z_{eb}$, $Z_{rl}$, $Z_{ia}$, $Z_{ib}$) and the capacitances present between the body, electrodes, and amplifier system to mains noise and GND. The capacitances and interference currents generated by this capacitive coupling are as follows:
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**Figure 2.30:** Interference model for biopotential recordings. [101]

\[ C_{\text{pow}} \] capacitive coupling of body to mains;

\[ C_{\text{body}} \] capacitive coupling of body to GND;

\[ C_{ca} \] capacitive coupling of cable “a” to mains;

\[ C_{cb} \] capacitive coupling of cable “b” to mains;

\[ C_{\text{sup}} \] capacitive coupling of amplifier common to mains if isolated;

\[ C_{\text{iso}} \] capacitive coupling of amplifier common to GND if isolated;

\[ i_1 \] interference current generated via \( C_{\text{pow}} \) through body;

\[ i_a \] interference current generated via \( C_a \) through “a” input;

\[ i_b \] interference current generated via \( C_b \) through “b” input; and

\[ i_2 \] interference current generated via \( C_{\text{sup}} \) through amplifier common;
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This model also accounts for the fact that the system may or may not be isolated from earth GND. If the amplifier common is connected to GND, then the pictured isolation switch is closed. In a battery-powered system however, the amplifier common is isolated from GND and the switch remains open; added capacitances $C_{\text{sup}}$ and $C_{\text{iso}}$ must be taken into consideration, changing the interference characteristics of the system. The next section will investigate how to reduce the total system noise based on this model.

2.4.3.1 Model-Based Noise Reduction

As previously mentioned in section 2.4.1.2, impedance mismatch leads to distortion of the signal and also interference due to the common mode voltage manifesting itself as a differential signal at the amplifier inputs. One method to reduce this effect is to minimize the impedance mismatch. Another tactic is to reduce the common mode voltage of the subject.

In a non-isolated setup, the amplifier common is connected to GND, and the common mode interference $v_{\text{cm}}$ across the reference electrode impedance $Z_{rl}$ is mostly due to the interference current $i_1$ generated by the capacitive coupling between the body and the mains ($C_{\text{pow}}$). In this case, minimizing the capacitance by keeping a far distance from the mains can reduce the common mode voltage. Minimizing the impedance $Z_{rl}$ can also result in a reduction of common mode interference, however in
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A non-isolated setup, this may pose a safety hazard to the patient since the reference is connected to GND. In an isolated setup, this safety concern is no longer as much of an issue. Also, now the four capacitances $C_{\text{pow}}$, $C_{\text{body}}$, $C_{\text{sup}}$ and $C_{\text{iso}}$ play a role. In this case, the $v_{cm}$ can be lower than in the non-isolated setup if $C_{\text{sup}} \ll C_{\text{pow}}$ and $C_{\text{iso}} \ll C_{\text{body}}$. If a small battery-powered amplifier is used, $C_{\text{sup}}$ can be considered negligible. $C_{\text{iso}}$ should also be minimized with proper isolation circuitry.

In both non-isolated and isolated cases, using a driven reference electrode circuit known as a “Driven Right Leg” (DRL) or “Right Leg Driven” (RLD) reference can be used to actually “cancel out” the common mode voltage on the body. The DRL reference drives the body to the amplifier common, and can greatly minimize the $v_{cm}$ potential difference between the body and amplifier common. As is shown in Figure 2.31, the DRL circuit uses the common mode voltage (average potential between the two input signals) and forms a negative feedback loop using the amplifier common at the positive input of the amplifier, effectively reducing the common mode voltage on the body.

For high-impedance electrode types, a biasing network may be needed to stabilize the DRL circuit. Based on this simplified resistive model, however, the concept of how the DRL circuit is effective can be derived. The equivalent circuit of the DRL setup is shown in Figure 2.32. In this equivalent circuit, $R_a$ represent the value of the averaging resistors used to extract $v_{cm}$ (these resistors are not shown in Fig. 2.31). The current $i_d$ represents a displacement current flowing to the subject from the power.
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Figure 2.31: Simplified driven right leg (DRL) circuit to reduce common mode potential. [101]

Figure 2.32: Equivalent circuit of DRL setup. [106]

lines. By summing the circuits at the negative input of the operational amplifier and
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solving for $v_o$, we get the equations:

$$\frac{2v_{cm}}{R_a} + \frac{v_o}{R_f} = 0 \rightarrow$$ (2.26)

$$v_o = -\frac{2R_f}{R_a}v_{cm}$$ (2.27)

Solving for $v_{cm}$ gives the equation:

$$v_{cm} = R_{RL}i_d + v_o$$ (2.28)

Finally, substituting Eq. 2.27 into Eq. 2.28 gives the final common-mode voltage:

$$v_{cm} = \frac{R_{RL}i_d}{1 + \frac{2R_f}{R_a}}$$ (2.29)

The resistance of the driven electrode itself $R_{RL}$ will depend on the electrode type. Typically, the value of $R_f$ is chosen to be large to limit current, on the scale of 5-10MΩ and $R_a$ is chosen to be relatively small, on the scale of 25kΩ. These values then result in a large denominator, which is able to then effectively minimize the value of the resulting common-mode voltage $v_{cm}$ [106].

As described in sections 2.4.1.4 and 2.4.1.5, power line mains and other electrical interference generate displacement currents in the cables of the measurement system via capacitive coupling. This is especially a problem with a high impedance electrode-skin interface and consequently high impedance cables ($Z_{ea}$, $Z_{eb}$). To reduce this interference, a common tactic is to shield the cables from this outside noise. Shielded cables can come with their own issues, however. One technique is to shield the cables with an active shield “grounded” or shorted to the amplifier common. While this
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can reduce the interference currents within the wires, it does not usually reduce the total level of interference in the system. This is because the grounded shields are now capacitively coupled to the input wires, and add parallel parasitic capacitances between the amplifier inputs and the amplifier common, lowering the overall input impedance of the amplifier (Fig. 2.33).

![Figure 2.33: Parasitic capacitance induced with grounded shields; the added parallel capacitances $C_{ia}$ and $C_{ib}$ reduce the overall impedance of the amplifier inputs.](image)

To avoid this parasitic capacitance while still actively shielding the cables, another technique is to actively shield the cables using the buffered input signals. This technique is known as “guarding.” Guarding drives each cable shield to the same voltage as the signal input, removing any parasitic capacitance. The buffered signal is also now a low-impedance signal, making it much less susceptible to any interference current-induced voltage error. While this is effective, it may not be practical since each input now requires an extra amplifier to drive the shield (Fig. 2.34).

Sometimes to reduce the sheer amount of instrumentation needed to guard the
Figure 2.34: “Guarding” technique where shields are driven by the input signals; cable capacitances and parasitic capacitances to amplifier input become virtually nonexistent. [101]

input signals, the average or common mode \( (v_{cm}) \) of each pair of differential inputs is used instead to guard the pair of input cables. In this case, the interference due to common mode signals in the cables is reduced, since the potential difference between the common mode voltage and the guarded shields is minimal, or theoretically, equivalent. While this significantly reduces the cable capacitance for common mode signals, guarding using \( v_{cm} \) does not shield the cables from other differential mode interference currents that arise from changing electric fields. Thus, while it reduces the number of output buffers needed, it does not shield the cables from all interference.

Another option for reducing noise in cables that is even better than shielding in terms of effectiveness is buffering the signals at the source. This is the case with an “active” electrode. Essentially, there is an amplifier on the back of the electrode as close to the source as possible that acts as an impedance converter. The high input
impedance signals at the electrode-skin interface are fed into the amplifier and “converted” to low impedance signals at the amplifier output before sending them down through the cables. As mentioned in section 2.4.1.4, the interference voltage error is the product of the displacement current and the impedance. Thus, if the impedance of the cables is very low (< 10Ω for typical amplifiers), the resulting voltage error due to the interference current is negligible. An operational amplifier can be configured as a unity-gain buffer as is shown in Figure 2.35 that buffers each input separately as recommended in the previous section (2.4.2.1); or in the case of a bipolar circuit, the amplifier can be used in the differential configuration, taking in both inputs and outputting the buffered differential signal (Fig. 2.36). A gain is also implemented in this case, increasing the overall signal strength in comparison to the noise floor. The active design also reduces impedance mismatch. Since the impedance conversion is

Figure 2.35: Active electrode design with amplifier in unity-gain configuration.
happening at the source, mismatch due to the different capacitive coupling between the inputs is significantly less likely to occur. As such, the desired lower electrode-skin interface as described in sections 2.2.1, 2.2.2 and 2.4.2.1 is theoretically not as necessary. Lowering the interface impedance reduces the interference-induced voltage errors as well as the significance of impedance mismatch due to the voltage divider effect. With an amplifier directly on or as close as possible to the signal source, however, the impedance conversion is done immediately, obviating the need of a lower interface impedance [83]. In Figures 2.37 and 2.38, an interference current on the cables and power line interference were induced as discussed in sections 2.4.1.4 and 2.4.1.5, respectively, on both a remote and active electrode. As can be observed by the resulting graphs, the active electrode’s noise rejection performance was much better than the remote electrode; the power line noise is minimal in comparison and
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Figure 2.37: Interference caused by an induced current on the electrode cable in (a) remote vs. (b) active electrode.

the induced current interference is negligible. While an active electrode is the most preferential, an extra differential amplifier per two electrodes or operational amplifier per single electrode are now required on the electrode backs.

Depending on the design constraints or requirements of the measurement system, the active design may not be optimal. For example, in a short-term setting such as monitoring ECG of a patient in the ER, a quick, reliable electrode application
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Figure 2.38: Power line interference induced in (a) remote vs. (b) active electrode.

method is needed. For this application, typically disposable adhesive-backed pre-gelled electrodes with shielded cables are the electrode setup of choice. Due to their adhesive-backed design and pre-gelled interface, application is quick and easy and requires no skin preparation. The adhesive also helps minimize motion artifact. Once the subject no longer needs to be monitored or is moved elsewhere, the electrodes can be thrown away and new ones are used for the next patient. An active type of electrode in this setting would be less useful; it would have to be washed in between subjects, and securing it onto the skin would prove to be a bulky task which is not straightforward.

In summary, based on this interference model, the factors that best reduce the overall noise in the measurement system are: common mode rejection with an RLD circuit, shielded cables using the guarding technique and/or an active electrode design.
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2.4.4 Summary

Taking both the amplifier circuit noise model and interference model into consideration, the optimal electrode design would be an active electrode secured well onto the skin. The amplifiers used in the measurement system should have low input-referred noise and a high input impedance with sufficient analog gain. To even further reduce common mode interference, a driven reference using an RLD circuit should be implemented.

If an active electrode is not possible, then effort should be taken to minimize the electrode-skin interface impedance of the remote electrodes, and measurement cables should be shielded with the guarding technique using individual input signals. Guarded cables using the $v_{cm}$ signal or shielded cables shorted to ground are also an option, but can lead to sub-optimal noise rejection and circuit instability.

The best solution for noise reduction is to remove it at the source with the recommendations made above. If the noise reduction is still insufficient due to poor implementation or certain design constraints, another option is to also filter the noise off-line during signal processing. Motion artifact is typically a low-frequency signal and can be mostly filtered out using a high-pass filter. Power line noise can be attenuated using a notch filter aimed at the 50/60 Hz signal. In the case of an envelope signal, using the notch filter is usually not a problem. For methods of control that require raw sEMG such as in pattern recognition, the notch filter may also remove signals of interest and is not usually recommended. Other interference currents re-
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resulting from changing electrical or magnetic fields and impedance mismatch cannot usually be removed off-line in a straightforward way, however, and must be dealt with at the source.
Chapter 3

Textile Electrodes

3.1 Textile Electrode Applications

Textile sensors can be used for a variety of applications, both medical and commercial. In both types of applications, the sensors measure biopotentials, mainly ECG, sEMG, and EEG. In the medical field, the sensors are geared towards acquiring accurate signals meant for diagnostics or long-term monitoring. In the commercial realm, the sensors are integrated into intelligent wearables and marketed as exercise or personal health monitors. The textile sensors are also seen in interactive clothing that responds to the wearer’s biopotential signals.
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3.1.1 Medical

Typically, wet gel electrodes are used to monitor biopotentials such as ECG, sEMG and EEG in a diagnostic or hospital setting. While these electrodes are good for short-term monitoring, they are not conducive to long-term monitoring needs. This is due partly to the fact that the gel used eventually dries out and signal quality becomes poor. As such, textile electrodes are being investigated for use in the medical field to monitor a variety of biosignals in an active or long-term setting [108–110].

3.1.1.1 ECG

Using textile electrodes to monitor ECG in an unobtrusive manner during everyday activities is a research area of interest [111–116]. An example of this is integrating textile electrodes into bedsheets to monitor ECG while a subject is sleeping. Ishijima et. al use conductive textile sewn onto a pillow and the bottom half of a bedsheets to create two large electrodes for monitoring ECG [111]. Peltokangas et. al use multiple textile electrodes sewn into the bedsheets underneath the subject to monitor ECG [115]. To monitor a subject’s ECG while driving, Yang et. al embed textile electrodes into the driver’s seat of a vehicle. This is then coupled to an alarm system in the car that could alert the driver if it detects the onset of any health issues that may require him or her to pull over and/or seek help [114]. Other researchers have worked on developing textile electrodes that can easily embed into clothing for monitoring ECG throughout the day while the subject is active [112, 113, 116, 117].
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With the electronics and a Bluetooth transmitter embedded in the clothing as well, this allows for monitoring ECG in a mobile setting.

3.1.1.2 sEMG

Research in using textile electrodes for monitoring and recording sEMG is also an area of interest [86, 118–122]. In a project by ConText, non-contact capacitive textile electrodes were sewn into a vest for monitoring muscle activity in the back and shoulders (Fig. 3.1(a)). Studies have shown that sEMG signals generated from these muscles can be related to stresses and fatigue induced by mental or postural loads. As such this vest was meant to unobtrusively monitor these stresses of a subject in his or her working environment [121,122]. Finn et. al and Lintu et. al both focused on integrating textile electrodes into compression shorts and shirts for monitoring muscle activation in a mobile setting [119,120]. By embedding the electrodes into clothing, the recorded sEMG can be used to determine active muscle groups during exercise, for example. Pylatiuk et. al and Farina et. al worked on developing textile sEMG sensors as alternatives to standard electrodes for the purpose of controlling prosthetic and rehabilitation devices, respectively [86,118].

3.1.1.3 EEG

Textile electrodes for monitoring EEG have also been investigated. The challenge is greater for this electrode type because the sensors must be able to work over any hair
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that is on the head of the subject. Lofhede et al and Lin et al have designed textile EEG electrodes using textile contacts with a foam backing to press the electrodes into the scalp. These electrodes could then potentially be used to more comfortably monitor EEG in a daily, mobile setting [85,123,124]. Lofhede et al worked on these electrodes more specifically as a softer electrode alternative for monitoring neonates, since maintaining good contact pressures with conventional hard electrodes can be potentially dangerous for the neonates [124].

3.1.1.4 Multiple Signals

In addition to monitoring ECG, sEMG or EEG separately, an area of interest has also been integrated clothing embedded with multiple textile electrodes for monitoring a number of different types of biosignals. Wearable clothing such as the WEALTHY health monitoring system [125] are designed to monitor multiple biosignals all over the body such as ECG, EMG, index of movement, skin temperature and respiratory rate (Fig. 3.1(b)) [125–127]. The benefit of this integrated clothing is the ability to monitor multiple signals in a natural and mobile setting. Bouwstra et al developed a prototype jacket embedded with sensors to monitor biosignals of a neonate (Fig. 3.1(c)) . This is beneficial since it is much easier and unobtrusive to put this integrated jacket onto a neonate than attaching individual, multiple electrodes which may be uncomfortable for the neonate.
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Figure 3.1: Textile electrode-embedded clothing for monitoring biosignals: (a) Work-monitoring vest by ConText [121], (b) “Smart Shirt” by Sensatex [128] and (c) “Smart Jacket” prototype for monitoring neonates [127].

3.1.2 Commercial

The commercial applications for textile electrodes overlap with medical applications in the sense that in both cases, the main purpose is to monitor biosignals in a long-term or mobile environment. In the commercial area, the focus is on integrating these sensors into clothing to be worn in every day life. These wearables are commonly used to monitor activity during exercise. A commercially available sensor wearable is a heart rate strap or bracelet. This strap is worn over the chest or over the wrist and uses embedded textile electrodes to monitor and report heart rate. Some have even integrated the textile heart rate monitor as part of sports clothing (Figure 3.2(a)).

Another application for textile electrodes in the commercial realm is less focused on acquiring accurate biosignals and more focused on using them in interactive clothing. This “smart” clothing can for example use a person’s ECG signals to light up LEDs on a shirt every time their heart beats (Fig. 3.2(b)).
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Figure 3.2: Commercially available wearables: (a) Sports bra with embedded heart rate monitor by Textronics, Inc (Chadds Ford, PA) and (b) Interactive ECG shirt Heart-Felt Apparel by Sean Montgomery (ProduceConsumeRobot.com).

3.2 Design Challenges with Textile

There are both signal quality and manufacturing challenges that must be addressed when designing textile electrodes. Hindrances to good signal quality arise from the nature of the dry electrode interface and are similar to those challenges faced with conventional dry electrodes. Manufacturing challenges arise from the nature of the textile itself as a soft fabric component. An additional set of considerations must be taken when integrating a fabric component versus a standard component into a device with standard electronics and for long-term use.
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3.2.1 Signal Quality

3.2.1.1 Impedance

Similar to conventional dry electrodes, a textile-based dry electrode will have a high interface impedance in comparison to wet electrodes. This impedance will depend on a variety of factors such as the conductivity and polarization of the material used, as well as the electrode-skin contact. One way of addressing this issue is to make the electrode contacts larger. In research literature, the size of textile electrodes range from square contacts 1 cm in width and height to the size of a pillow [111,124]. As expected, with larger textile electrodes, interface capacitance increases and the overall interface impedance decreases [81]. Another method to lower impedance is to use a conductive hydrogel membrane between the electrode and the skin (Fig. 3.3(a)) [81,109,129]. Due to the tacky nature of the hydrogel, more surface area on the skin maintains good contact with the electrode, increasing the interface capacitance. Furthermore, the membrane acts as a gel electrolyte layer between the textile and the skin, lowering the interface resistance and helping facilitate charge transfer of the sEMG signals to the electrode [81,109,129]. Lin et. al describe an EEG electrode that uses conductive foam wrapped in a conductive textile layer [85]. With this design, the foam-backed textile electrode can be pressed up against the scalp and can conform to irregularities on the surface that arise from hair or dirt. This helps increase the surface area of the contact to the skin, increasing the interface capacitance and lowering the
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overall impedance (Fig. 3.3(b)).

![Diagram](image)

**Figure 3.3:** Examples of design considerations taken to lower interface impedance: (a) Hydrogel membrane on top of a textile electrode [109] (b) Model showing how foam-backing can maximize contact surface area by conforming to irregularities such as hair follicles [130].

### 3.2.1.2 Motion Artifact

The novel textile electrode designs discussed at the beginning of this chapter do show promise in acquiring accurate and useful biosignals in a long-term setting, however they do not fully address how to maintain the signal quality in an active setting. Motion artifact is still very much an issue and stems from the challenge of maintaining good electrode contact while the subject is mobile. The typical method for securing the electrode-skin contact is to achieve a contact pressure that can minimize motion artifact while still being comfortable [119, 120, 125–127]. Another method is to use a foam-backed electrode as discussed in the previous section. The foam backing can be used both as a way of lowering impedance and also maintaining contact during motion. The reasoning is that the foam can act as a damper for any movement inflicted
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on the electrode, absorbing the movement without compromising the electrode-skin contact itself (Fig. 3.4).

Figure 3.4: Model showing how a foam-backed electrode can minimize electrode lift-off and consequently motion artifact under conditions of movement [130].

3.2.2 Manufacturing

3.2.2.1 Hard-to-Soft Connections

The invention of conductive textiles has opened up an entire field of “soft” electronics. The electrical properties allow for fabric circuits to be designed using conductive fabrics and threads as pads and traces. By sewing into a conductive textile patch with conductive thread, an electrical connection is made. The use of these soft materials give rise to challenges that are not normally faced with conventional “hard” printed circuit boards (PCBs) and their associated components. Same challenges arise when interfacing with soft sensors made of fabric rather than hard sensors made with metal components.

The biggest challenge is designing a stable and secure interface from a soft fabric
component to a hard metal component; this is known as the “hard-to-soft” connection [131]. This challenge is two-fold: the connection must (1) electrically couple the components and (2) be mechanically robust. At a solder joint between two metal components, the interface is mechanically sound and the integrity of the electrical connection is high. Solderable thread and textile mimic this hard-to-hard connection by allowing for metal components to be soldered directly onto the fabric or vice versa (Fig. 3.5(a)). Conductive fabrics that are not solderable can be made so by using a technique called electroplating. Another common interface is the use of metal fasteners such as button snaps or clasps. Solderable or non-solderable thread can be used to sew into a metal fastener electrically coupling the two components. The mating connector can then be attached to a mating hard or soft component. Sewing conductive thread directly onto hard components such as wire or the legs of an integrated circuit (IC/chip), or sewing into vias of a PCB can also establish the hard-to-soft connection (Fig. 3.5(b,c)). Another method is to directly pierce through the conductive textile with hard components such as header pins, crimp connectors or IC legs [131].

The above methods all describe how to electrically couple the soft and hard components, however due to the nature of textile, the interface points can easily weaken, tear or even break. Added measures need to be taken to keep these connection points durable. Sometimes epoxy or solder must be used at the interface points to strain relief the area and maintain mechanical stability. In the case of biopotentials and
sensors, the stability of the connection is especially important. If the interface points are not mechanically sound, any movement between two components on the high impedance sensor side (before the signal reaches the processing electronics) can lead to momentary breaks in the electrical connection that appear as artifacts in the signal.

### 3.2.2.2 Moisture Susceptibility

Another issue with using textiles in electronics is their susceptibility to moisture. Measures can be taken to seal hard components from moisture such as applying a layer of conformal coating or epoxy. With fabrics, however, flexibility is typically a desired characteristic, and so applying hard coatings to seal from moisture is not always an option. Especially in the case of wearables, this becomes a problem, since accumulated sweat can lead to shorting between components on the conductive fabric. Flexible insulation, moisture wicking fabric or insulation between conductive fabric components can help reduce the adverse affects of moisture.
CHAPTER 3. TEXTILE ELECTRODES

3.2.2.3 Durability

The longevity of the fabric is also of concern. Unlike hard plastic or metal, fabric can wear out fairly easily over time. Wearables, again, are especially susceptible to this problem since the fabric is exposed to constant friction with the skin and impurities such as sweat and dirt. Fraying and tearing can break electrical connections and reduce the overall integrity of textile circuits or sensors.

3.3 Evolution of the Textile Electrode

Design

The design of the newly developed textile electrodes addressed in this thesis was a research process, beginning with hand-made prototypes and evolving into more finalized prototypes manufactured by a third party. The electrode can essentially be split up into three elements: the electrode module that houses the electrode contacts, the cable which transmits the signal down the processing electronics, and the interface between the cable and the electronics.
CHAPTER 3. TEXTILE ELECTRODES

3.3.1 The Electrode Module

3.3.1.1 Contact Material

The electrode module comprises the base fabric (non-conductive) and the electrode contacts (conductive). For the initial hand-made prototypes, three conductive materials were investigated for the fabrication of the electrode contacts: MedTex™P-180, Ripstop, and 117/17 2-ply conductive thread by Statex (Bremen, Germany) (Fig. 3.6).

![Conductive textile materials](image)

(a) MedTex™ P-180, (b) Ripstop and (c) 117/17 2-ply conductive thread by Statex (Bremen, Germany). Source: Sparkfun.com

The textiles were chosen for their low resistance properties. It was soon found that the thread had a high linear resistance (> 100Ω per in.) and also required more advanced sewing techniques for creating an embroidered contact area. As such, focus was geared towards the other two materials whose properties are shown in Table 3.1.

The MedTex™P-180 fabric had a resistance of < 5Ω per square and the Ripstop
had a lower resistance of $<0.02\, \Omega$ per square. In terms of handling and use, the fabrics were similar, however due to the lower resistance, the hand-made prototypes used the Ripstop fabric. Moving to the manufactured prototype, however, the chosen material was 99.9% silver-plated cotton, similar to the Med-Tex™ material. The reason for this material choice was that this material was known to be compatible with skin, whereas Nickel, a material present in the Ripstop fabric, can lead to an allergic skin reaction.

### 3.3.1.2 Contact Shape

The original electrode shape was based on a circular contact and a “guard ring” reference electrode around the contact (Fig. 3.7(a)). The size of the circular contact was chosen to be 1 cm, similar to conventional remote metal electrodes on the market. The guard ring reference size was chosen arbitrarily. This was then evolved to a pair of bipolar electrode contacts surrounded by a guard ring reference (Fig. 3.7(b)). While somewhat effective, this design was not practical given the size constraints of the overall liner design. If potentially eight electrode pairs needed to be embedded

<table>
<thead>
<tr>
<th>Name</th>
<th>Materials</th>
<th>Resistance per square (Ω)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Med-Tex™ P-180</td>
<td>99.9% Silver, Cotton</td>
<td>$&lt;0.02$</td>
</tr>
<tr>
<td>Ripstop</td>
<td>Tin, Nickel, Silver, Nylon</td>
<td>$5$</td>
</tr>
</tbody>
</table>

**Table 3.1:** Conductive textile properties.
Figure 3.7: Hand-made prototypes: (a) center electrode with guard ring reference and (b) bipolar electrode with guard ring reference.

in a liner, then a smaller design would be required. With this in mind, the design of the electrode was shifted to more closely resemble the Otto Bock or Touch Bionics electrode, with two differential contacts placed 2 cm apart, and a reference electrode located in the middle.

In an attempt to achieve better contact with the skin, exploration was made into using foam-backed electrodes to press the electrode surfaces deeper into the skin and beads underneath the contacts to create raised individual contacts (Fig. 3.8). This was in an attempt to lower the contact impedance. The foam-backed electrodes were too large however, and the raised contacts were difficult to manufacture without introducing any frayed edges. As such, both design directions were not pursued.

The final hand-made prototype was similar to the current manufactured version, comprising a base fabric and three flat conductive fabric contacts: two bipolar contacts and a larger middle reference. Prototype runs were then made with a third-party
CHAPTER 3. TEXTILE ELECTRODES

Figure 3.8: Hand-made prototypes: (a) contacts raised using beads and (b) foam-backed electrode.

manufacturer producing a two-contact (2C) version with a separate single contact (1C) reference, and a three-contact (3C) version where the middle contact serves as the reference. The most current prototype design is also silicone-backed for the purpose of making it easier to embed into a silicone liner and also adding height that would allow for the electrode to protrude and make good contact with the skin. These electrodes are shown in section 3.4.

3.3.2 The Cable

The cable is responsible for transmitting the signal from the electrode-skin interface to the electronics. For initial testing purposes with the early electrode designs, conventional multi-wire shielded cable was used. To interface the cable with the electrodes, small metal button snaps were sewn onto the back of the electrode patches. Each snap was electrically coupled to a contact using conductive thread and the mat-
CHAPTER 3. TEXTILE ELECTRODES

ing snaps soldered onto its respective wire (Fig. 3.9). The cable shield was grounded.

![Figure 3.9](image1.png)  ![Figure 3.9](image2.png)

**Figure 3.9:** Hard-to-soft connections for electrode prototypes: (a) snaps sewn using conductive thread and (b) snaps interfaced with cable with mating snaps.

Moving from the cable to a textile solution, the electrodes were interfaced with a conductive “ribbon cable” made by Fabrickit (New York, NY, U.S.) (Fig. 3.10(a)). This particular ribbon cable was chosen because of its very low resistance and because the traces were solderable. Because it was a three conductor cable, it could also conveniently transmit the signals from the three contacts in the early 3C design. To attempt to shield the cable, a fabric sheath with a layer of conductive fabric shorted

![Figure 3.10](image3.png)  ![Figure 3.10](image4.png)

**Figure 3.10:** Fabric cable (a) and fabric shield (b).
CHAPTER 3. TEXTILE ELECTRODES

to the circuit ground was drawn over the ribbon cable (Fig. 3.10(b)). The shielding layer restricted the flexibility of the cable, however, and the use of a shield was not pursued.

The manufactured prototypes instead used the manufacturer’s own proprietary cable they termed the textile data cable (TDC). The cable is “textile” in the sense that it comprises a polyester sheath surrounding the conductive traces. The conductive traces are made from thin multi-stranded copper wire plated with tin to prevent corrosion. The cable remains unshielded to preserve its thin and flexible nature. The TDC is pictured in section 3.4.

3.3.3 Cable-Electronics Interface

Interfacing to the electronics goes hand-in-hand with the cable design and involves the initial hard-to-soft connection. Initially when using conventional cables, header pins could be soldered directly onto the cables. Moving to the textile ribbon cable, the traces on the textile cable had a standard pitch of 2mm and as such could be interfaced with off-the-shelf header pins. Because the traces on the textile ribbon cable were solderable, the pins were, at first, soldered directly onto the cable and also epoxied for strain relief. A “piercing” solution was then adopted where the header pins were simply pierced through the traces, electrically coupling each pin to a trace. To maintain better contact, the joints were usually epoxied.

For interfacing with multiple electrodes, a “dock” solution was developed. On
CHAPTER 3. TEXTILE ELECTRODES

Figure 3.11: Prototype docking solution for textile cables: (a) unplugged and (b) plugged.

this dock, the multiple cables could be drawn through slits and a mating piece containing the appropriate header pins would pierce through all the traces when plugged into the dock. The docking solution was attempted with the TDCs of the manufactured prototype as shown in Figure 3.11. The polyester sheath was difficult to pierce through, however. The thin multi-stranded wire would also often shift or break as the pins pierced through the cable, creating a loose or broken electrical connection. Furthermore, the header pins did not always align properly with the traces, since the tolerance was very tight. As such, these cables were instead terminated with connectors to achieve a hard-to-soft connection that was also a “pluggable” solution (as shown in section 3.4).
CHAPTER 3. TEXTILE ELECTRODES

3.4 The “Final” Electrode Assembly

This section will describe the remote and active textile electrode prototypes that were addressed in this thesis. Experimental methods to test these prototypes along with their results and discussion are presented in Chapter 4.

3.4.1 Remote

The “final” or more appropriately, the most current, electrode assembly is shown in Figure 3.12; the full assembly comprises the electrode module, the TDC and the pluggable connector. This is a remote version of the electrode, in the sense that there are no components on the back of the electrode and all the amplification and filtering is done downstream. As will be shown in Chapter 4, however, it becomes very apparent that, as expected, this remote design is very susceptible to noise interference. Shielding the TDC could be an option but would require additional layers of shielding fabric. This would increase the thickness of the cable, reducing its flexibility and also increasing the chance of producing artifacts arising from the shield coupling capacitance. As such, an active design was also investigated.
CHAPTER 3. TEXTILE ELECTRODES

Figure 3.12: Manufactured remote textile electrode prototype: (a) TDC with connectors plugged into mating electronics and (b) 3C, 2C, and 1C electrode versions.

3.4.2 Active

The most current active electrode prototype comprises a housed protoboard located between the electrode module and the cable (Fig. 3.13). The active circuit layout on the protoboard is shown in Figure 3.14. The circuit comprises a dual op-amp configured as two unity-gain buffers which serve to buffer the electrode input signals. The amplifier is attached to the 2C electrode version, converting the signals from high impedance to low impedance signals, and essentially eliminating any cable interference that can be induced downstream of the amplifier. This also significantly reduces impedance mismatch and power line interference. Results from noise analysis experiments are presented in Chapter 4. The amplifier circuit for the active textile electrode prototype is located slightly downstream of the electrode, however in a proper active electrode, the amplifier circuit should be encased and shielded directly.
CHAPTER 3. TEXTILE ELECTRODES

on the back of the electrode. The next, future prototype will address this.

Figure 3.13: Active textile electrode prototype with packaged “pre-amp”.

![Active textile electrode prototype with packaged “pre-amp”]

Figure 3.14: Active textile electrode circuit.

3.4.3 Dimensions

Dimensions of all three electrode versions (3C, 2C, and 1C) are shown in Figures 3.15, 3.16 and 3.17.
CHAPTER 3. TEXTILE ELECTRODES

Figure 3.15: Dimensions for 3C textile electrode.

Figure 3.16: Dimensions for 2C textile electrode.
Figure 3.17: Dimensions for 1C textile electrode.
Chapter 4

Experimental Results

A total of six different experiments were used to investigate the remote and active textile electrode prototypes. Impedance analysis investigated the nature of the electrode-skin impedance using the remote textile electrodes. Noise analysis focused on the noise characteristics of the electrodes. For this analysis, two experiments were conducted to investigate how well the remote and active textile electrode prototypes can reject power line interference and induced current cable interference. A fourth experiment analyzed the envelope quality of the sEMG signal acquired with an active textile electrode prototype to investigate if the resulting envelope signal is suitable for amplitude-based direct control of a prosthesis. Finally, the last two experiments tested the electrode prototypes in a functional setting to investigate their performance using both both a direct control and pattern recognition system. This chapter will present the methods, results and discussion for each experiment.
CHAPTER 4. EXPERIMENTAL RESULTS

4.1 Impedance Analysis

Revisiting the electrode types from Chapter 2, a textile electrode falls under the “dry” electrode category and is expected to have a high electrode-skin interface impedance. The purpose of this experiment was to investigate the textile electrode-skin interface impedance versus that of conventional metal dome electrodes. A few parameters were of interest: the change in the interface impedance over time, the effect of contact size and the effect of spraying down the arm with water as part of the skin preparation process.

4.1.1 Methods

To investigate the impedance of the textile electrodes, impedance analysis was performed using an LCR meter (LCR-800 by GW Instek, New Taipei City, Taiwan) using the remote textile electrode prototypes. By measuring the impedance through two textile electrode contacts placed on the skin, we essentially measure the electrode-skin-electrode impedance or 2 x the electrode-skin impedance assuming matched impedances at the contacts and negligible body resistance $R_b$ (Fig. 4.1) [65]. An impedance frequency sweep was done manually at a set of 10 frequencies over

---

Figure 4.1: Electrode-skin-electrode interface impedance.
CHAPTER 4. EXPERIMENTAL RESULTS

<table>
<thead>
<tr>
<th>Electrode</th>
<th>Shape</th>
<th>Size (mm)</th>
<th>Area (mm$^2$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>2C</td>
<td>Rectangle, Flat</td>
<td>3 x 10</td>
<td>30</td>
</tr>
<tr>
<td>1C</td>
<td>Circle, Flat</td>
<td>10 dia.</td>
<td>78.5</td>
</tr>
<tr>
<td>MD</td>
<td>Circle, Round</td>
<td>10 dia.</td>
<td>&gt;78.5</td>
</tr>
</tbody>
</table>

Table 4.1: Electrode properties.

the range of 30-2000 Hz and the average of 5 impedance measurements at each frequency was recorded. The LCR meter reported the equivalent R and C values of the electrode-skin-electrode circuit. Once the equivalent R and C values were found, the electrode-skin impedance of one electrode contact was calculated as $\frac{Z}{2}$ where:

$$Z = \frac{RX_C}{\sqrt{R^2 + X_C^2}}.$$  \hspace{1cm} (4.1)

and

$$X_C = \frac{1}{2\pi fC}.$$  \hspace{1cm} (4.2)

Both the remote 2C (two-contact) bipolar electrode and a pair of remote 1C (one-contact) reference electrodes were investigated and compared against a pair of conventional remote metal dome (MD) electrodes (by Liberating Technologies, Inc.). Shape properties of these three electrode types are listed in Table 4.1.

When testing each electrode pair, the same silicone cuff was used on the inner forearm on the same subject and arm location within the same day. The measurements were also taken indoors where the temperature was kept constant. Between testing electrode types, the forearm was wiped down with 80% isopropyl alcohol and
rested for 15 minutes. The electrode-skin impedance was observed at 5 minutes, 20 minutes and 40 minutes after application of the electrode to the skin. The effect of spraying down the arm with water as part of the skin preparation process was also investigated for the textile electrodes. For those trials, the arm was sprayed down with 3 sprays of tap water prior to application of the electrode on the skin. The reason this particular form of skin preparation was investigated was because it is more accessible and comfortable for an amputee to spray down his or her arm before donning a prosthesis than it is for him or her to abrade the skin.

4.1.2 Results

All three types of electrodes exhibit the expected frequency relationship of decreasing interface impedance with increasing frequency with both skin preparation techniques. Figure 4.2 shows the interface impedance versus frequency relationship in a short-term setting after 5 minutes and in a long-term setting after 40 minutes for all electrodes with the normal skin preparation process.

Observing Figure 4.2, it is shown that the 2C electrode exhibits an interface impedance that is almost a factor of magnitude greater than the metal dome electrodes. The 1C electrode is much more similar to the metal dome, but still exhibits an interface impedance between 3-4 times greater than the metal dome at lower frequencies. Interface impedance of all three electrode types drop over the 40 minute period. Focusing on the normal skin preparation trials, the estimates of the equivalent
Figure 4.2: Interface impedance versus frequency for textile and metal dome electrodes over time using “dry” normal skin prep process.

individual impedance components (R and C) of the electrode-skin-electrode interface are plotted (Fig. 4.3 and 4.4).
Figure 4.3: Interface resistance versus frequency for textile and metal dome electrodes over time with normal skin prep process.

Figure 4.4: Interface capacitance versus frequency for textile and metal dome electrodes over time with normal skin prep process.
CHAPTER 4. EXPERIMENTAL RESULTS

The separate R and C components also follow the same trend as the overall impedance, with the 2C electrodes exhibiting the greatest resistance and smallest capacitance. As is listed in Table 4.1, the 2C contacts have a surface area of $30mm^2$ whereas the 1C versions have a surface area of $78.5mm^2$. While the circular metal dome and 1C electrodes have the same diameter, the overall surface area of the metal dome electrodes is greater than that of the 1C because of the added curvature. As such, the observations match the expected outcome based on the electrode designs: as the surface area of the electrode increases, the interface capacitance increases and the overall interface impedance decreases. Furthermore, over the 40 minute period, resistances of all three electrode types decreases while capacitance increases.

Figure 4.5 shows the interface impedance versus frequency relationship of the electrodes with adding water as part of the skin preparation process for the textile electrodes.
CHAPTER 4. EXPERIMENTAL RESULTS

Figure 4.5: Interface impedance versus frequency for textile and metal dome electrodes over time with adding water as part of the skin preparation process for textile.

A significant decrease in the textile electrode interface impedance for both textile electrode types is observed when water is added as part of the skin preparation process (Fig. 4.5). At the 5-minute mark, the 2C interface impedance decreases by almost 10-fold in comparison to the dry skin preparation trials. The 1C interface impedance decreases almost 2-fold and all electrode interface impedances are now within the same order of magnitude. As such, with the tap water skin preparation process, the overall interface impedance of both the 2C and 1C electrodes decreases to within the $<200\text{k}\Omega$ range and much more closely resembles that of the dry metal dome electrode (no water added). Again, impedance of all three electrode types drops over the 40 minute period, but the observed change for the textile electrodes is less than was observed with the dry skin preparation trials, possibly since the effect of
sweat is masked by the already wetted skin. Focusing now on the added water skin preparation trials, the estimates of the individual impedance components (R and C) of the electrode-skin-electrode interface are plotted (Fig. 4.6 and 4.7).

**Figure 4.6:** Interface resistance versus frequency for textile and metal dome electrodes over time with adding water as part of the skin prep process for the textile electrodes.
Figure 4.7: Interface capacitance versus frequency for textile and metal dome electrodes over time with adding water as part of the skin prep process for the textile electrodes.

By adding a few sprays of water to the skin preparation process, it can be observed that the resistance of both textile electrode types more closely resemble one another. For the lower frequencies of interest, the values are also less than the initial contact impedance of the metal dome electrode with the dry skin preparation, although after 40 minutes, the metal dome electrode interface resistance does decrease and more closely resembles that of the textile electrodes. The water also did help increase the capacitance of the textile electrodes, although especially after a 40 minute period, the metal domes still exhibit higher capacitance values. This can be attributed to the assumption that the tap water more significantly affects the resistance component of the contact interface by creating an ionic concentration gradient at the skin surface.
CHAPTER 4. EXPERIMENTAL RESULTS

and opening up appendageal pathways.

Comparing the overall impedance of all the electrode types across both skin preparation techniques at 20 minutes, it becomes clear that with the water skin preparation, the overall interface impedance can be reduced significantly to more closely resemble that of the standard metal dome electrodes (Fig 4.8).

Figure 4.8: Interface impedance vs. frequency without versus with adding water as part of the skin prep process for the textile electrodes.

4.1.3 Discussion

The interface impedance of all the electrodes was observed over a 40 minute period. As can be seen by Figures 4.2 and 4.5, the impedances drop over time with both skin preparation processes. This can be possibly attributed to the assumption that over time, the subject begins to sweat underneath the silicone, opening up
appendageal pathways, which could also explain the observed decrease in interface resistance over time (Fig. 4.3 and 4.6). Capacitance values also increased, which could be attributed to the accumulating sweat increasing the overall effective surface area of the contact interface (Fig. 4.4 and 4.7). In general, the interface impedance of the textile electrodes is decreased by a factor between 1-2 over the 40 minute period. The metal dome electrodes are affected to a lesser degree. Over time, the interface impedance of the textile electrodes could potentially decrease to match that of the metal dome electrodes depending on the degree of sweating the subject experiences at the electrode interface.

Spraying the arm down with a few sprays of tap water was meant to imitate the sweating process in order to observe its effects. By observing the graph in Figure 4.8, it is apparent that using water prior to application does significantly lower the interface impedance. This observation suggests that with a few sprays of tap water prior to application of the electrodes, the process of decreasing the impedance over time due to sweat can be sped up. Furthermore as can be observed in Figure 4.5, the lowered interface impedance is maintained over the 40 minute period, implying that the silicone is able to effectively seal in the moisture at the interface.
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4.2 Power Line Interference

The noise characteristics of the remote 2C textile electrode and active 2C textile electrode prototypes were investigated when exposed to power line interference. The use of a notch filter and a right-leg driven (RLD) reference for optimized noise reduction was also investigated.

4.2.1 Methods

For the following tests, a 2C textile electrode and a 1C textile reference electrode were placed on the forearm of a subject underneath a silicone cuff. To induce a large power line interference signal, the subject grabbed the power cable of a laptop charger. The electrode noise performance was tested under four conditions: “Nothing” (no RLD, no notch), notch only, RLD only and notch + RLD. The RLD reference was set to derive its signal using the common mode signal of the bipolar inputs from the electrode being tested, and the notch filter was configured to attenuate 60Hz signals. For each recording, the subject was asked to perform a 75% MVC contraction sustained for 4 seconds, followed by a 4 second rest period. This was done before then inducing a 4-second power line interference in order to show the relative magnitude of the interference to the sEMG signal and baseline noise. Both a remote 2C textile electrode and an active 2C textile electrode were tested using a remote 1C textile reference electrode.
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4.2.2 Results

The results for the remote 2C electrode are shown in Figure 4.9. The last 4 seconds of each recording show the resulting signal on the electrode while the subject is grabbing the power cable with the opposing hand. Note that during this last 4 second period, no contractions are being formed and the subject’s forearm is at rest.

As can be seen by Figure 4.9(a), with neither the RLD reference or notch imple-
CHAPTER 4. EXPERIMENTAL RESULTS

mented, the high-amplitude power line signal between 11 and 15 seconds can easily mask the raw sEMG signal (between 4 and 8 seconds). Implementing the notch filter alone or the RLD reference alone significantly attenuated the power line interference and yielded similar results (Fig. 4.9(b-c)). Both features implemented resulted in the best attenuation, as can be seen by Figure 4.9(d). Focusing on the power line interference area on the graph between seconds 11 and 15 however, it is easy to visually point out the characteristic 60 Hz signal, even after the attenuation facilitated by the RLD reference and the notch filter. Figure 4.10 displays a one second interval during the interference-induced period to better visually confirm that the signal is purely 60Hz interference.

Figure 4.10: One-second interval during power line interference on remote 2C textile electrode for all four test conditions.

The Power Spectral Density (PSD) of these signals was also computed to analyze the relative power of the signals’ frequency spectrum; an obvious peak near 60 Hz can be observed under all four conditions (Fig. 4.11).

Compared to the pure sEMG signal power whose true power is approximated
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Figure 4.11: PSD analysis on the remote textile electrode signal show peaks at power line frequency.

to be -15dB around the power line frequency, when no noise removal techniques are implemented, the interference signal at power line frequency is observed to be approximately 6.5dB. With the notch or RLD implemented, the power of the this interference signal decreases to about 0dB. This peak is best attenuated when both RLD reference and notch filter are implemented, attenuating the power line frequency to about -5dB (Fig. 4.12).
Figure 4.12: PSD analysis on the remote textile electrode signal show peaks at power line frequency; zoomed in.

The same experiment was done using the active 2C textile electrode prototype. In this case, the power line interference is negligible on the sEMG signal, if even present, under all four conditions. PSD analysis of the signals do not show a significant peak at the power line frequency under all four test conditions (Fig. 4.13, 4.14).
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Figure 4.13: PSD analysis on the active textile electrode signal shows no significant peak at power line frequency.

Figure 4.14: PSD analysis on the active textile electrode signal shows no significant peak at power line frequency; zoomed in.

4.2.3 Discussion

By comparing the PSD graphs of the remote and active textile electrodes, it becomes clear that using the active textile electrode is more effective in eliminating
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power line interference than using noise reduction techniques (RLD, notch) with the remote textile electrode.

As a side observation, the power of the lower frequencies (< 20 Hz) is also observed to be lower in the active textile electrode sEMG recording versus the remote textile electrode sEMG recording. Since motion artifact is typically observed in the range of the lower frequencies especially approaching DC, this observation suggests that the active textile electrode also better attenuates the low frequency signals that can show up as a result of movement or cable interference (as will be investigated in the following section (4.3).

An observation regarding the power line peak frequency observed on the remote textile electrode PSD is also made that suggests the implementation of the digital notch filter is not entirely accurate. By observing Figure 4.12 it can be seen that the power line peak observed on the remote textile electrode sEMG signal is not centered around 60 Hz, but rather a frequency closer to 61.8 Hz. This does not imply that the power line frequency is actually 61.8 Hz, but rather implies that our sampling frequency is actually incorrect and as a result, an actual 60 Hz signal is manifested as an apparent 61.8 Hz signal on our electronics. To apply the digital notch filter, the limitations of our microprocessor’s clock rate obligated us to use varying sampling periods in order to achieve an averaged, effective 600 Hz sampling frequency. Currently, a number of two different sampling periods are used such that when averaged will result in 600 Hz. In Figure 4.12 however, an obvious attenuation is
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seen at the apparent 60 Hz frequency. This means the digital filter in itself is effective
and is attenuating the desired power line frequency. Due to our sampling error,
however, the filter is attenuating the signal closer to the actual 58.8 Hz frequency;
because the notch filter isn’t extremely sharp, the immediate surrounding frequencies
(60 Hz included) are also attenuated, and power line noise is still reduced. A more
robust combination of these two sampling periods that give us a more accurate average
will need to be used in the future to bring the sampling frequency closer to the true
600 Hz in order to attenuate the proper notch frequency.

4.3 Induced Current Cable Interference

The noise characteristics of the remote 2C textile electrode and active 2C tex-
tile electrode prototypes were investigated when exposed to induced current cable
interference.

4.3.1 Methods

The same electrode setup as mentioned in the previous section was used for test-
ing the performance of the electrodes in the presence of cable interference. In this
experiment, both the RLD and notch were implemented at all times. Again, prior
to inducing interference, a 75% MVC contraction was first sustained for 4 seconds,
followed by a 4 second rest period. This was done to show the relative magnitude of
CHAPTER 4. EXPERIMENTAL RESULTS

the interference to the sEMG signal. To induce cable interference, the cable was hit 4 times with a piece if silicone. Because silicone is an insulator, it possesses a static electric field that can induce an interference displacement current in the cable. If the cable is susceptible to this type of interference, the displacement currents will create voltage errors on the sEMG signal.

4.3.2 Results and Discussion

As can be observed in Figure 4.15, the interference induced on the cable of the remote 2C electrode is significant and over three magnitudes greater than the sEMG signal itself. Both the raw sEMG and envelope of the signal are shown in order to show how this interference affects both signal output types. With the active 2C electrode, this interference is completely absent. These results suggest that the low-impedance cable of the active textile electrodes is not susceptible to voltage errors induced by the static field. As such, the results imply that an active electrode design is imperative to prevent any interference on the electrode cables once they are within a prosthesis.
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Figure 4.15: Cable interference comparison on the remote and active textile electrode signal shown on the (a) high-pass filtered raw EMG signal and (b) filtered envelope signal.

4.4 Envelope Quality (SNR)

This experiment was meant to investigate the quality of the envelope signal obtained using the active 2C textile electrodes using an envelope signal-to-noise ratio (SNR) test. This envelope SNR experiment was not used to look at signal-to-noise ratio in the traditional sense where “signal” is power of the observed sEMG above
the baseline and “noise” is the power of the baseline caused by environmental noise. For that purpose, the noise analysis shown in the previous section has proved that noise caused by power line or cable interference is effectively minimized using the active textile electrode prototype. This is further enforced by the observations that the baseline of the envelope of the sEMG signal is a flat zero when the electrode is placed on a subject’s arm and the subject is at rest.

Instead, this envelope SNR test is meant to quantitatively compare the quality of the sEMG envelope obtained for the purpose of direct control. The “cleanliness” of the sEMG envelope signal is essentially the variance of the envelope signal at a constant force. This is used as a measure for envelope SNR, which is calculated by comparing the normalized sEMG signal to a normalized force reading using the following equation:

$$SNR_{env} = \frac{\sum(Force)^2}{\sum(Force - EMG)^2}. \quad (4.3)$$

Based on equation 4.3, the closer the sEMG signal resembles the force signal, the higher the envelope SNR. While sEMG is not a direct representation of the force generated, when using amplitude-based direct control, the sEMG signal is used as an estimator of force in order to properly drive the prosthesis. This is especially true with myoelectric devices that use proportional control [132]. As such, the envelope SNR test is more a functional measure of the sEMG envelope quality for direct control. From an absolute standpoint, the envelope SNR value is not necessarily meaningful. From a relative standpoint, however, the comparative results can be used to com-
CHAPTER 4. EXPERIMENTAL RESULTS

pare the envelope quality obtained using conventional Otto Bock electrodes to that obtained using the active 2C textile electrodes.

Due to the nature of this test, the envelope SNR measure is thus affected both by the electrode design itself as well as our signal processing. It is important to note that over-filtering could also result in a cleaner envelope and result in an apparent “better” overall envelope SNR reading. On the other hand, because the filtering is being achieved digitally and not with an analog filter, over-filtering could incur delays in the signal and result in inaccurate sEMG. As such, the degree of low-pass filtering for our electronics was chosen in a way as to not incur a visible delay on the envelope signal that would interfere with hitting standard sEMG triggers discussed in the next experiment (Cue Test).

4.4.1 Methods

For this IRB-approved study, four subjects were asked to perform contractions at three levels of force using a conventional Otto Bock electrode and an active 2C textile electrode, and both the force and sEMG were recorded. A primer-bulb attached to a pressure sensor was used as our measure of force generated. The electrode location was determined by trying to pick a spot on the forearm flexors that would best acquire the sEMG signal when the bulb was squeezed. An extra textile 1C reference electrode was placed on the forearm for the textile electrode trials.

The experiments began by putting an Otto Bock electrode on the flexors without
skin preparation underneath a cuff. The subject was asked to perform three maximum voluntary contractions (100%) with 2 minutes of rest in between. An average reading of the sEMG was then taken and used for normalization of the sEMG signal. An averaged maximum force reading was taken concurrently and used for normalization of the force signal.

To set the same gain across all electrodes, the Otto Bock trial was performed first and the electrode’s gain was adjusted based on a desired set maximum. The Otto Bock electrode is active and packaged with a potentiometer on the back that adjusts the analog gain on the electrode’s signal. As such, the gain was adjusted on the Otto Bock by turning the potentiometer, and by trial and error was chosen to achieve a maximum average sEMG envelope output of 2.5V to avoid clipping of the signal on our analog-to-digital converter (ADC).

Once the maximum values were recorded and the gain set, subjects were asked to generate 10 repetitions of 25%, 50% and 75%MVC contractions. The contractions were held for 4 seconds each and interspersed with 20-second rests. Five minutes of rest were given between each force level of contractions to prevent fatigue. Twenty minutes of rest were also given between the Otto Bock electrode trials and the textile electrode trials.

For testing the textile electrodes, an active 2C textile electrode was placed on the same location as the Otto Bock electrode, again without any skin preparation and underneath a silicone cuff. The subject was also asked to perform three 100%
contractions. Based on the averaged maximum value previously obtained by the Otto Bock electrodes, a digital gain was then applied to scale up the textile electrode maximum sEMG signal to the same value as the Otto Bock electrodes. By doing so, we are also putting under scrutiny how well our digital gain can achieve the equivalent analog gain provided by the Otto Bock electrode.

To analyze the results, both the sEMG and force signals were normalized to their maximums. A constant-force window of 2 seconds for each 4 second contraction was used to compute the envelope SNR based on equation 4.3.

### 4.4.2 Results

The results of the SNR test were non-parametric. As a result, we were obliged to use a non-parametric test of comparison, of which we chose the Wilcoxon Rank Sum test. This statistical test essentially ranks all the SNR observations of both electrode types in order and applies an algorithm to test whether or not the observations could be from the same population by chance. Based on the Wilcoxon test and the pooled results of all 4 subjects, the envelope SNR obtained using the textile electrodes was determined to be significantly different than the envelope SNR obtained using the Otto Bock electrodes ($p < 0.01$). Overall, the envelope SNR values of the textile electrodes were lower than the envelope SNR values for the Otto Bock electrodes, suggesting that the envelope quality of the textile electrodes is not comparable to the Otto Bock envelope quality. The p-values generated by comparing the data from
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<td>50% MVC</td>
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<td>75% MVC</td>
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Table 4.2: Wilcoxon rank sum p-values for SNR test results.

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<th>75% MVC</th>
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<td>No</td>
</tr>
<tr>
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<td>Yes</td>
<td>Yes</td>
<td>Yes</td>
</tr>
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<td>Yes</td>
<td>No</td>
</tr>
<tr>
<td>AB4</td>
<td>No</td>
<td>Yes</td>
<td>Yes</td>
</tr>
</tbody>
</table>

Table 4.3: Significance table of Wilcoxon rank sum p-values for individual subject SNR results.

Each contraction level are displayed in Table 4.2.

In attempt to see if this could potentially be subject-based, the Wilcoxon test was also run on the individual results for each subject using a significance level of 5% ($p < 0.05$) (Table 4.3). No obvious pattern was seen either, and with such a small sample size, it was assumed that the pooled results offered a more accurate representation of the overall results.
4.4.3 Discussion

When looking at a few examples of subjects’ contractions using the textile electrodes and the Otto Bock electrodes, it can be understood why the observed envelope SNR is lower for the textile electrodes. In Figure 4.16, a pair of sEMG envelopes representing 75%MVC contractions is shown from subject AB3. Figure 4.16(a) shows an sEMG envelope obtained with the active textile electrode with a digital gain of about 16. Figure 4.16(b) shows an sEMG envelope obtained with the Otto Bock electrode. The blue represents the sEMG signal, the red the force and the black dashed line the normalized max level.

![EMG envelopes](image)

Figure 4.16: sEMG envelope at 75%MVC from subject AB3 obtained with (a) the active textile electrode and (b) the Otto Bock electrode. The blue represents the sEMG signal, the red the force and the black dashed line the normalized max level.

As can be observed by this example, the overall variance of the sEMG envelope
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is greater with the active textile electrode than the Otto Bock electrode. This is representative of the resulting lower SNR observed. The reason for this greater variance can be attributed to the use of a high digital gain in conjunction with a low analog gain. The amplifier chip used in our central electronics has a low maximum analog gain of 12. To make up for this low gain, a 24-bit high-resolution ADC is on board. The trade-off, however, does not seem to be sufficient as can be observed in this particular example and as is suggested by the overall results of the envelope SNR test. The apparent sEMG signal is discrete-like and suffers from low resolution. As such, because the amplitude of the actual signal is very low, we are obligated to use a high digital gain, resulting in a “noisy”, digitized sEMG envelope. The problem with such a wide variance of the signal is that in order to be completely above a set amplitude-based threshold of a myoelectric device, the amputee may have to contract even harder to keep the envelope in its entirety above the threshold.

To address this problem of a lower quality sEMG envelope, it is apparent that a higher front-end analog gain must be used prior to signal processing. A future design direction as is discussed in Chapter 5 and Chapter 6 involves the use of a differential biopotential amplifier on the back of the electrode with a higher analog gain. This will improve the initial resolution of the sEMG signal before it is sent to the central processing electronics and will reduce the need for such a high digital gain. With a higher resolution signal, gain adjustments made in software will not as negatively affect the overall integrity and quality of the sEMG envelope.
4.5 Functional Cue Test

The Cue Test is a functional test that was used to measure how well subjects could achieve four different standard sEMG triggers using conventional Otto Bock electrodes versus the active 2C textile electrode prototypes. The purpose of this pilot study was to test specifically the quality of the active textile electrode in a more functional prosthesis-based test.

4.5.1 Methods

For the Cue Test, electrodes were placed on the forearm flexors and extensors underneath a silicone cuff without any skin preparation. An extra 1C textile reference electrode was placed on the forearm for the textile electrode trials. Locations were chosen in a way that would minimize cross-talk between the flexors and extensors signals. Both electrode types were placed on the same locations. A total of four subjects were required to hit four cues: co-contraction (CC), hold open (HO), double-pulse (2P) and triple-pulse (3P) (as described in section 1.3.1.3). A co-contraction requires the flexion and extension signals to reach a threshold at the same time. A hold open trigger requires that the extension signal stay above a threshold for 3 seconds. The double-pulse and triple-pulse triggers require that the extension signals be pulsed above a threshold twice and thrice, respectively, within a set window. For this test, this window was set to 800 milliseconds and the maximum pulse duration set to 175
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milliseconds. The cues were presented in a random order five times each. No cue was repeated twice in a row. Two subjects performed the Cue Test using Otto Bock electrodes first and two subjects performed the test using the active textile electrodes first. This was done to eliminate any bias that may be introduced practicing with one electrode first versus the other. The subjects all had a window of 30 seconds to hit each cue, and the timer button was pressed by the assessor after the hand fully performed the grip associated with the cue/sEMG trigger presented. If the subject could not perform the correct trigger within the 30 second period, the cue was considered to be “incomplete” and considered a failed attempt.

4.5.2 Results

The results of the Cue Test were non-parametric as can be seen in Figure 4.17. As a result, we were obliged to use a non-parametric test of comparison, of which we again chose the Wilcoxon Rank Sum test. This statistical test ranks all the observations of the successfully achieved cues from both electrode types in order and applies an algorithm to test whether or not the observations could be from the same population by chance.

By simply looking at the plot, we can observe that in general, the distribution of times (first - third quartile) is wider for the pulse triggers than the co-contraction triggers. This can be attributed to the fact that in general, the subjects were observed to have more difficulty overall hitting the pulse triggers. The median time required
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Figure 4.17: Box plot showing distribution of time needed to achieve four cues using Otto Bock and textile electrodes.

to perform the co-contraction or hold open trigger was relatively similar across both electrode types, while for the pulse triggers, the median time required with the textile electrodes was observed to be higher. Furthermore, the distribution of times was clearly wider for the textile electrodes when considering the pulse triggers.

Using the Wilcoxon Rank Sum test with a significance level of 5%, we determined if these observed differences were statistically significant (Table 4.4).

Observing the results, we see that for three out of the four trigger types, there was no significant difference in the times needed to achieve the cue. More specifically what this means is that based on the Wilcoxon Rank Sum test and the non-parametric distribution of times collected, the times obtained using the textile electrodes were not different enough to be considered as part of a separate population at the significance level of 5%. While this does not necessarily prove equivalence, it does suggest that
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<th></th>
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<tr>
<td>Hold open</td>
<td>0.0583</td>
<td>No</td>
</tr>
<tr>
<td>Double-pulse</td>
<td>0.0363</td>
<td>Yes</td>
</tr>
<tr>
<td>Triple-pulse</td>
<td>0.1687</td>
<td>No</td>
</tr>
</tbody>
</table>

**Table 4.4:** Wilcoxon rank sum p-values for Cue Test results.

The effort needed to achieve the co-contraction, hold open and triple-pulse triggers with the textile electrodes is comparable to the effort needed while using Otto Bock electrodes. The double-pulse trigger times obtained with the textile electrodes on the other hand, were shown to be significantly higher than the times obtained with the Otto Bock electrodes ($p < 0.05$).

When looking at actual successful cue completion, the results are promising. The graphs in Figures 4.18 and 4.19 show what percentage of total presented cues of a certain type is achieved at different points in time within the 30 second window. It is observed that 95-100% of all the presented cues could each be achieved within the 30 second window using both electrode types. The results of the co-contraction and hold-open times are shown to be similar as is also suggested by the Wilcoxon test (Fig. 4.18). Observing the double-pulse and triple-pulse graphs, it can be observed that the disparity in times is greater, especially for the double-pulse trigger (Fig. 4.19). This reinforces the results given by the Wilcoxon test.
Figure 4.18: Cue completion time for (a) co-contraction and (b) hold open triggers.
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Figure 4.19: Cue completion time for (a) double-pulse and (b) triple-pulse triggers.
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<table>
<thead>
<tr>
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<tr>
<td>Co-contraction</td>
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<td>6s</td>
</tr>
<tr>
<td>Hold open</td>
<td>6s</td>
<td>8s</td>
</tr>
<tr>
<td>Double-pulse</td>
<td>8s</td>
<td>14s</td>
</tr>
<tr>
<td>Triple-pulse</td>
<td>10s</td>
<td>16s</td>
</tr>
</tbody>
</table>

Table 4.5: Time required to achieve the desired cue with 80% confidence.

In Table 4.5, times are listed that suggest the time needed to achieve each trigger with 80% confidence. Again, this suggests that the times needed with the textile electrodes to hit the co-contraction and hold open triggers are more comparable to the time needed with Otto Bock electrodes than the pulse triggers.

4.5.3 Discussion

In general, the amount of time needed to achieve the co-contraction and hold open triggers was less than the time needed to achieve the pulse-triggers. This is due to the fact that the co-contraction and hold open triggers themselves are much more straightforward in comparison to the pulse triggers. Challenges for subjects were generating pulses that were short enough in duration and also generating the right number of pulses within the set window. The Wilcoxon Rank Sum test suggested that the times needed to achieve the co-contraction, hold open and triple-pulse triggers using the textile electrodes versus the Otto Bock electrodes are from the same
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population, and as such, are not significantly different. As can be seen by observing Figure 4.17, however, the distribution of times using the textile electrodes was still wider than the distribution of times using the Otto Bock electrodes. This can possibly be attributed to the prototype nature of the textile electrodes used for this experiment. It is important to note that due to project time constraints, the active textile electrode prototype used for the test was not optimal (pictured in Fig. 3.13). In this prototype, the packaged pre-amplification circuit was located slightly downstream from the electrode itself and not directly on the back of the electrode. As such, the proximal portion of the electrode and cable (upstream of the amplifier) was still very much susceptible to interference and motion artifact, especially if the packaged amplifier circuit was moving. These movements were sometimes misinterpreted as triggers of a different type and would sometimes send the hand into a the wrong grip, increasing the overall time needed to achieve the trigger.

This observation could also explain why the double-pulse trigger times were significantly longer with the textile electrodes than with the Otto Bock electrodes as suggested by the Wilcoxon Rank Sum test. The reason being that due to the limitations of the active textile electrode prototype, the proximal cable motion artifact would create extra pulses as the subject’s arm retracted from a trigger attempt. Due to these possible extra pulses, the resulting signal could transform a wanted double-pulse trigger into an unwanted triple-pulse trigger.

In summary, these preliminary results suggest the with some improvements, the
textile electrodes could be comparable to the conventional Otto Bock electrodes when considering functionality with the use of a directly-controlled trigger-based prosthesis. In terms of the actual ability to hit triggers, the results show that within 30 seconds, 95-100% of the triggers can be achieved. Time required to hit three out of the four triggers were also shown to be comparable. This experiment must be undoubtedly repeated with the next active textile electrode prototype which houses the amplifier directly on the back of the electrode. With this design, unwanted interference that is currently observed on the cable proximal to the packaged amplifier circuit will become negligible. This could then also drastically improve the reliability of the textile electrode and decrease the overall time needed to achieve the cues.

4.6 Pattern Recognition

An IRB-approved pilot study was also conducted to obtain preliminary results of the textile electrodes in a pattern recognition system. The pattern recognition system used was developed at Johns Hopkins University (Baltimore, MD, U.S.) in collaboration with Infinite Biomedical Technologies. The system uses an LDA classifier to analyze sEMG data collected from eight different channels of electrodes to process what sEMG patterns the subject is generating and to predict what movement he or she is performing. To use the system, a subject must first position the eight channels of electrodes accordingly and then train the classifier during one or more
training sessions. For an able-bodied subject or an amputee who has not undergone targeted muscle reinnervation (TMR) surgery, these electrodes are typically evenly spaced circumferentially around the (residual) limb below or above the elbow. For an amputee who has undergone TMR, these electrodes are positioned on and around the reinnervated sites. Prior to starting the training session, the subject connects to an associated graphic user interface (GUI) and selects which cues he or she would like to perform and use with his or her system. During the training session, the GUI displays the the subject chose in a random order a set number of times and the subject must attempt to perform and hold these cues as they are displayed on the screen.

The LDA classifier computes the needed analysis during training and then fits the training data against the classifier, determining the probability of the provided cue matching the predicted cue based on the training data. The classifier uses three features of the sEMG to do this: mean, variance and waveform length. After a training session is completed, a confusion matrix is generated. A confusion matrix is meant to represent the apparent separability of the patterns generated during the training data. A perfect confusion matrix will have matched provided and predicted cues, achieving a probability of 1. During use, the system continuously processes sEMG in real-time and makes a prediction as to what the user is attempting to achieve. While separability can be apparent by observing values of the confusion matrix, one or more evaluation sessions are needed to verify the ability of the subject to essentially reproduce the training data and generate the cues accurately as if he
or she were using a prosthesis [133].

An evaluation session is similar to a training session in the sense that cues are displayed on the screen that the subject must perform and hold for the duration the cue is presented. In the evaluation sessions, all cues aside from “rest” are presented a set number of times randomly throughout the session and are interspersed with a “rest” cue. The outputs from an evaluation session are cue completion percentage, scaled cue completion time and accuracy. Cue completion percentage represents the percentage of total cues presented that the subject was able to achieve at any point during the cue presentation. Scaled cue completion time is the ratio of the time it took for the subject to achieve a cue once it had been presented versus the minimum time needed to achieve that cue. The best scaled time can be 1.0, meaning the subject performed the cue within the minimum time allotted for it. Accuracy focuses more on the ability for the subject to achieve and hold a cue. To calculate accuracy, the total time a cue of interest is achieved by the subject while the cue is presented is divided by the total time the cue was presented. A pooled accuracy is then reported which represents the accuracy of hitting and holding all the cues that were trained.

### 4.6.1 Methods

The subject for this pilot study was chosen because the subject was familiar with the training procedure and pattern recognition system. For the conventional interface, eight triplets of remote metal dome electrodes by Liberating Technologies, Inc. were
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embedded evenly-spaced circumferentially in a silicone cuff and worn over the forearm. Eight remote 2C textile electrodes were placed evenly-spaced circumferentially underneath a silicone cuff with a separate textile reference electrode for the textile interface. The subject was allowed up to 20 minutes per electrode type for practicing and becoming familiar with the electrode interface and the pattern recognition system. This was to reduce any learning that could occur throughout the trials. The subject then performed a training session using six cues: rest, hand open, hand close, (wrist) rotate in, (wrist) rotate out and fine pinch. The training session presented each cue in a random order three times each for five seconds. The first second of data was always trimmed. A confusion matrix was then generated at the end of the session.

At the end of the training session, the subject was asked to perform two evaluation sessions where the subject was presented the non-rest cues three times each for five seconds each in a random order and interspersed with rest. The subject was allowed to rest for at least five minutes between sessions to prevent fatigue.

At the end of the evaluation sessions, the subject was allowed to rest for up to 10 minutes and a second training session was performed.

4.6.2 Results

The confusion matrices generated by the first training sessions with the metal dome and textile electrodes are shown in Figures 4.20 and 4.21, respectively.
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Figure 4.20: Confusion matrix for first training session using conventional LTI electrodes.

A “perfect” confusion matrix would be all white down the diagonal, representing perfect match of provided to predicted cue. As is observed with both electrode types, the probability of matched provided and predicted cues for all six cues is determined to be at least 93%. This suggests that separability of patterns is not only possible but also comparable to conventional electrodes with the textile electrodes.
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![Confusion Matrix for Training Session](image)

**Figure 4.21:** Confusion matrix for first training session using remote textile electrodes.

The same observations are made for the second training session with each electrode type (Figures 4.22 and 4.23).
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Figure 4.22: Confusion matrix for second training session using conventional LTI electrodes.
Figure 4.23: Confusion matrix for second training session using remote textile electrodes.
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<table>
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<th>Metal Dome</th>
<th>Textile</th>
<th>% Difference</th>
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<td>0</td>
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<tr>
<td><strong>Accuracy %</strong></td>
<td>80.65</td>
<td>92.20</td>
<td>+12.53</td>
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Table 4.6: Cue completion % and Accuracy % during pattern recognition evaluations sessions using conventional LTI electrodes and remote textile electrodes.

Results of the evaluation sessions were pooled and are presented in Table 4.6 and 4.7. In terms of cue completion percentage of all presented cues, the subject’s performance is identical across both electrode types, achieving 28 out of the 30 total presented cues. Accuracy, which represents the ability for the subject to hold the presented cue, was observed to be 12.53% better when using the textile electrode interface. This could suggest that better accuracy is achieved using the textile electrode interface due to the fact that the interface is more conformal to the subject’s arm and can maintain solid contact during each movement.

Results regarding the average scaled cue completion times across all presented cues are presented in Table 4.7. The relative difference in the scaled cue completion time for textile versus metal dome electrodes are also presented in the last column. For three cues (hand close, hand open and rotate in), the subject needed more time to achieve the cue using the textile electrodes versus the metal dome electrodes. The time needed to achieve the cue was over 60% longer for all three cues. For rotate out and fine pinch, however, the performance seemed to be better with the textile electrodes, and the completion time was observed to be 10% faster using the textile
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<table>
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</tr>
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<td>Rotate In</td>
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<tr>
<td>Fine Pinch</td>
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<tr>
<td>All Cues</td>
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<td>+23.02</td>
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Table 4.7: Scaled cue completion time of cues during pattern recognition evaluations sessions using conventional LTI electrodes and remote textile electrodes.

Electrodes versus the metal dome electrode. When observing all cues overall, the cue completion time for the pooled cues is observed to be about 23% greater with the textile electrode interface than with the conventional metal dome interface.

4.6.3 Discussion

It is important to first note that the data presented in this pilot study represents a very small sample size of \( n = 1 \). For a better and more insightful comparison, more subjects who are familiar with the pattern recognition system are needed. It is also important to emphasize that these results were obtained using the remote textile electrode prototypes, and as such, the electrodes were susceptible to environment noise. Again, for a more insightful comparison, active textile electrodes will need to
be used in the future.

In general, it took longer to achieve the cues using the textile electrodes versus the metal dome electrodes. Again, the fact that remote and not active textile electrodes were used for this study could have contributed greatly to this observation. Due to the fact that the electrodes were susceptible to movement and interference, external factors such as power line interference or motion/cable artifact generated by switching to a grip could possibly explain why it took longer for the sEMG patterns to stabilize initially. Furthermore, this could explain why for the textile electrodes, there is apparent confusion (although minimal) between rest and other presented cues. Apparent noise on the signal could prevent a stable baseline and prevent a “perfect” rest pattern.

Overall, however, the study shows the potential of these textile electrodes in a functional pattern recognition system. The cue completion percentages were comparable, suggesting that patterns could successfully be decoded with the sEMG signals obtained using the textile electrodes, and by using the active textile electrode prototypes in the future system, the other inconsistencies can potentially be addressed.

4.7 Summary

This chapter presents experimental results from six different experiments. The first experiment investigated the impedance of the electrode-skin interface using the
remote textile electrode prototypes. The results suggest that with sweat present at
the electrode-skin interface, the impedance may be comparable to the electrode-skin
impedance of conventional metal dome electrodes. Spraying down the arm with tap
water can also significantly lower the interface impedance within 5 minutes. When
using a prosthesis with these electrodes, spraying down the arm first with tap water
as a skin preparation step can be an easy and accessible solution for an immediate
and initial decrease in the interface impedance. With a silicone liner, the moisture is
expected to be retained until sweat inevitably begins to accumulate at the surface.

Noise analysis investigated the performance of the active textile electrode proto-
types in rejecting power line interference and induced current cable interference. Both
experiments suggested that while the remote textile electrode prototypes are clearly
affected by these sources of interference, the active textile prototypes could effectively
reject both sources of interference, rendering their effect on the sEMG signal as neg-
ligible. This suggests that using an active textile electrode design will be necessary
for obtaining good noise rejection characteristics in the final system.

Envelope SNR was investigated to analyze the quality of the sEMG envelope
signal obtained with the active textile electrode prototype and our central processing
electronics. The results suggested that the quality of the active textile electrode
envelope is not matched to the quality of a standard Otto Bock signal envelope.
By analyzing the results and data collected, it is observed that our low analog gain
coupled with a high digital gain resulted in a “noisy” envelope. As such, moving
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forward with the next active textile electrode prototype, a higher front-end analog gain must be applied to ensure for a good envelope SNR.

The last two experiments tested the performance of the textile electrodes in a functional setting. The Cue Test was used to test the active textile electrode prototypes for use with a direct-controlled prosthesis by requiring the subjects to hit certain sEMG triggers using the active textile electrodes and conventional Otto Bock electrodes. The results suggested that using the textile electrodes, all four triggers could be achieved. Three out of the four cues were also achieved within a time frame comparable to that achieved with Otto Bock electrodes. The overall distribution of the times was greater for the textile electrodes, suggesting inconsistencies in the sEMG signal. These inconsistencies can be attributed to the limitations of the active textile electrode prototypes, which were still susceptible to some interference proximal to the active circuit. This strongly suggests that for the next future prototype iteration, the components must all be placed directly on the back of the electrode prior to sending the signal down through the TDC.

The pattern recognition case study was used to test the textile electrode prototypes with a pattern recognition system. The results showed that separability of patterns during a training session is achievable using the remote textile electrodes. In comparison to the results obtained with conventional metal dome electrodes, the time needed to achieve the cues was in general greater for the textile electrodes, although the accuracy achieved was also greater. Overall, the study does show potential for the
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use of the textile electrodes in a pattern recognition system, but it is greatly weakened by the limitations of the remote textile electrodes because of their susceptibility to noise and the small sample size of the study (n = 1). In the future, this study must be repeated with active textile electrodes and a larger number of subjects to produce more informative comparative data.
Chapter 5

MyoLiner Integration

The previous sections have focused more heavily on the characteristics and performance of the individual textile electrodes. This section describes the envisioned integration process needed to eventually interface the textile electrodes with a myoelectric prosthesis. We’ve termed this integrated solution as the “MyoLiner.” The MyoLiner design concept representing the final product is discussed in this section. In the following Future Work chapter, the steps needed to move towards this final integrated solution are also laid out.

5.1 Overall Vision

More on each component of the full MyoLiner system will be discussed in the following sections, but first the fully-integrated design concept is presented. The
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overall vision of the final system is displayed in Figure 5.1. In addition to providing a flexible, secure electrode interface and increased suspension, one of the main design goals for the entire system is to minimize the number of connections and parts needed to use the MyoLiner, such that the device can be straightforward and easy to use for the amputee.

Figure 5.1: Overall vision of the MyoLiner system.

In this model, an integrated liner and electromechanical dock is pictured (Fig. 5.1). The integrated liner will be embedded with the desired number of active textile electrodes, their textile data cables and a distal pin attachment that will interface the integrated liner with the prosthesis shell. The electromechanical dock will then interface the liner with the prosthesis both mechanically and electrically.
CHAPTER 5. MYOLINER INTEGRATION

5.1.1 The Integrated Liner

The integrated liner houses the active textile electrodes and their cables and the distal attachment (Fig. 5.2).

![Figure 5.2: The integrated liner.](image1)

The active textile electrodes will be silicone-backed during their manufacturing process, with the fabric cable (TDC) emerging from the center of the electrode module as pictured in Figure 5.3.

![Figure 5.3: Silicone-backed electrode prototype with TDC emerging from center of the module.](image2)
CHAPTER 5. MYOLINER INTEGRATION

With this configuration, the embedding process will be made easier for the prosthetist. One slit will need to be made per electrode in order to string the TDC through. The silicone-backing then allows for a simple silicone-to-silicone attachment on the inside of the liner. By attaching the electrode module to the inside of the liner, the slit is also then sealed from the inside. This maintains suction and suspension of the liner itself. This attachment process was attempted with an in-house fabricated silicone liner and the integrity of the seal was tested (Fig. 5.4). As can be observed in Figure 5.4(b), an air pocket is formed when the open end of the liner is sealed, suggesting that the attachment points on the liner where the cable is drawn through are also sealed.

Figure 5.4: (a) In-house experimentation with embedding the textile electrodes in a roll-on silicone liner; as is suggested by the second figure (b), a seal is maintained via the proposed attachment process.
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Due to the fact that active textile electrodes will be used, the TDCs will be minimally susceptible to noise. In order to prevent the wear of the fabric on the TDC itself, however, the cables can either be stitched underneath an outer fabric typically seen on a liner or even embedded into the silicone liner itself.

The distal attachment will be a permanent one. The purpose of this attachment is to provide a locking mechanism into the prosthesis, to provide a central “hub” for the electrode cables, and to house the pre-processing electronics needed to multiplex the electrode signals. In the modeled system (Fig. 5.1), the mechanical attachment is a pin lock. The electrodes, once embedded into the liner, will then be permanently plugged into this attachment.

The ultimate goal would be to generate amplified, raw sEMG signals from each active electrode. These signals will then be sent to a small microcontroller with an on-board eight-channel analog-to-digital converter (ADC) housed in the digital attachment. This will allow for the signals to then be digitized, multiplexed and sent to the central electronics via one to two lines of communication as is used with SPI or I2C. As such, only four points of contacts are needed when communicating with downstream processing electronics (two for Power/Gnd, two for data transfer). Four spring pins on the distal attachment (Fig. 5.2) will then interface with four rings on a mating ring board PCB located on the electromechanical dock. This will allow the digitized signals to be sent to the central electronics and power to be supplied to the active textile electrodes. All the electronics in this distal attachment will also be
properly sealed and protected from exposure to moisture.

5.1.2 The Electromechanical Dock

The electromechanical dock will be a permanent fixture in the distal part of the shell and will secure the liner into the shell mechanically, using a pin lock on the distal end of the liner and a ratchet system within the electromechanical dock (Fig. 5.5). A key or button can be used to then release the pin lock (a key is used in this example). A prototype we developed of this dock that can mechanically withstand up to 30 lbs of weight is shown in Figure 5.5.

![Electromechanical dock prototype](image)

**Figure 5.5:** Electromechanical dock prototype: (a) internal components of the ratchet lock and (b) 4-contact ring board.

The dock will also house the central processing electronics. A “ring board” PCB on the top side of the dock will make contact with spring pins on the distal end of the integrated liner when the liner is plugged into the dock. When these pins make contact, they will deliver power to the liner and transfer data via a standard com-
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Communication protocol (SPI/I2C) in order to send the signals to the central electronics. The rest of the electronics will also be housed within the dock, and appropriate outputs will then be connected to the prosthetic device being used. Essentially, the final design will operate in a “plug and play” fashion: once the integrated liner is plugged into the dock, it will be secured into the shell and also interfaced with the electronics of the prosthesis.

5.1.3 The Central Electronics

The central electronics are responsible for post-processing the signal and choosing the appropriate output for the desired prosthesis application. A custom development board designed in collaboration with Infinite Biomedical Technologies (IBT) known as the CX-3 is used as the basis for the hardware. The digitized signal received from the integrated liner will first be high-pass filtered to remove any low-frequency motion artifact. A digital notch filter will be added if necessary. If the raw signal is needed for a pattern recognition system, then the processing stops here. If the envelope of the signal is needed for driving a myoelectric device using direct control, then additional rectification and low-pass filtering will be done to obtain the sEMG envelope. In both cases, a digital gain can be added to scale up or down the output signal as needed using a graphic user interface (GUI) software. This data workflow is summarized in section 5.1.4.

The CX-3 has an on-board digital-to-analog converter (DAC) and four analog
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outputs, meaning it can output up to four out of the eight channels of filtered and amplified sEMG. In addition, the CX-3 can output I2C, SPI and UART (Bluetooth) data, as well as other analog and digital signals needed for driving prostheses motors directly. For the purposes of two-site control, two channels of processed EMG envelopes corresponding to electrodes placed on two antagonist muscles can be outputted to the prosthesis via the DAC. For the purpose of pattern recognition, Bluetooth or I2C communication can be used to communicate with the pattern recognition software and prosthesis, respectively.

5.1.4 Data Workflow

The data workflow described in the previous section is depicted in Figure 5.6.

![Figure 5.6: Data workflow in the integrated system.](image-url)
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The diagram is divided into three components boxed out in blue dotted lines: the liner body of the integrated liner, the distal attachment of the integrated liner and the central electronics housed in the electromechanical dock. As can be observed on the diagram, sEMG data from the embedded active electrodes is sent via the textile data cables (TDCs) to the distal attachment. Within the distal attachment the signals are digitized and multiplexed before sending them to the central electronics’ microcontroller. The signals first pass through a high-pass filter (HPF). If the system is configured to work with a pattern recognition system (PR), then the signals are sent through the LDA classifier and pattern recognition code. From there the appropriate output module is used to transmit the signals needed to control the prosthesis. If the system is configured for direct control, the high-pass filtered data is then rectified and low-pass filtered (LPF) to obtain the sEMG envelope. From there the signals are sent to a digital-to-analog converter (DAC) and outputted to the prosthesis.

5.2 Using the MyoLiner

Another design goal is to make the MyoLiner system easy to configure for the prosthetist. From a prosthetist standpoint, an aim would be to minimize the number of steps the prosthetist must take during the fabrication and fitting process. Whether the prosthetist would like to embed the electrodes himself or herself or have a custom liner pre-fabricated may vary from one case to another. Ultimately however, once
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proper locations are determined, the textile electrodes will be embedded into the liner and plugged into the distal attachment permanently. The textile electrodes now become a permanent fixture of the integrated liner and are no longer manipulated by the prosthettist. To adjust the gain or choose from multiple channels, the prosthettist will simply need to connect to the compatible GUI software (Fig. 5.7).

![Preliminary MyoLiner GUI](image)

**Figure 5.7:** Preliminary MyoLiner GUI.
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Figure 5.7 shows the preliminary software used currently. As can be seen there are options to select Flex and Extend channels for a dual-site control device, to send a digital gain adjustment, and to view the signals of up to three channels on the screen. The final version will offer a clearer interface with more options for channel selection, mode of control (raw vs. envelope), and a signal viewer for all eight channels.

Another design goal is to make the MyoLiner system easy to use for the amputee. From the amputee standpoint, the integrated liner will be a device that can be donned like a typical roll-on silicone liner. Once the liner is donned, the liner can then be plugged into the prosthesis. With the electromechanical dock, only one connection is needed. Once removed from the prosthesis, the integrated liner can then be doffed and hand-washed. Any electronics in the distal attachment or on the back of the electrodes will be entirely sealed and embedded, making the liner a washable component.

To address the possible issue of electrode location alignment with a direct-control myoelectric prosthesis, we’ve also developed an auto-calibration feature. This auto-calibration feature can allow the amputee to re-calibrate his or her integrated liner every time it is donned, or whenever necessary, to choose the optimal electrode sites out of multiple available electrode sites on the integrated liner. With this feature, the amputee will not necessarily need to don the liner in the same manner every time it is worn. Also, if the amputee experiences fatigue over time and would eventually like to switch to different movements/sites for direct control, the auto-calibration feature would allow him or her to reset the sites as desired.
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For example, for a dual-site controlled device, eight electrodes can be evenly spaced around the internal circumference of the liner. When the calibration button is pressed, the user will then be prompted to make two distinct movements (such as flex and extend). Based on those two movements, two channels will be selected out of the available eight that will offer optimal dual-site control with the least amount of crosstalk based on the calibration data. The calibration feature can be extended to also employ an automatic gain adjustment. These added auto-calibration features could help prevent the need for the amputee to visit a prosthetist every time an electrode location or gain adjustment is needed. With the calibration button, the adjustments are done automatically. The preliminary auto-calibration algorithm works for a dual-site controlled device as follows:

1. System collects sEMG envelope data from all $n$ channels during the calibration process.

2. Root-mean-square (RMS) of the sEMG envelope is computed for both movement 1 (RMS.1) and movement 2 (RMS.2) for all $n$ number of channels ($RMS.1_1, RMS.1_2, RMS.1_3 \ldots RMS.1_n; RMS.2_1, RMS.2_2, RMS.2_3 \ldots RMS.2_n$).

3. To choose the best electrode channel based on movement 1, the $RMS.1/RMS.2$ ratio is computed for each channel; the channel with the maximum ratio value is chosen.

4. To choose the best electrode channel based on movement 2, the $RMS.2/RMS.1$ ratio is computed for each channel; the channel with the maximum ratio value is chosen.

5. Automatic gain is applied to both channels to scale up or down signals to a previously specified voltage level.

6. The two selected channels with their applied gain are used for prosthesis control until device is turned off or subject re-calibrates.
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This preliminary algorithm has shown to correctly chose channels for optimal control, however it still needs to be tested for robustness. The algorithm must also be optimized to work for any number of channels and any number and type of control outputs.

5.3 Summary

The concept of the fully-integrated MyoLiner system was developed based on feedback from both prosthetists and users. A common request was to minimize the number of connections that needed to be made during the fitting process and prosthesis donning process. To address this, we’ve separated the MyoLiner system into just two physical parts: the integrated liner and the electromechanical dock. In terms of the proposed attachment process for integrating the textile electrodes into a liner, we have manufactured a proof-of-concept liner suggesting a proper seal can be made when embedding the textile electrodes, preserving proper suction and suspension of the liner. The central electronics developed in collaboration with IBT are also currently able to interface with our textile electrodes and conventional direct-controlled devices as well as the IBT pattern recognition system. Finally, a compatible GUI software and auto-calibration features, although in their preliminary stage, have been developed to more easily interact with and control the MyoLiner system.

Although much progress has been made overall in developing particular aspects
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of the overall design, future work is still needed to finalize the design of this fully-integrated system. The next chapter will discuss in more detail some of the system components we have already designed and their current state, as well as the work needed to take each component to completion.
Chapter 6

Conclusion and Future Work

6.1 Conclusion

To conclude we will review the thesis objectives and how they were met, as well as discuss the implications of our results. To reiterate, the primary objectives of this thesis were to characterize the skin-electrode interface, model and discuss sEMG electrodes and their design, report preliminary results obtained by various experiments conducted using the newly developed textile electrodes and establish the groundwork for a fully-integrated version of the MyoLiner system. This final system will aim to provide the patient with an integrated textile-electrode interface that can be interfaced with his or her prosthesis.

In Chapter 1, we briefly reviewed prostheses types, discussed the existing electrode-skin interface issues with current myoelectric prostheses and proposed the MyoLiner
CHAPTER 6. CONCLUSION AND FUTURE WORK

solution. In the second chapter, we modeled and characterized the the skin-electrode interface as well as the sEMG electrode and its noise characteristics. The overall outcomes of this modeling hopefully better familiarized the reader with these concepts. Design implications for developing an optimal sEMG recording system were also reviewed. These design implications were taken into consideration when designing the electrode as discussed in Chapter 3 and also when designing the impedance and noise analysis experiments discussed in Chapter 4. Experimental results for impedance and noise analysis as well as three more functional-based tests (Envelope SNR, Cue Test, Pattern Recognition) were reviewed in Chapter 4. The experiments laid out in Chapter 4 were meant to generate preliminary results to show the potential of the newly developed textile electrodes and to provide the author with information to better approach the next design iteration.

During the time frame of this thesis, both a remote textile electrode prototype and active textile electrode prototype were developed as described in Chapter 3 (section 3.4). Impedance testing of the remote textile electrodes showed that with the natural presence of sweat that will undoubtedly accumulate within the silicone liner over time or by spraying down the arm with water as part of skin preparation, a skin-electrode impedance similar to that of conventional metal dome electrodes can be achieved. Noise analysis results showed that in the presence of power line and other environmental interference, the active textile electrodes are able to effectively minimize the negative effects, rendering the interference negligible. From a functional standpoint,
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preliminary results obtained using the Cue Test and the Pattern Recognition case study have suggested that the textile electrodes, once finalized, can reliably control a myoelectric prosthetic device.

Finally, the MyoLiner integration and overall vision of the system was laid out in Chapter 5. An extensive Future Work section in this final chapter describes in detail the next steps needed to move forward towards the final design. The results of the envelope SNR test strongly suggest future design improvements be made in the overall active electrode design. These improvements and more are discussed in the following Future Work section.

The experimental results presented in this thesis show promise in terms of acquiring good signal quality with an active textile electrode design. Furthermore, in-house testing of the embedding process into a silicone liner has suggested that the integration into a roll-on liner can be achieved. We were also able to successfully interface with conventional prostheses, both direct-controlled and pattern recognition-controlled, validating the design of our central electronics. Finally, preliminary testing with added features such as the software interface and auto-calibration settings has also shown the potential of making the MyoLiner a system that can be easily controlled by the prosthettist and the amputee. While each aspect of the system has been addressed, there are still a number of steps needed to finalize their development in order for the MyoLiner system to be a full, working solution. These steps are laid out in the next Future Work section.
6.2 Future Work

6.2.1 Electrode Design

There are several challenges we face with the most current design. The recently manufactured prototypes were hand-made, meaning that openings in the base fabric on the electrode module that expose the contacts were hand-cut. Due to the nature of the cut as well as the nature of the material used (cotton), the edges of these openings begin to fray. Frayed edges can pose a significant problem when the frayed fabric begins to cover part of the contact area. This reduces the effective surface area of the contact touching the skin, increasing impedance and lifting the electrode lightly off the skin. This can also make the electrode even more susceptible to motion artifact. A solution is to laser cut the edges in the future. The heat of the laser cut can create more finished edges, preventing cut edges from unraveling.

Another possible problem with the module is the fact that the contacts are not well insulated from one another. The two contact fabric patches are sandwiched between non-conductive fabric, and are only separated from one another because of their location. Fabric will absorb moisture, however, and eventually, enough sweat can short the contacts via the base fabric. A proposed solution is to add a silicone barrier between the two contacts on the 2C patch. Using the same silicone that back the electrodes, an added silicone barrier in between the contacts can better electrically insulate the contacts from one another in the case of sweat build-up underneath the
electrode.

The durability of the connectors may also pose an issue. The connector on the distal end of the cable achieves the hard-to-soft connection needed for interfacing with the electronics. The life of this connector is limited however, especially if it is to be plugged in and out of electronics frequently. As such, the connection is intended to be a one-time permanent connection, and the proper attachment will need to be defined.

Finally, when moving to the active electrode design, many factors need to be considered. The active solution is the right one in terms of obtaining a good quality sEMG signal with minimal interference. The trade-off, however, is the increased complexity of the overall electrode design. The question arises if the flexibility of the overall module will suffer. The proposed solution is to use a “fabric PCB” or flex PCB with very small (<3mm) components on the back of the electrode to maintain flexibility. Durable and long-lasting hard-to-soft connections will need to be made on the back of the electrode that are strong, but that also do not add rigidity to the overall module. Proper measures will also need to be taken to make sure to insulate the PCB and its components from exposure to moisture that could potentially soak up through the textile. Furthermore, there may be the need to move towards a 3C version. By doing so, a reference is now available at the electrode site, allowing for the use of a differential biopotential amplifier. With this design, an added front-end analog gain can also be added to the electrode signal at the interface site. This
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will promote better resolution of the signal and improve the overall SNR. The added contact in the middle, however, may also pose a problem in terms of flexibility and the textile electrodes achievable bend radius.

6.2.2 Overall Integration

6.2.2.1 The Integrated Liner

Integrating the textile electrodes into a silicone liner in a reliable fashion also needs investigation. Currently, we have been able to open communication with several prosthetists who are trying to embed the silicone-backed electrodes into their current liners or silicone. Open questions about the embedding process include material preference for backing the electrodes, flange or skirt options to make the embedded surface smoother and to promote a tight seal, and embedding options for the fabric TDC. Once the preferred options are determined using their feedback, the next prototype can then be backed with the specific silicone and shape needed to easily attach the electrodes.

6.2.2.2 The Central Electronics

In terms of the electronics, the current interface is in stand-alone form and represents a table-top version of the electronics unit. A low noise, high CMRR 8-channel amplifier chip with 24-bit ADC resolution and a maximum programmable analog gain
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of 12 (ADS1298, Texas Instruments, Dallas, TX, U.S.) is used for acquiring the sEMG differential signals. The textile electrodes plug into mating connectors on a routing board that interfaces the electrodes with the amplifier chip and the CX-3. Outputs are available on the unit for a reference electrode shorted to the amplifier common and a reference electrode connected in RLD fashion.

For the final design, we will be using active electrodes with amplified, differential signals. As such, the 8-channel amplifier chip is no longer needed. The signals can instead be fed directly into a microcontroller for processing. Furthermore, the reference electrode will either become part of each individual electrode (3C version with biopotential amplifier) or simply shorted to the amplifier common. The routing board will also be replaced by the distal attachment on the integrated liner. As such, the changes in electronics and signal processing will go hand-in-hand with the active electrode design.

When moving to the final design, the size of the electronics will need to be minimized and integrated into the electromechanical dock. A stand-alone version can also be an option, where the electronics simply are placed freely in the shell, however size is still of concern.

6.2.3 A Note on Experimental Design

Once a final prototype is achieved, verification and validation of the electrode design is of utmost importance. Conducting studies to prove the signal quality and
functionality of the MyoLiner will be crucial prior to moving from the final prototype to a production run, however. The experiments conducted in this thesis are good for preliminary studies in order to show the potential of the newly developed textile electrodes and device. To make them more robust, however, a statistical power analysis should be done prior to experimentation to decide on how many subjects are needed [134, 135]. Also, choosing the statistical tests to be used ahead of time can also improve the overall design of the experiment itself.

In most cases, we are doing points of comparison between data obtained with a conventional electrode and data obtained with a textile electrode. For this reason, it is of interest to see if the observations collected from various studies across the electrode types are significantly different from one another. For most tests a typical unpaired two-sample two-tail t-test would be sufficient. A two-one-sided-test (TOST) alternatively can also be used to prove significant equivalence based on an arbitrarily set range of equivalence [136]. These are parametric tests, however, and assume normal distribution of the data. For the most part, the studies conducted obtained results that were non-parametric. This is most likely due to the small sample size, but can also be attributed to the nature of the test or the data itself [134, 137]. As a result, enough observations should be made with the conventional electrode type prior to experimentation in order to determine or help make assumptions about the expected distribution of data. By convention, this number is usually at least 30 samples although this number will vary based on the experimental design [138].
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If a parametric distribution can be assumed, statistical analysis software such as R or MATLAB can be used to compute the number of samples needed based on a desired statistical power in order to show significance (R: pwr package, or powerTOST package; MATLAB: sampsizepwr function). Expected or desired deviations from the mean are termed “effect sizes” and are also arbitrarily set. A common parameter used is known as Cohen’s $d$ and is represented by:

$$d = \frac{\mu_1 - \mu_2}{\sqrt{\frac{SD_1^2}{2} + \frac{SD_2^2}{2}}}$$

(6.1)

Several effect sizes can be investigated in order to calculate how many subjects or observations would be needed to prove significance [136]. Same can be done for the equivalence (TOST) test. Based on these numbers, the studies can then be conducted using the textile electrodes.

If a non-parametric distribution is still observed, the experimental design must be investigated to see if there are any confounding factors that could create outliers or a skew in distribution [137]. Otherwise, non-parametric tests should be used to test for significance and more advanced methods are required for power analysis. Some argue that if enough data points are collected, then a power analysis can be done assuming parametric data, and the resulting sample size simply increased by a set factor such as 15% [135].

Aside from the above suggestions for overall experimental design, suggestions for improving the existing Envelope SNR, Cue Test and Pattern Recognition experiments and also additional experiments are discussed in the following sections.
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6.2.3.1 Envelope SNR

The experimental setup of the Envelope SNR test as is designed is ideal, however there may be some flaws in the actual post-analysis of the experiment. For example, to normalize the force, a max force reading is first recorded, and subsequent contractions are divided by this number in order to normalize the force reading to 1. To normalize the sEMG signal, an averaged sEMG value at the subject’s constant-force 100% MVC is used. This max value calculation is done automatically based on the assumption that the subject can hold the same force contraction steady for two seconds. A more manual approach to calculating this normalizing value where the assessor makes sure the 100% MVC contraction is steady may be required to obtain a more accurate number, since it was seen that the sEMG normalized signals did not always match up with the force readings. Furthermore, averaging the EMG signal to obtain a number to normalize the data against may not be entirely accurate, and other options should be explored.

6.2.3.2 Cue Test

One major drawback from the Cue Test as is, is that the time it takes to achieve a cue may not necessarily be directly representative of the effort needed by the subject to generate the needed sEMG trigger. Firstly, the timer isn’t pressed until the grip is completed by the hand. If the grips take different amounts of time to actually be completed by the hand, then the timing will be different across cues even if the
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trigger was generated correctly within the same time frame. This can somewhat be addressed by subtracting the given time it takes for the hand to complete a grip or normalizing the time based on this value, but then it becomes even more complicated when considering time differences based on which grip the hand is switching from. Furthermore, if an incorrect cue is generated, the subject must then wait to see if the grip is correct before proceeding. As such, a more accurate representation of effort may be the number of attempts needed to achieve the correct cue. In this case, the inconsistencies with time are eliminated, and the observations more accurately describe how hard it is for the subject to generate the correct sEMG trigger.

6.2.3.3 Pattern Recognition

The Pattern Recognition experiment as is may be sufficient, however it would greatly benefit from using subjects who are already trained with pattern recognition. Although we give subjects up 20 minutes to practice, this is most likely not enough to remove possible differences due to learning. Subjects who are already familiar with the system should prove to be more consistent across evaluation sessions. As a result, differences in the evaluation sessions across the different electrode types can be more confidently attributed to the electrodes themselves and not variability in the subject’s own performance.

Due to a limitation in resources, we were not able to fabricate a set of eight active textile electrodes for the study and remote textile electrodes were used. While this is
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an option for showing preliminary data that suggests pattern recognition is possible with these electrodes, it is imperative that the future studies be conducted with the active textile electrode prototypes. This will undoubtedly be more representative of the actual textile electrode design intended for the final device.

6.2.4 Added Noise Analysis

While the interference analysis carried out in this thesis is somewhat effective in showing the ability for the active textile electrode prototypes to reject noise, a more quantitative measure should be used in the future. To quantify the degree of noise rejection, traditional SNR should be reported. This SNR is computed using the raw sEMG signal and is represented by the ratio of RMS voltage of the sEMG signal at full-scale (100% MVC) versus the RMS baseline signal voltage. Traditional SNR can be computed as follows [139]:

\[
SNR(dB) = 20 \log \left( \frac{FullScaleV_{rms}}{BaselineV_{rms}} \right)
\]  
(6.2)

Another parameter worth investigating is the CMRR of the entire system. While the amplifiers of choice will have their own CMRR specification, investigating and reporting the electrode+amplifier CMRR may also be of interest. This can be accomplished by investigating the system’s response to a common mode signal sent to two electrode contacts at the amplifier’s differential inputs [140,141].
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6.2.5 Motion Artifact Study

One drawback of this thesis is that due to the limitations of our current active electrode prototypes, a reliable motion artifact study was not possible. The supposed motion artifact observed on the signal during movement of the subject or the subject’s arms can actually be attributed to cable interference resulting from the fact the the amplifier in the active prototype is not directly on the back of the electrode. As such, the artifact is not representative of the actual motion artifact that we are concerned with. With the next iteration of active textile electrode prototypes, however, a motion artifact study will be imperative.

6.2.6 Long-Term Studies

Another drawback is the absence of any long-term testing. Once the textile electrodes can be integrated and embedded into a silicone liner, the liner prototype must be worn by an active amputee and long-term studies must be carried out to test the performance of the textile-electrode interface over time. Key points of interest will be: sweat accumulation in the silicone liner and its effect on signal quality, the liner’s ability to maintain the electrode-skin contact in an active setting, subjects’ range of motion, the durability of the textile electrode material over time and in the presence of sweat, the durability of the integrated liner as a whole during continuous donning and doffing, ease of electrode location alignment between donnings, and/or reliability
CHAPTER 6. CONCLUSION AND FUTURE WORK

of the auto-calibration feature for automatic channel selection.
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Vita

Deena Jamal received a Bachelor of Science degree in Biomedical Engineering from the University of California, Irvine in 2010. During her undergraduate career, Deena received monetary awards for her research on the mechano-transduction properties of endothelial cells, and also served as an active board member/President of the Lebanese Social Club. Following graduation, Deena worked at a start-up located in Orange County, CA as a design engineer for one year and assisted with the development of a novel root canal device. She then joined the Master’s of Science in Engineering program at Johns Hopkins University in 2011. Her research throughout her JHU Master’s program was in collaboration with Infinite Biomedical Technologies and focused on the development and research of products that aim to make prostheses more accessible and reliable for amputees. In September of 2013, she presented her work on an alternative textile-electrode interface for myoelectric prostheses at the American Orthotic and Prosthetic Association World Congress.
VITA

After graduation, Deena plans to work as a consultant at Infinite Biomedical Technologies, and eventually relocate back to Southern California to pursue a career in the biomedical device industry.