DESIGN SPECIFICATIONS FOR A COMMERCIAL VIABLE 
ELECTROOCULOGRAPHY ACQUISITION DEVICE FOR THE DISABLED

by

Joshua E. Keys

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Supervisors: Philip A. Parker, PhD, Electrical and Computer Engineering
Yves G. Losier, PhD, Electrical and Computer Engineering

Examining Board: Kevin B. Englehart, PhD, Electrical and Computer Engineering
Bernard S. Hudgins, PhD, Electrical and Computer Engineering
Kenneth B. Kent, PhD, Computer Science

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Dean of Graduate Studies

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ABSTRACT

In this thesis, several key aspects in the design of a commercially viable electrooculography (EOG) acquisition device are discussed. The context for the needs of such a device, along with associated background in the EOG signal and previous work are presented.

The topic of electrooculography electrodes is introduced with a thorough investigation into seven different types, all chosen for their relatively small size and practicality. This investigation includes the impedance settling time, sweep response, Warburg model parameters and motion artifact data for each electrode, providing an objective overview.

EOG signal variability from subject to subject is explored with strong correlation found. The majority of the face area is assessed for levels of electrooculography signals present with all areas showing potentially useful levels. Finally, a novel pattern recognition algorithm is used to assess the accuracy of EOG signal classification while exploring the effect of the number of electrodes used and their positioning.
DEDICATION

This thesis is dedicated to the many people suffering from neurological diseases in this world. I hope my work can lead to the development of devices to improve their lives.
ACKNOWLEDGEMENTS

The author would like to thank his co-supervisors Dr. Yves Losier and Dr. Phil Parker, along with Mr. Adam Wilson, for their knowledge, guidance and assistance. The author would also like to thank the Stan Cassidy Center for Rehabilitation for both their financial support and accommodating work schedule.
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<td>ABS</td>
<td>Acrylonitrile butadiene styrene</td>
</tr>
<tr>
<td>ADC</td>
<td>analog-to-digital converter</td>
</tr>
<tr>
<td>AF</td>
<td>amplitude factor</td>
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<tr>
<td>ALS</td>
<td>amyotrophic lateral sclerosis</td>
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<tr>
<td>AP</td>
<td>action potential</td>
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<tr>
<td>AT</td>
<td>assistive technology</td>
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<td>ATS</td>
<td>assistive technology services</td>
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<tr>
<td>Ag-AgCl</td>
<td>silver-silver chloride</td>
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<tr>
<td>CE</td>
<td>counter electrode</td>
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<tr>
<td>CM</td>
<td>common-mode</td>
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<tr>
<td>CMRR</td>
<td>common-mode rejection ratio</td>
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<tr>
<td>CNS</td>
<td>central nervous system</td>
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<td>CRP</td>
<td>corneal-retinal potential</td>
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<tr>
<td>DC</td>
<td>direct current</td>
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<tr>
<td>DMD</td>
<td>Duchenne muscular dystrophy</td>
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<td>EEG</td>
<td>electroencephalography</td>
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<td>EMG</td>
<td>electromyography</td>
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<td>EOG</td>
<td>electrooculography</td>
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<td>ERG</td>
<td>electroretinography</td>
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<tr>
<td>HF</td>
<td>high frequency</td>
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<td>IA</td>
<td>instrumentation amplifier</td>
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IR  infrared
ISCEV  International Standard for Clinical Electrophysiology of Vision
LED  light emitting diode
LiS  locked-in syndrome
LPF  low pass filter
LTI  Liberating Technologies Inc.
MD  muscular dystrophy
MS  multiple sclerosis
MSE  mean squared error
NIA  Neural Impulse Actuator
OT  occupational therapist
PC  personal computer
PIC  peripheral interface controller
REDOX  reduction and oxidation
RE  reference electrode
RMS  root mean square
RPE  retinal pigment epithelium
SCI  spinal cord injury
SE  sense electrode
SENIAM  Surface EMG for Non-Invasive Assessment of Muscles
SF  spread factor
SLP  speech language pathologist
SNR  signal to noise ratio
<table>
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<th>SS</th>
<th>stainless steel</th>
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<tr>
<td>TENS</td>
<td>transcutaneous electrical nerve stimulation</td>
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<tr>
<td>TF</td>
<td>threshold factor</td>
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<tr>
<td>TMSI</td>
<td>Twente Medical Systems International</td>
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<tr>
<td>VP</td>
<td>vector projection</td>
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<tr>
<td>WE</td>
<td>working electrode</td>
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1 Introduction

1.1 Neurological Disorders

A neurological disorder is any disorder that affects the nervous system of the body [1]. The effects of these disorders can vary greatly in symptom and progression, with severe cases leading to a state of locked-in syndrome (LiS), more precisely, a cognitively aware patient with complete inability to move or speak while retaining vertical eye movement or blinking [2] [3]. As one approaches a LiS state, the ability to carry out even the simplest of tasks becomes increasingly difficult to the point of becoming impossible, leaving the patient with very limited, if any, solutions for communication or the control of their environment. Of particular interest to this thesis, are those disorders most often responsible for leaving a patient in a near, or actual, LiS state, where little more than eye movements are possible.

1.1.1 Spinal Cord Injury

Spinal cord injury (SCI) results in nearly 12,000 paralyzed Americans each year with close to a quarter million cases already existing. The highest rate of incident occurs in patients between the ages of 20 to 40 with 70% of all SCI patients below the age of 40 [4]. Simply put, a large percentage of SCI patients will be severely debilitated for extended periods of time requiring lifelong care.

1.1.2 Muscular Dystrophy

Muscular Dystrophy (MD) is a broader term used to describe a group of hereditary muscle disorders that can occur with varying levels of severity at any age [4]. Once
thought to be a result of a lack of proper nutrition (*dys* – lack and *trophy* – nutrition), MD has since been attributed to a genetic disorder that affects the body’s ability to create the proper proteins for muscle growth and maintenance [5]. Duchenne’s MD (DMD), caused by a deficiency of the dystrophin protein in the body [6], is the most common and usually most severe subset of MD. DMD affects approximately 3 out of 100,000 of the general population and will most likely leave the patient confined to a wheelchair and severely disabled for the remainder of their life [4].

1.1.3 Multiple Sclerosis

Multiple Sclerosis (MS) is a disease of the central nervous system (CNS) affecting nerve conduction. Myelin, produced by oligodendrocytes cells in the CNS, covers the axons of motor neurons in discrete segments providing significant increases in conduction velocities (saltatory conduction), necessary for motor control. In a process known as demyelination, the oligodendrocyte cells and associated myelin are damaged to the point of disrepair, thereby affecting, or even halting, nerve conduction [7] [4]. MS is the leading cause of disability in young adults [8] with only a modest negative effect on longevity of a patient’s life, translating to a possibility of years with severe disablements [4].

1.1.4 Amyotrophic Lateral Sclerosis

Amyotrophic Lateral Sclerosis (ALS), also known as Lou Gehrig’s disease, is a neurodegenerative disorder that primarily affects both the upper and lower motor neurons [4]. The disease name is of Greek origin where *a* means without, myo refers to muscle and *trophic* to nourishment, or simply put, the muscles lack nourishment [9]. This
malnourishment leads to the *sclerosis*, or hardening, of the motor neurons located in the *lateral* area of the corticospinal tract responsible for movement [10]. The disease will often spare movements of the eyes until the very end [11], leaving the patient in a near or complete LiS. The disease is progressive with the majority of patients eventually succumbing to respiratory failure [4], although, the use of bilevel positive airway pressure (BiPAP) devices have been shown to slow progression and improve symptoms [12].

![Figure 1-1 Common BiPAP masks](image)

As shown in Figure 1-1 [13], the BiPAP devices have varying types of masks used for operation that come in contact with nearly all parts of the face between them.

### 1.2 Assistive Technology

Many patients of neurological disorders, particularly those suffering from LiS, may lack any form of unassisted communication or control of their environment. It is for this reason that Assistive Technology (AT) has been developed. AT is a term used to describe any device or service used in assisting the disabled.
1.2.1 Assistive Technology Services

An assistive technology service (ATS) describes a process of aiding any disabled patient in the selection, acquisition and use of an assistive technology device for their personal use [14]. Typically, at neurological rehabilitation centers, the ATS department consists of occupational therapists (OT), speech language pathologists (SLP) and rehabilitation engineers or technicians specialized in their respective disciplines for providing technological solutions to their patients. These professionals work together with the patients to address all needs concerning communication, transportation, computer access and many other forms of technology. Each profession provides their respective disciplinary perspectives on the long-term goals of the patient's access and needs in terms of function, safety, repeatability, ease of use and other factors.

ATS members will use their skillsets to select or create devices based on many factors such as movements that may be remaining, rate of disease progression, environmental considerations such as indoor or outdoor usage etc. Once identified, the ATS members will work to integrate the device into the routine of the patient such that a local caregiver, family member or therapist can provide any additional support needed from that point forward.

1.2.2 Assistive Technology Devices

"An assistive technology device is any item, piece of equipment, or product system, whether acquired commercially, modified, or customized, that is used to increase, maintain, or improve functional capabilities of individuals with disabilities" [14].
As described by the Assistive Technology Act of 2004 above, these devices can be separated into 3 broad categories: commercially available devices, modification of existing devices and fully customized devices.

In their simplest form, a device such as a letterboard provides, as might be expected, letters on a board to enable a caregiver to track the eyes of a patient as they spell out words and sentences. Such devices are referred to as ‘low-tech’ and play a very important role in AT due to their low cost, minimal setup time, operation in non-ideal environments and as guaranteed back-up systems when all else fails.

Modification of an existing device provides the patient with a more simplistic form of access or use. For example, the process might involve the left click button on a mouse, altered to enable an external, more accessible, switch or button as in Figure 1-2.

![Figure 1-2 Switch Adapting Left Click Button on Mouse](image)

Typically, the connections from the mouse to the external switch are done with a standard 3.5mm headphone jack. This method of connection provides the therapist with a wide selection of commercial or custom switches and buttons, all with a 3.5mm
headphone plug, tailored to the patient’s current needs without preventing alternate options in the future as the patient’s disease progresses or other factors arise.

As the severity of the disorder increases, the availability of assistive devices decreases. For this reason, a patient approaching a LiS state will begin to pose a challenge for an ATS team to provide a reliable solution. Although some ‘low-tech’ solutions can be applied, they often do not provide the level of independence requested by the patient. In the next section, a few of the more commonly prescribed high-tech assistive devices, along with their applications, advantages and disadvantages, will be examined.

1.2.2.1 Head Tracking

If head mobility remains, several options exist for capturing these movements and translating them to cursor movements on a computer. One such commonly prescribed device, the SmartNav [15], requires less than an inch of head movement for the cursor to travel the entire span of the screen. The device uses an infrared (IR) camera and IR light emitting diodes (LED) to detect reflections from a small silver sticker placed on the patient’s hat, forehead, glasses, or any other mobile location. These reflections are then translated to mouse movements with clicking achieved by either “dwelling” on the area of interest for a preset adjustable time period or use of an external switch connected to the SmartNav.

Although many patients demonstrate success with use of the device, there are potential drawbacks and nuances that can sometimes plague the experience. The most obvious requirement for device operation is controlled movements of the head. Many factors may affect the reliability of these movements, from fatigue limiting the initial range of motion
used for set-up, to the changes in angle a patient may experience throughout the day (upright in chair, reclined, bed, etc.), where the forces of gravity may affect different muscles of the neck and head. Because the device relies on tracking IR light reflections, any other sources of IR, such as the sun, will introduce errors on the system rendering it non-functional outdoors or near any natural sources of light (window) indoors.

Furthermore, users who wear glasses often times will experience these same issues due to unwanted reflections off their lenses. Last, and most obvious, the device requires the use of a computer which may be of no interest to the patient due to cost or complexity.

1.2.2.2 Eye Tracking with Camera

When head movement does not exist, tracking of the eye may provide another viable option for mouse control. The device can be as simple as software for an existing webcam to track eye movements, to more expensive units providing a tablet with integrated IR LED’s, camera and proprietary software such as those designed by Alea or Tobii [16] [17]. As technology improves, these devices are demonstrating greater ability to accurately track the eye with fewer drawbacks. However, as with the head tracking devices, the eye gaze systems rely on a camera to follow the movements of the eye with IR light for subject illumination. This camera tracking approach can often lead to reflection issues with contact lenses, glasses, the sun and other bright sources of light. There is typically also a great deal of set up time required to mount such a system while accounting for any of the anomalies as previously mentioned, all of which need be repeated if the patient requires a different location or angle for use. The systems have a very large associated cost with some requiring proprietary hardware that does not utilize
the computer a patient may already have. Finally, as with head tracking devices, the use of a computer or tablet is required introducing additional costs and unwanted complexity to the patient.

1.2.2.3 Motion Sensitive Switches

If head movements are not possible, there are a variety of motion sensitive switches to allow very minute movements to be detected. One such device, a SCATIR switch [18], uses the reflection of infrared light from its optical emitter to detect subtle movements. If detected, the 3.5mm jack output will close, emulating a switch for adapted device control. Placement is typically done near the end of fingertips if the patient has any movement or to detect eyelid or eyebrow movements. The device can be difficult, especially for eye blinks, to set up and will respond to all blinks. The EyeBlink Switch [19], allows for regular blinks to be ignored, but can be difficult to mount and blocks part of the peripheral vision of the patient.

1.2.2.4 Electromyography (EMG) Switch

Even though a patient may not have the strength to move, this does not necessarily mean that the EMG signals from the muscles are too small to measure. The EMG Switch [20] allows the placement of an electrode at any possible viable location for EMG detection. The device’s threshold is then set to activate with the patient’s given muscle contraction strength. Often times this can prove a viable solution, but in low functioning patients, a reliable and repeatable site for EMG detection cannot be found. Furthermore, the system is based on a single input, single output scheme such that if multiple outputs are required, multiple unique input sites must be present.
1.2.2.5 Electrooculography (EOG)

As discussed in 1.1.4, ALS will typically spare the movements of the eyes until the very end. Therefore, it is very useful to capture the movements of the eye for AT in such low functioning patients. This measuring technique, known as electrooculography (EOG), exists in AT for the most part in research, prototypes and papers with the exception of very few commercial devices.

One such commercial device, the Brainfingers by Brain Actuated Technologies [21], is made up of 3 electrodes attached to a headband, tethered to a computer, where electroencephalography (EEG), electromyography (EMG) and EOG signals are all used to control various mouse and keyboard functions. The eye movements are restricted to horizontal detection only and the system is sold for just under $2000, not including the computer. The device can only be interfaced to a computer, again limiting its uses outside of mouse and keyboard control.

Along the same lines, the Neural Impulse Actuator (NIA) developed by OCZ in conjunction with Brain Actuated Technologies offered most of the same functionality as the Brainfingers device at a much more reasonable rate of $300. The product however has since been discontinued.

Finally the EOG switch developed by Dr. Tohru Yagi, and commercialized by SeaStar Corporation in Tokyo in 2002 [22], offers a simple switch toggle output controlled with lateral eye movements. The device is limited to a single output and sensitive to placement and patient. However, at time of publication, the author has been unable to find a distributor or receive a response from the company for more information.
1.3 Objectives

With the devastating severity of many neurological disorders, it is evident the important role AT plays in a patient’s life. Although many devices exist, some previously discussed, the potential for EOG has not yet been fully realized. With the multitude of papers on EOG research and design, no single device to date exists that can truly meet the needs of a patient in terms of ease of use, low cost and aesthetic appeal. Most attempts fall short of bridging the gap between experimental and practical due to a lack of a full understanding of the patients’ needs and the EOG signal itself. There is also a lack of literature and/or understanding with respect to alternative locations on the face that may be used for EOG processing.

This thesis aims to objectively, through experiment and analysis, provide information for the understanding and design of a commercially viable EOG acquisition system. The system will meet the needs of patients at, or near, a LiS state by accommodating electrode placement sensitivity and patient specific eye movements while providing multiple switch outputs for adapted device control. The thesis can be separated into the following three objectives:

1) A thorough examination into the physiological origins of EOG to provide a full understanding of the signal and its associated difficulties in acquisition.

2) A scientific comparison of several types of small and unobtrusive dry and wet electrodes to provide a variety of options for electrode choice.

3) An investigation into EOG signal strength mapping and alternative locations for EOG acquisition along with the use of a pattern recognition algorithm for verification of the locations feasibility for use.
2 Electrooculography

2.1 Introduction

With the design of any product, it is important to fully understand the nature of the problem one is dealing with. In the case of EOG, it is necessary to first explore the origins of the signal before designing a device that will capture and use these signals for processing. The following chapter provides an overview of the anatomy of the eye, its operation and, finally, how it applies to EOG with previous research discussed.

2.2 Anatomy of the Eye

The ostensibly complex process of vision begins at the anterior portion of the eye, the cornea (C) (Figure 2-1), where light from our surroundings first enters. From here, the light passes through the fluid filled anterior chamber (A) where it is then focused by the lens (L) to accommodate for varying focal distances in our field of view. Finally, the light travels through the gelatinous vitreous body (V) and onto an inner surface of the eye, known as the retina (R), where visual processing begins.

The retina, a thin filmy piece of brain tissue, is made up of millions of densely packed nerve cells arranged in 10 distinct layers with the first 9 photoreceptive and the remaining one, the retinal pigment epithelium, being nonreceptive [26].
Due to the path that light must take to reach the photosensitive cells at the back of the retina, the layers are often thought of as inverted. This layout is however a result of the vital support mechanisms required of the photoreceptors of the eye which are provided through direct contact with the retinal pigment epithelium (RPE) [27]. The RPE provides metabolic support to sustain the photoreceptors of the eye and controls the local environment [28].

2.3 Operation of the eye

As light enters the retina, the first level of excitation begins at the second innermost layer, the layer of rods and cones as shown in Figure 2-2. Here, the varying photosensitive opsin pigments serve to depolarize these photoreceptors in an inversely proportional relationship to the illumination intensity. In the absence of light, sodium
gated channels within the photoreceptors open to invoke a depolarizing ‘dark current’ (Figure 2-3 [29]) resulting in a steady release of the neurotransmitter glutamate.

As the light increases, the gated channels restrict the flow of sodium and eliminate the dark current thereby hyperpolarizing the photoreceptor and stopping the transmission of
glutamate. From there, the synaptic mode of signal transmission continues on through inner nuclear layer of bipolar cells and finally on to the layer of ganglion cells with horizontal and amacrine cells intertwined along the way to aid in the provision of the first stages of our visual processing beyond the scope of this thesis. Once transmission has reached the ganglion cells, action potentials (AP) carry the signal the remainder of the way to the brain through the optic nerve for final processing.

Through all of these intracellular processes, various potentials are developed that form the basis of many clinical diagnostic techniques for the eye. The constant sourcing of current due to the ionic movements within the retina must all eventually return, thereby establishing potential gradients along this loop (Figure 2-4).

![Figure 2-4 Currents of the Eye](image)

Despite the fact that most current source and sink loops will remain within the retina, some will find their way beyond the eye enabling the use of non-invasive measurement techniques. With the majority of neurons and glial cells within the retina aligned radially (with the exception of the horizontal and amacrine cells) [27], summation of these potentials serves to magnify the resultant signal for easier acquisition.
2.4 Electrooculogram

2.4.1 Origin and Clinical Uses

Electrooculography (EOG) is a technique employed to measure a physiologic electrical potential difference between the anterior pole of the eye, the cornea, and posterior pole, the retina, with the cornea by convention being positive [31]. This potential, known as the resting potential or corneal-retinal potential (CRP), was first discovered in 1848 by Emil du Bois Reymond [32]. The potential originates in the retinal pigment epithelium (RPE) layer of the eye, due to a sodium ionic imbalance across its nearly impermeable tight junction membrane [33]. When ambient light changes occur, a hyperpolarizing of the apical (side closest to photoreceptors) membrane of the RPE propagates over a 1 minute period to the basal (opposite of apical) side, altering the resting potential and is referred to as ‘fast oscillations’ due to its relatively quick response [34] [27]. As these changes are occurring, a release of an unknown ‘light messenger’ substance is believed to bind to a receptor on the apical membrane of the RPE [35]. This binding is also believed to trigger the release of a second unknown messenger within the RPE leading to a depolarization of the basal membrane, known as the ‘light-peak’, typically 5-10 minutes after onset. If the light source persists, the mechanisms whereby the fast oscillation and light peak occur will continue in a cyclical fashion with decreasing amplitude known as ‘slow oscillations’. The combination of the standing and light sensitive potentials propagate through the tissues of the face to the electrodes used for the electrooculogram.
In Figure 2-5 [36] [37], the EOG signal follows the movements of the eyes; the greater the rotation of the eye towards an electrode, the greater the resultant signal. A typical signal level is in the 10μV to 5mV range [27], and is highly dependent on electrode proximity to the eye.

The routine use of the EOG as a diagnostic tool has resulted in the publication of several standards [38] [39] that ensure uniformity between clinical centres. These standards set out guidelines for not only the way tests are conducted but also on such factors as electrode placement and amplifier characteristics. These guidelines, unfortunately, do not cover the use of alternative electrode material types or locations relevant to use in the AT field. In a typical clinical EOG setup, a patient is placed in an ordinary lit room for upwards to an hour before testing begins. Electrodes are then configured as in Figure 2-6 [38], the head is immobilized and the patient is asked to fixate on red lights located 15° left and right of center, alternating once per second, for 10 seconds out of every minute.
Recording begins when the lights are turned off, leaving the patient in a state of complete darkness, save for the alternating fixation lights, for a period of 15 minutes. During this time period, the EOG amplitude will slowly decrease until it then begins to rise again. The amplitude level at the point in time where the change from decreasing to increasing occurred is known as the ‘dark trough’. Following this, illumination is restored for at least 15 minutes, or until a ‘light peak’, as earlier discussed, is noted on the EOG recording. It is the ratio of the amplitude of the light peak compared to that of the dark trough that results in the clinical diagnosing tool of the EOG known as the Arden ratio (Figure 2-7 [37]).

![Figure 2-7 Arden Ratio](image)

Although no established normal reference data is currently available, the International Standard for Clinical Electrophysiology of Vision (ISCEV) standard for clinical EOG
reports an Arden ratio <1.5 as abnormally low and >2.0 as normal [38]. However, it is recommended that each clinic develop their own normative data to ensure proper diagnosis.

As the origins of the EOG lie in the RPE, the Arden ratio provides clinicians with an insight into the integrity of this layer. This in turn has led to the EOG being used in the diagnosis of such eye diseases as Best dystrophy, pattern dystrophies, Stargardt’s disease and vitamin A deficiency [40]. In addition, recent evidence suggests that the EOG can also be used to identify asymptomatic “carriers” of Best disease in skipped generations [41]. Another well-established application of the EOG is in the study of the movements of the eye. This has been used extensively in the study of sleeping patterns [42] and in the diagnosis of abnormal nystagmus [43].

2.4.2 Development & Acquisition Problems

As mentioned briefly earlier, the use of EOG in AT is very limited and lacking in developed and practical solutions. There have been many novel approaches to solve some of the issues surrounding EOG acquisition such as the slow drifting of the EOG signal over time and artifacts caused by blinking [44], but none that fully realize the complex demands of an ALS patient. Venkataramanan et al. [45] developed an EOG device to aid bed restricted hospital patients but neglected to account for skin irritation caused by the use of silver-silver chloride (Ag-AgCl) electrodes placed at the same location on the skin for extended periods of time. In 2009, Uchtiomi & Hori [46] developed a pointing device for the disabled using EOG but the system required a computer for operation thereby eliminating portability. Zheng et al. (2009) utilized a wireless approach to EOG
acquisition however it required electrodes to be placed in a typical EOG electrode placement pattern causing problems for any patients using BiPAP.

To date, there have been few alternate solutions for electrode placement. Kirbis & Kramberger (2009) have partially developed an EOG device that incorporates many practical and intelligent designs aspects however it will require the use of goggles for electrode placement. Manabe & Fukumoto (2006) have designed what appears to be the most promising alternative to traditional electrode placement and would allow placement of a BiPAP device. Their design incorporates electrodes into headphones that could then be worn by the patient; however, this would affect the hearing of the patient and therefore is not desirable.

The author has also spent a great deal of time developing simple EOG devices for patient use. The devices were sensitive to equal and opposite electrode placement at the outer canthus of both eyes (the temple area), and looked for specific hardcoded eye movements that were not indicative of all patients. Furthermore, the adjustment of the sensitivity in both eye movement and time allotted for the movement, proved difficult and intimidating for some caregivers.

In short, there has been no single device that meets the needs of many ALS patients or other acute care patients where aesthetics (wires on and across the face) or existing facial obstructions may be a factor. Any of the aforementioned devices only exist in research and, as of the date of this thesis, have not been made available to the general clinical community or realized commercially.

As the first step of acquisition is at the electrode, it is prudent to begin by investigating their various properties. With the multitude of choices, it was necessary to present a
subset of electrodes that best represented possibilities for future EOG designs. The subsequent chapter introduces the basic theory behind the function of an electrode followed by experiments aimed at providing objective measures for electrode choice.
3 Electrodes

3.1 Introduction

As the eyes and their associated movements generate potential gradients throughout the face, it necessary to acquire, process, and interpret these signals for AT purposes.

The first step, the acquisition, begins with the application of electrodes to the body. By nature, the flow of current throughout the eye (and body) is a result of ionic exchanges and movements in response to both environmental stimuli (light, heat, touch, etc.) and internal mechanisms (metabolic changes, action potentials, synapses, etc.). In contrast, the flow of current from electrodes to lead wires is carried by electrons, necessitating an ion to electron transduction to occur. Fortunately, electrodes provide this transduction function inherently allowing the ion-induced signals in our body to be passed through the electrode and on to the equipment for acquisition. However, this transduction is not without its consequences and greatly complicates electrode application by placing constraints on their operation [47]. In the following sections the issues of electrode half-cell potential, impedance, and motion artifact are considered.

3.1.1 Half-Cell Potential

The transduction of ions to electrons and electrons to ions by electrodes is accomplished via oxidation and reduction, or REDOX, reactions. That is, when an atom is reduced, it gains an electron, thereby reducing its overall charge. When an atom oxidizes, it gives up an electron, thereby increasing its overall charge.

When a piece of metal, or electrode, is introduced into an ion rich solution, or electrolyte, the metal will have a tendency to either oxidize or reduce. If it is assumed that
oxidization predominately occurs, that is, leaves behind an electron and enters the solution as a cation, this process will result in a net negative charge distribution on the electrode. At the same time, some of these cations will reduce, or be reabsorbed onto the electrode by taking back an electron, until the rate of oxidation and reduction reaches equilibrium. The net charge and electric potential at which this equilibrium occurs is known as the half-cell potential and is dependent on electrode type, ion concentration, temperature and other second-order factors [47].

3.1.2 Noise & Overpotentials

Having established a standard for measuring half-cell potentials and an understanding of its origins, one would expect that if two identical metals and ionic solutions were used as electrodes, such as those used in a differential recording, a net zero electric potential difference between the two (signal excluded) would occur. Aronson & Geddes [48] experimented with this concept and found differences in potentials of some identical electrode pairs to be on the order of several hundred millivolts, due to metal/solution differences.

Further complicating the matter, as current begins to flow in our electrodes, such as the currents developed in the acquisition of EOG signals, additional offsets occur, known as polarization or overpotentials ($V_p$), that deviate from standard half-cell potential values. All real world electrode overpotentials are made up of three distinct components [49]:

$$V_p = V_r + V_c + V_a$$
Ohmic Overpotential ($V_r$):

As current passes from the body through to the first stage of amplification, it will encounter impedances resulting in small voltage drops. These voltage drops are known as ohmic overpotentials and contribute to the overall overpotential, $V_p$.

Concentration Overpotential ($V_c$):

As discussed in a previous section, when a metal is introduced into an electrolytic solution, a distribution of ions via REDOX reactions occurs at the electrode-electrolyte interface until an equilibrium is achieved, resulting in a half-cell potential. When a current is developed, this equilibrium rate of REDOX reactions is altered such that the rate of oxidation to reduction is no longer the same. This change affects the half-cell potential and is known as the concentration overpotential, $V_c$.

Activation Overpotential ($V_a$)

In order for the atoms of metal to be oxidized into metal ions and enter the electrolyte solution, an energy barrier, or activation barrier, must be overcome [47]. The reverse also holds true for the reduction of ions to be deposited back onto the electrode. With the application of current, the activation energy barriers for either the oxidation or reduction reactions are significantly modified dependent on the direction of current flow [49]. This difference is manifested as an activation overpotential, $V_a$, and contributes to the overall overpotential components affecting the half-cell potential.

3.1.3 Polarizable and Nonpolarizable Electrodes

"Theoretically, two types of electrodes are possible: those that are perfectly polarizable and those that are perfectly nonpolarizable." [47]. Perfectly polarizable electrodes are
those in which no electron or ion charge transfer occurs across the electrode interface [50]. Instead, displacement current as in a capacitor is responsible for the current flow negating the need for ion transfer between the skin and electrolyte. Because of this, these electrodes are considered safe for long term use. However, movement of the electrode with respect to the electrolyte solution will alter the charge distribution, resulting in motion artifact.

In contrast to this, in a perfectly nonpolarizable electrode: ions from the electrolytic solution pass electrons to and from the electrode with no build up or accumulation of charge, resulting in no overpotentials [51]. Therefore, movement of the electrode with respect to the electrolyte will have no effect on the measured potentials. However, due to the ion exchanges which can occur into and out of the skin, nonpolarizable electrodes are not recommended for long term use. Most inert metals exhibit the characteristics of polarizable electrodes whereas the common silver/silver-chloride (Ag/AgCl) electrode more closely models the nonpolarizable class.

Although it is not possible to construct either category of these ideal electrodes, all real world electrodes will fall somewhere in between and exhibiting some of the characteristics as shown in Table 3-1.

<table>
<thead>
<tr>
<th></th>
<th>Ideal Polarizable</th>
<th>Ideal Nonpolarizable</th>
</tr>
</thead>
<tbody>
<tr>
<td>Overpotentials</td>
<td>Ohmic &amp; Concentration</td>
<td>None</td>
</tr>
<tr>
<td>DC</td>
<td>No</td>
<td>Yes</td>
</tr>
<tr>
<td>Long term use</td>
<td>Yes</td>
<td>No</td>
</tr>
<tr>
<td>Motion Artifact</td>
<td>Yes</td>
<td>No</td>
</tr>
</tbody>
</table>

Table 3-1 Ideal Electrode Type Characteristics
3.1.4 Electrode Circuit Model

Given the complexity involved at the electrode-electrolyte interface, it is of no surprise that electrically modeling this process proves just as difficult. Despite this fact, the characterization of electrode/skin interface impedances has been the subject of review for many years leading to multiple interpretations. Of specific interest to this thesis is that of the modified Warburg Model developed by Warburg in the early 1900's [52] [53] and shown in Figure 3-1.

Figure 3-1 Modified Warburg Electrode Model

Figure 3-1 depicts the half-cell potential, $E_{bc}$, modeled as a battery source. $C$ represents the capacitance across the double layer of charge at the electrode-electrolyte interface with a parallel resistance, $R_p$, to signify the associated leakage. Finally there is $R_s$, a measure of the resistance in the electrolyte, in series with the former impedances. From this model it is apparent that at low frequencies, $C$, will act as an approximate open circuit resulting in an overall impedance equal to the sum of $R_s$ and $R_p$ (the Faradic Region, Figure 3-2 a.). At high frequencies, $C$ will act as an approximate short circuit shunting $R_p$ such that the overall impedance will be roughly equal to $R_s$ (the Subject Region, Figure 3-2 b.). The remaining mid-range frequencies are represented by the full modified Warburg model (the Warburg Region).
The combination impedance of all three regions is shown in Figure 3-3.

Through straightforward circuit analysis, the impedance of the Warburg Model can be shown as in (1).

\[ Z(s) = R_s + \frac{R_p}{R_p s C + 1} \] (1)

From (1) the associated circuit pole and zero frequencies, \( f_z \) and \( f_p \), are derived as in (2)

\[ f_z = \frac{1}{2\pi C R_p || R_s} \quad f_p = \frac{1}{2\pi C R_p} \] (2)

By measuring the impedance in the Subject-Impedance Region it is possible to estimate the value of \( R_s \). Knowing this, the impedance value in the Faradic Region can be
used to derive $R_p$. Finally, using these values, and the pole and zero equations of (2), the model parameter capacitance $C$ can be calculated. These models provide a significant oversimplification of the electrode-electrolyte interface, not only in the omission of certain components such as the diffuse layer and its associated modeling [54], but also in negating the frequency dependency of each of the models components and subject variation. Therefore, it is often necessary to adjust the value of $C$ in an iterative fashion until the mean squared error (MSE) of the calculated value is minimized with respect to the experimental value.

3.1.5 Motion Artifact

As discussed in 3.1.4, the electrode-skin interface can be modeled and subdivided into multiple parameters. One such parameter, a capacitor, is representative of two layers of charge, known as a double Helmholtz layer [55], and therefore directly proportional to the area of the skin and electrode/electrolyte in contact, and the distance between them (3).

$$C = \varepsilon_r \varepsilon_0 \frac{A}{d} \quad (3)$$

$$I(t) = C \frac{dV(t)}{dt} \quad (4)$$

It is then evident that if the area or distance between the skin and electrode/electrolyte is affected, the capacitance will also be affected. Since the voltage across a capacitor cannot change instantaneously (4), if the capacitance is altered, the capacitor current must also change. This sudden change in capacitance will draw more or less current through the impedance of the tissues and electrode/electrolyte interfaces, thereby altering the
measured voltage in the associated recording system. These unwanted spikes are known as motion artifact.

A second possible cause of motion artifact, known as 'skin stretch reflex', is a result of the alterations of the thickness of the outer layers of the epidermis due to stretching or movement [56]. Just like the alterations to charge layers of the skin-electrode/electrolyte interface, the charge layers in the epidermis are affected by these same movements and result in similar motion artifacts.

3.2 Electrode Impedance Analysis

The selection of an electrode type in any application, let alone EOG, must take into account factors such as shape, size, material, and usage duration. All of these parameters ultimately affect the overall impedance of the electrode, as shown in Table 3-2 [57], thereby altering its associated model parameters.

<table>
<thead>
<tr>
<th>Property</th>
<th>Change in Property</th>
<th>Changes in Electrode Impedance</th>
</tr>
</thead>
<tbody>
<tr>
<td>Surface area</td>
<td>↑</td>
<td>↓</td>
</tr>
<tr>
<td>Polarization</td>
<td>↑</td>
<td>↑At low frequencies</td>
</tr>
<tr>
<td>Surface Roughness</td>
<td>↑</td>
<td>↓</td>
</tr>
<tr>
<td>Radius of Curvature</td>
<td>↑</td>
<td>↓</td>
</tr>
<tr>
<td>Surface Contamination</td>
<td>↑</td>
<td>↑</td>
</tr>
</tbody>
</table>

Table 3-2 Electrode Impedance Properties

In order to characterize the varying types of electrode materials possible for use in an EOG AT device, the impedance of 7 electrode types shown in Table 3-3 were tested and analyzed to aid in an objective selection process. These electrodes were chosen based on their small size and availability, making them potential candidates for EOG applications.
As the variation due to the electrode/electrolyte/skin interface will be much larger than that of the variations between electrodes of the same material, the same electrodes of a given material were used from trial to trial.

<table>
<thead>
<tr>
<th>Material</th>
<th>Note</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gold (Au) [58]</td>
<td>Ear-Clip</td>
</tr>
<tr>
<td>Silver (Ag) [58]</td>
<td>Ear-Clip</td>
</tr>
<tr>
<td>Ag-Ag/Cl (Wet) [59]</td>
<td>TMSI REFA</td>
</tr>
<tr>
<td>Ag-Ag/Cl (Dry) [60]</td>
<td>Ear-Clip</td>
</tr>
<tr>
<td>SS Dome [61]</td>
<td>LTI for Prosthetics</td>
</tr>
<tr>
<td>Conductive Rubber [62]</td>
<td>TENS stim ear electrodes</td>
</tr>
<tr>
<td>Zoflex [63]</td>
<td>CD45.1 with FL45 Adhesive</td>
</tr>
</tbody>
</table>

Table 3-3 Electrodes Investigated Electrode Settling Time

The first experiment conducted of electrode comparison investigated the time course of impedance and half-cell voltage following electrode application. As discussed in 3.1.2, a half cell potential is developed as an electrode is introduced to the skin. These reactions that occur are not instantaneous and will therefore vary over time as equilibrium is reached. Similarly, the impedance will change over time. Many factors will affect the settling time, including the material, shape, size, location, environment thereby necessitating empirical data to provide objective comparisons of the previously discussed electrode material types.

3.2.1.1 Experimental Procedure

The VersaSTAT 3 by Princeton Applied Research [64] was utilized to provide the data necessary for impedance settling time measurements. The device applies a small AC potential to the Counter Electrode (CE), establishing a current that travels to the Working Electrode (WE) where the current and phase shift is measured via the Sense Electrode.
connected in parallel (Figure 3-4). These measurements were taken with respect to a 3M Red Dot [65] reference electrode, placed on the underside of the forearm one hour before the experiment commenced.

![Figure 3-4 VersaSTAT 3 Electrode Setup](image)

<table>
<thead>
<tr>
<th>Electrode</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>Working Electrode (WE)</td>
<td>Electrode of interest whose impedance will be measured.</td>
</tr>
<tr>
<td>Sense Electrode</td>
<td>Connected in parallel to the WE to measure the associated current amplitude and phase shift.</td>
</tr>
<tr>
<td>Reference Electrode</td>
<td>Provides a reference for measurements.</td>
</tr>
<tr>
<td>Counter Electrode (CE)</td>
<td>Driving electrode that controls power output of VersaSTAT</td>
</tr>
</tbody>
</table>

Table 3-4 VersaSTAT Electrode Descriptions

The VersaSTAT 3 does not provide a direct method for measuring the change in impedance of an electrode over a period of time. To overcome this, the VersaSTAT was set to sweep from 60Hz to 60.01Hz linearly with a total of 2161 points as shown in Table 3-5. With a small bandwidth and large number of points, it can be thought of as a time sweep of impedance at a given frequency over the range of time required to perform the test, which in this case, was 27 minutes and 9 seconds.
### Parameter Settings

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Settings</th>
</tr>
</thead>
<tbody>
<tr>
<td>Start Frequency</td>
<td>60 Hz</td>
</tr>
<tr>
<td>End Frequency</td>
<td>60.01 Hz</td>
</tr>
<tr>
<td>Point Spacing</td>
<td>Linear</td>
</tr>
<tr>
<td>Points Per Decade</td>
<td>2161</td>
</tr>
<tr>
<td>Time</td>
<td>~27 min 9s</td>
</tr>
<tr>
<td>Applied Voltage Amplitude</td>
<td>10mV RMS</td>
</tr>
</tbody>
</table>

Table 3-5 Electrode Impedance Settling Procedure Settings

Before each electrode was placed on the forearm, alcohol swabs were used to clean both the skin and the electrode of interest with ample time to ensure complete dryness. Following this, the VersaSTAT commenced recording immediately upon placing the electrodes on the forearm. The patient was asked to remain motionless for the 27 minute period to avoid any possible motion artifact effects.

#### 3.2.1.2 Settling Time Results and Discussion

For each of the 7 electrodes, 3 trials were performed on the forearm, with the cleaning and placement procedures as outlined above performed. The data was interpolated to represent time in seconds with each of the 3 trials per electrode shown in Figure 3-5.

The first observation is that of the low and stable impedance level of the TMSI electrode in comparison to the others, as expected for a relatively non-polarizable electrode. Second, the large variation in settling curves between even identical electrodes over the 3 trials is clearly evident. The effect of impedance magnitude is considered in 3.2.1.2.1. The effect of variation is further explored in 3.2.1.2.2.
Figure 3-5 Impedance Value vs Time (3 Trials Each)
Figure 3.6 Electrode Impedance and 95% Confidence Interval (CI) over Time
3.2.1.2.1 Electrode Impedance and Signal Attenuation

If only the magnitude of the impedance of one electrode, \( R_{e1} \), is examined, a simple voltage divider equation can be derived from Figure 3-7 as shown in (5). This equation can be used to assess the effect of electrode impedance on signal attenuation where \( R_{in} \) is the input impedance of one of the differential inputs.

\[
\text{% Attenuation} = \left(1 - \frac{R_{in}}{R_{in} + R_{e1}}\right) \times 100
\]  

(5)

If the Surface ElectroMyoGraphy for the Non-Invasive Assessment of Muscles (SENIAM) Project recommended 1000 M\( \Omega \) value of input impedance for use with dry electrodes is used, there is at most a 1% resultant signal attenuation given the highest value of initial electrode impedance tested. This demonstrates that, if a design uses the
recommended SENIAM input impedance value, the overall electrode impedance has very little effect on signal attenuation.

### 3.2.1.2.2 Electrode Impedance Mismatch

Due to the finite input impedance of an instrumentation amplifier (IA), not only must the impedance of each electrode be considered, but also the difference in impedance between ‘identical’ electrodes placed on the body. As seen in Figure 3-5, there is a large discrepancy in impedance curves even between identical electrode types per trial. As only 3 trials were performed for each electrode, Figure 3-6 demonstrates the 95% confidence intervals over time with an increasing level of confidence shown as time goes on. Once again, looking at Figure 3-7, the effects of impedance mismatches can be derived. Due to the capacitive coupling of the body to the domestic line supply and ground, a 60Hz voltage is developed on the body and is common to both electrodes and referred to as common mode (CM) voltage. In an ideal scenario with identical impedance electrodes experiencing the same level of CM, the differential nature of the amplifier, V1-V2, will eliminate the 60Hz CM signal entirely.

Figure 3-8 and Figure 3-9 exhibit the standard deviation of impedance between all trials for every electrode tested, highlighting the evident large mismatches possible.
Figure 3-8 Impedance Standard Deviation between Trials vs Time
Looking at Figure 3-7, the voltages present at the input, Va and Vb with respect to Vcm, can be calculated, once again, using the voltage divider rule and is shown in (6). Combining (6) and (7) with the assumption that Rin >> Re1 or Re2, (8) is arrived at.

\[
V_a = \frac{R_{in}}{R_{in} + R_{e1}} V_{CM} \quad V_b = \frac{R_{in}}{R_{in} + R_{e2}} V_{CM} \quad (6)
\]

\[
V_{out} = A_d (V_a - V_b) \quad (7)
\]

\[
V_{out} \approx A_d \frac{R_{e2} - R_{e1}}{R_{in}} V_{CM} \quad (8)
\]

If one standard deviation, as shown in Figure 3-8, is used to represent the impedance mismatches (Re2-Req), and the CM voltage on the body can be on the order of 15V [56], the CM voltage introduced to our amplifier at 60Hz, as shown in Figure 3-10 and Figure 3-11, can be derived. Considering a typical 10uV-5mV range for EOG signals with traditional electrode placement [27], it is evident that the 60Hz CM voltage, without any low pass filter (LPF), will completely bury any signal of interest.
Figure 3-10 Amplifier Input CM Voltage (60Hz) vs. Time (No Filter, mismatch of one SD)
Although a 0-100Hz bandwidth is common for a clinical setting [27], the author of this thesis has found a 15Hz LPF filter to be an acceptable frequency for interpreting EOG signals from previous work. Figure 3-12 depicts the results of applying a 15Hz 1st order LPF to the CM voltage. The benefit of the filter on the 60Hz CM noise results in a 10 fold factor of attenuation in noise levels.

Utilizing the lower range of 10uV for EOG signal level, the signal to noise ratio (SNR) per filter order was calculated and displayed as shown in Figure 3-13. It is evident the effect that settling time has on the overall SNR by noting the upward shift of the lines as time goes on, implying a higher SNR for the same order of filter. Comparing the first minute with the last, it can be seen how the individual electrodes are affected with time and with the application of a LPF. Although the TMSI wet electrode demonstrates the best SNR values after 1 minute, the final 25 minute plot displays the dry Ag-Ag/Cl electrode, the Zoflex and conductive rubber (hidden under Zoflex in plot) actually
achieving higher SNR values than that of the TMSI. The associated times for these
electrodes to achieve a higher SNR value than that of the TMSI are shown in Table 3-6.

<table>
<thead>
<tr>
<th>Electrode</th>
<th>Time for SNR &gt; TMSI SNR (s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ag-Ag/Cl</td>
<td>758</td>
</tr>
<tr>
<td>Conductive Rubber</td>
<td>891</td>
</tr>
<tr>
<td>Zoflex</td>
<td>936</td>
</tr>
</tbody>
</table>

Table 3-6 Settling Time for SNR of Electrodes to Exceed SNR of TMSI Electrode

Although positive and encouraging, the settling experiment results alone do not
provide a full picture of electrode behavior and characteristics. These items are further
explored in the sections below.
Figure 3-12: Amplifier Input CM Voltage (60Hz) vs. Filter Order vs. Time
Figure 3.13 SNR vs Filter Order over Time
3.2.2 Electrode Sweep

It is often important to understand how a system will respond at varying frequencies. In the case of electrodes, without a frequency sweep analysis, there may be unknown frequencies where the impedance significantly drops. Despite the fact that most systems will have a low pass filter at the input, there may be enough power in an unexpected bandwidth to cause unwanted noise and complications. It is therefore desirable to be prepared and know what you are dealing with so a proper filter order can be applied. Furthermore, utilizing the results of a frequency sweep, it is possible to derive Warburg model parameters as shown in Figure 3-14 below. One such parameter, the capacitance, can play a significant role in the DC response of an electrode and its associated motion artifact.

![Figure 3-14 Experimental Setup Model](image)

Utilizing this simplified model, the impedance of the skin and tissue was neglected due to their relatively small values compared to that of the electrodes and the assumption
made that Electrode 1 and Electrode 2 were identical. This assumption allowed the final results to be divided in half to arrive at an equivalent model for one electrode.

3.2.2.1 Experimental Procedure

Immediately following an impedance settling experiment as outlined in 3.2.1.1, such that the electrodes remained in a ‘settled’ state, the parameters of Table 3-7 were set on the VersaSTAT 3 to enable a frequency sweep from 0.01 Hz to 100 KHz. The patient was once again instructed to remain steady to avoid motion artifact corruption of the signal while the data was being captured.

After all data was collected, the experiment was repeated as outlined in 3.2.1.1, with only a 3 minute settling time waiting period to provide sweep results for a relatively unsettled state.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Settings</th>
</tr>
</thead>
<tbody>
<tr>
<td>Start Frequency</td>
<td>0.01 Hz</td>
</tr>
<tr>
<td>End Frequency</td>
<td>100 KHz</td>
</tr>
<tr>
<td>Point Spacing</td>
<td>Logarithmic</td>
</tr>
<tr>
<td>Points Per Decade</td>
<td>10</td>
</tr>
<tr>
<td>Total Points</td>
<td>71</td>
</tr>
<tr>
<td>Time</td>
<td>~17 min</td>
</tr>
<tr>
<td>Applied Voltage Amplitude</td>
<td>10mV RMS</td>
</tr>
</tbody>
</table>

Table 3-7 Electrode Impedance Sweep Procedure Settings

The sweep, however, began at 0.1Hz as opposed to the previously used 0.01Hz as the post 27 minute settling time results indicated a relatively flat low frequency response, where the extra decade was felt to not be necessary. All results are the average combination of 6 trials, with standard deviation bars added where applicable.
3.2.2.2 Sweep Results and Discussion

The results of the sweep can be found in Figure 3-15 and Figure 3-16 as the 'experimental' curves. The value of $R_s$ was taken from the Subject-Impedance Region of the experimental plots where the Subject Impedance Region was evident. Rubber, Au, Ag and Zoflex however did not seem to exhibit a Subject-Impedance Region in the 100 KHz range of sweep available from the VersaSTAT. Due to this, an approximation was made that the value of impedance at the 100KHz frequency was that of $R_s$.

The value of $R_p$ was derived by subtracting $R_s$ from the average impedance value of the Faradic Region. The capacitance was then calculated by rearranging the Impedance Equation (1) and substituting the appropriate 3dB frequency. With $R_s$, $R_p$ and $C$, a 'calculated' curve for each of the electrodes in Figure 3-15 and Figure 3-16 was produced using Equation (1). The mean of the squared error between the experimental and calculated values, the mean squared error (MSE), was used to iteratively adjust the value of the calculated capacitance $C$. The value of capacitance was reduced by 1% each iteration until the MSE between the calculated and experimental curves was at a minimum. This new value of $C$ was then used to produce the final 'MSE Correction' curves of Figure 3-15 and Figure 3-16.
Figure 3-15 Impedance Sweep after 3 Minutes of SETing
Figure 3-16 Impedance Sweep after 27 Minutes of Setting
The Zoflex and rubber electrodes appear to exhibit the largest initial MSE from experimental to calculated, with Zoflex continuing this trend after the iterative MSE correction routine. This is most likely an unfortunate result of the oversimplified model parameters lacking the accuracy to represent the more complex nature of these electrodes.
After 27 minutes of settling, the averaged sweep results are plotted in Figure 3-16. With the extra 24 minutes of settling provided, the results appear to lessen the initial and final MSE values of the Zoflex electrodes but at the same time the results for stainless steel and gold electrodes appear worse.

Figure 3-17 depicts the extracted capacitance model parameters, after 3 minutes, and how they compare with the expected values based on their surface areas. The TMSI electrodes appear to exhibit the largest error from surface area to expected capacitance with an unexpectedly large value. The rubber electrode exhibited the opposite effect.

Figure 3-18 provides the same surface area values alongside the updated capacitance and MSE values. The settled results appear to agree much better with the exception of TMSI and SS electrodes. The calculated capacitance of the SS electrodes is much less than the expected value based on surface area. This however may be due to the actual surface area in contact with the skin being reduced due to its domed shape. The TMSI value is significantly lower (roughly 50%) than its initial unsettled value but still higher than the expected value based on surface area. It is possible that an excessive amount of electrode paste was used or perspiration existed under the outer adhesive ring, thereby increasing effective surface area of the electrode.

3.2.3 Motion Artifact Experiment

As discussed in 3.1.5, motion artifact is a direct result of changes in the area or distance between charge layers present in the epidermis or skin-electrode/electrolyte interface. Because it is not possible to realize the ideal non-polarizable electrode as in 3.1.3, motion artifact will always exist to some extent in all real-world biomedical
systems utilizing electrodes. With this in mind, it was necessary to assess the selected electrodes of Table 3-3 for their associated response to motion. These results provide the clinician or designer with another factor to consider when selecting the type of electrode for their EOG application.

3.2.3.1 Experimental Procedure

One of each of the 7 electrodes under test was affixed to a solid plate of ABS plastic and spaced 20mm apart (Figure 3-19). The plate had a vertical protrusion connected at its center with which the perturbations could be delivered, causing all electrodes to move in a uniform and equal fashion.

The electrodes were cleaned with alcohol before each trial and attached to the forearm of the subject with a Red Dot [63] reference electrode placed nearby.

In order to ensure all electrodes received a consistent level of perturbation throughout each trial and from one trial to the next, the apparatus shown in Figure 3-20 was designed. The motion artifact device delivered a consistent perturbation at a 2 second interval and the resulting artifact voltages were measured using 7 channels of an

Figure 3-19 Electrode Array

Figure 3-20 Motion Artifact Experiment Apparatus
instrumentation rack [64], with respect to the reference electrode. The gains were set to 100 and accounted for in the processing. Four trials were conducted, with each one 29 minutes in length.

3.2.3.2 Results and Discussion

The collected data, a sample of which is shown in Figure A-1, were processed at one minute intervals. At each interval, the first 4 peak motion artifact voltage levels, that occurred every 2 seconds due to the apparatus perturbations, were averaged after removing the DC offset. Figure A-3 contains the average motion artifact voltage levels every 4 minutes with 95% confidence intervals (black) and standard deviations (red) overlaid. As with Figure 3-6, the confidence intervals become smaller with time. The averaged results of the 4 trials over the 29 minute length experiment are displayed in Figure A-3.

The results shown indicate the TMSI electrode, being the only wet and adhesive electrode, was outperformed by the Zoflex, rubber, dry Ag-Ag/Cl and gold electrode. Of particular interest is the consistency and performance of the Zoflex electrodes with respect to overall spike levels and time. The confidence levels and standard deviations of the Zoflex electrodes throughout the experiment are superior to all others with the average magnitude of motion artifact surpassing all others by the 5 minute mark.

It is also worth noting the exceptionally poor results from the stainless steel electrodes in both overall average motion artifact magnitude and also variation between trials. These electrodes are a common choice in the prosthetic industry for myoelectric control.
4 EOG Signal Analysis

4.1 Introduction

Electrode choice plays a significant role in biomedical system design as it is the first point of acquisition and therefore sets the bar for maximum signal to noise ratio (SNR) possible. However, even with the most ideal electrode possible, limitations are set by the amount of information available based on factors such as proximity to the signal origin, number of electrodes used, inter electrode spacing, etc. On the same token, these parameters must be balanced with the actual needs of the patient such as aesthetics, practicality and functionality. Hundreds of electrodes cannot be placed on the face without affecting the practical and aesthetic aspects of design. Alternatively, a single electrode at an extreme proximity to the face or at a location occupied by other equipment such as a BiPAP is also not a feasible option.

To help balance these parameters, it is necessary to first attempt to gather as much objective data concerning all electrode number and position scenarios possible. With these metrics, it is then possible to weigh them against the more subjective ones to enable informed adaptation to each patient’s needs.

4.2 The EOG Signal

In order to gather the data required to make objective and subjective design decisions, it is important to first understand the signal being dealt with. The amplitude levels for EOG of 10 µV to 5 mV are based on a standard electrode placement as shown in Figure 2-6. This range is not indicative of some electrode placement positions as shown in
Figure 4-1 where an EOG signal captured from a non-standard location with a 4uV peak voltage is shown.

![Typical EOG Signal](image)

Figure 4-1 Typical EOG Signal

4.3 EOG Signal Subject Repeatability

As with any biomedical experiment, it is desirable for the signal of interest to exhibit repeatability from subject to subject. In the case of EOG, the question is how the EOG signal amplitude and spatial distribution vary, or repeat, from one subject to another. With good repeatability it is possible to make subsequent measurements and designs based on representative subjects.

4.3.1 Experimental Procedure

To facilitate a true comparison of signal repeatability from subject to subject, a high electrode-density bio-potential acquisition system was used (REFA™, TMS International [59]). Five healthy subjects, aged 20-40, were selected with fifty-three electrodes placed
from the hairline of the forehead down to the area surrounding and including the ears, along with a reference electrode at the back of the neck (Figure 4-2).

![Figure 4-2 Electrode Placement](image)

Each subject was asked to sit in a chair with their heads positioned in roughly the same location. Markers were placed at five different positions in the room to provide consistent rotation angles of the eye from trial to trial and subject to subject. A marker was placed on the ceiling for an ‘up’ glance, the floor for a ‘down’ glance, to the right of the subject for a ‘right’ glance and the left of the subject for a ‘left’ glance. A final marker was placed on the wall directly in line with the resting level of the subject’s eyes and referred to as the ‘center’ or ‘rest’ marker. Once the subject was comfortable with looking at each of the markers, the experiment began. The subject was instructed to perform each of the following glances upon request, while each of the 53 electrodes recorded their associated signal:

- Sharp left glance with return to center.
- Sharp right glance with return to center
- Sharp up glance with return to center
- Sharp down glance with return to center
- No eye movement, eyes on center
4.3.2 Processing

The data were segmented into the appropriate files representative of the eye glance. They were then processed through a 15Hz software low-pass filter (LPF) and the DC offset removed by subtracting a ‘resting’/‘center’ glance data capture from the signals of interest. The signals were normalized to subject one and the amplitude for each subject/electrode found by averaging the signal (S1) window of time as shown in Figure 4-1.

Subject 1 was arbitrarily chosen with which to compare all other subjects to. The mean of the square of the difference between normalized amplitudes between any given subject and subject 1 was calculated as in (10). This value was then divided by the mean square of the normalized amplitudes of Subject 1 as in (11) providing the percent difference from any subject to subject 1.

\[
MSD(S_x) = \text{mean}[(S_1 \text{Amplitudes} - S_x \text{Amplitudes})^2]
\]  

(10)

\[
\% \text{Difference}_x = \frac{MSD(S_x)}{\text{mean}(S_1 \text{Amplitudes}^2)}
\]  

(11)

4.3.3 Results

Figure B-2 depicts the normalized values for each subject for each glance type per electrode. Visually, it is evident the strong repeatability of the EOG signal on the surface of the face covered by electrodes for each subject.

To further illustrate this point, Table 4-1 displays the actual percent differences between the average of each subject’s electrodes and that of the average of subject 1’s electrodes. The largest discrepancies occur with an up glance for most subjects with the largest being 6.5% and the overall average difference being 2.1%.
4.4 EOG Amplitude Distribution Map

With the very strong repeatability results previously demonstrated, a map of the steady state signal amplitude distribution was obtained for subject one (Figure 4-3).

4.4.1 Experimental Procedure

The same procedure as outlined in 4.3.1 was utilized to gather data about the distribution of the EOG signal.

4.4.2 Processing

The captured data underwent the same preliminary LPF and pseudo-HPF methods as previously discussed in 4.3.2. The data was then processed to take the average amplitude of the EOG signal for each electrode once the subject’s eye was in the instructed position. The value of these amplitudes were then used to create a colored contour map that was overlaid onto pictures of the subjects face for each eye movement scenario including the ‘no-movement’ case (Figure 4-3).

4.4.3 Results

The contour maps shown in Figure 4-3 give a visual representation of not only signal strength throughout the facial region, but also how the EOG signal’s potential varies with
the movement of the eyes. Each of the four defined eye movements along with a 'no-
movement' case are presented with contour maps overlaid on three angles of the face
(front, right side and left side) to ensure a full view of all electrodes used in the study.
Presenting the results in a colored contour manner provides immediate visual
representation of how the signal strength varies and changes throughout the facial region
for varying eye movements without the need to sift through multiple data plots. Contour
colors are scaled by the largest signal strength of all glance types (red) with respect to the
smallest of all glance types (blue).

Figure 4-3 Colour Contour Map of EOG Signal Presence Over Face
4.4.4 Discussion

The first area of interest arises in the ‘no-movement’ case where the resting potential measured at all electrode locations shows little variation in color other than a slight deviation most likely due to the subject not looking directly ahead but instead somewhat up and to the left. Next it can be noted that as the eyes move to the right, left, upwards or downwards, the potentials measured at each electrode rises or falls based not only on distance from the eyes, but also based on the direction the eyes are pointing. However, certain locations appear to be insensitive to the movements of the eyes. These locations appear where the electrode was placed in line with the axis of horizontal rotation of each eye (directly above the center of the eyes). This fits in line with our understanding of the dipole-like behavior of the eyes such that any potential changes measured at any given location are a direct result of a change in distance from the more positively charged cornea with respect to the retina. Therefore if the electrodes are placed in line with the axis of rotation of the eyes, the movement of the cornea with respect to those electrodes is minimal.

Finally it can be seen that, aside from the aforementioned electrodes placed in-line with axes of rotation, that all locations exhibit a change in potential (color variation) as the eyes rotate. This means that nearly all electrode placement locations are viable based purely on signal presence.

4.5 Signal Interpretation

With an understanding of the locations with which EOG could be acquired on the face, a more quantitative measure was needed to assess the integrity of these signals and
interpret them for ATS use. As was discussed in 2.4.2, the difficulties experienced in previous EOG devices was due to the variation of electrode placement, eye movement and environmental factors from day to day and patient to patient. When used with high care ALS patients, the adjustments of any parameters and settings proved a challenge for the caregivers, who in most cases change from day to day, with all having little to no time to spare.

It was evident that an algorithm capable of handling these issues was necessary. One such method for recognizing repeated patterns is known as pattern recognition. Pattern recognition can be defined as the ‘classification of data based on knowledge already gained or extracted from patterns and/or their representations’ [67]. This method typically involves first training the system with the intended patterns such that it can more easily recognize, or classify, them during regular operation.

4.5.1 Vector Projection

Developed by Dr. Yves Losier of the University of New Brunswick [68], Vector Projection (VP) is a pattern recognition derived algorithm. Although most pattern recognition systems could address the bulk of the aforementioned issues, VP was designed to address the limitations of many traditional pattern recognition algorithms by enabling proportional outputs and real-time adjustment parameters to lessen the need for re-training every time a factor affecting the sensor inputs is varied (temperature, slight position adjustment, etc). As VP was specifically targeted for clinical operation by providing very intuitive control and operation it was chosen to interpret the EOG signal
and provide a means of classification accuracy. The algorithm operates through 3 fundamental stages:

4.5.1.1 Stage 1: Training Data

Training data is used to take the output value (amplitude) of each sensor (electrode) and create localized vectors from the origin to class centroids representing, in the case of EOG, the movements of the eyes the patient wishes to train (ex. up, down, left and right). Figure 4-4 [68] provides a visual of such a vector, $\vec{V}_{OX}$, as it points from the input space origin to the class centroid $C_x$, which could represent any of the aforementioned eye glances. By including a rest centroid in the training, $\vec{V}_{ORest}$, a class specific vector, $\vec{V}_{RestX}$ can be created from the rest centroid $C_{Rest}$ to a class centroid $C_x$. Finally, a class specific vector can be created from the rest centroid to each of the centroids representing a trained input as shown in Figure 4-5.

![Figure 4-4 Local and Class Vectors](image1)

![Figure 4-5 Class Centroids](image2)
4.5.1.2 Stage 2: Real-Time Input

Once the class centroids and vectors have been created with training data as previously described, the system may then be used to gather real-time data. At a chosen time interval, each of the sensor values are updated and used to create a new centroid, $C_{current}$ (Figure 4-6). This centroid has an associated class vector, $\vec{V}_{Rest \ current}$, which is compared to each of the existing class vectors (for example, $\vec{V}_{Rest_1}$, $\vec{V}_{Rest_2}$ as in Figure 4-6) such that an angle and pseudo-projection magnitude are passed to the output stage for classification.

![Figure 4-6 Input Vector Pseudo-Projections](image)

4.5.1.3 Stage 3: Output Algorithm Tuning Parameters

Utilizing the previously computed values, the final stage applies tunable parameters to the input vector projections to determine the output strength of each class. These directly adjustable parameters used are listed below with the examples assuming a patient trained the system with 4 class vectors, Up, Down, Left and Right (along with Rest).
Threshold Factor (TF):

The Threshold Factor allows the user to create a region of inactivity whereby the associated class output will not respond if the projected input vector magnitude falls within that region.

**Example:** With threshold set to 0.3 and an upward glance, the Up class output will begin to respond sooner than that of an upward glance where the threshold is set to 0.7. If an upward glance is performed while below the threshold, the associated class output remains at zero. This setting helps to remove any unwanted false positives as the eyes remain in a relatively rested position by giving a wider inactive zone to accommodate for small movements (Figure 4-7).

![Threshold Factor](image)

Figure 4-7 Threshold Factor

Amplitude Factor (AF):

The Amplitude Factor scales the level at which the projected input vector will reach 100% of the associated class output. If the value is set above 1, a projected input vector
will require a larger amplitude than that which was set in training to reach the maximum output whereas if a value of less than 1 is used, a smaller amplitude is required.

**Example:** As the user moves their eyes upwards, the Up class will output 100% at a glance level lower than that which previously trained if the AF was set below 1 whereas it will reach 100% only at a glance further upwards than the trained level if the AF was set above 1. In practice it has been found that during training, a user will train with eye glances farther than they will typically reach when in use where setting the AF below 1 allows the user to reach 100% output without straining their eyes. Furthermore, as fatigue sets in, the level with which a user glances lessens and would result in little to no class outputs if the AF could not accommodate without re-training (Figure 4-8).

![Figure 4-8 Amplitude Factor](image)

**Spread Factor (SF):**

The spread factor is used to adjust the angular sensitivity of the projected input vector to that of the output class vector, that is to say, the associated class output will weaken at
a rate proportional to the SF as the angle between the input vector and class vector increases.

**Example:** If a large SF value is used, a glance to the upper left or upper right will result in a larger Up class output value than if a smaller SF value was used where the glance must stay more true to an exact upwards glance as was performed in training. In practice, most eye movements used in training are not exactly repeatable and are affected by factors such as head position and environmental changes. A larger SF value allows a level of error with respect to the original glance direction that minimizes the need for retraining if any environmental, physical, or mental changes occur (Figure 4-9).

![Figure 4-9 Spread Factor](image)

With a basic understanding of how the VP pattern recognition algorithm operates, the following chapter utilizes the VP algorithm output results to help assess the attainable accuracy at different locations and combinations of electrodes on the face.
5 EOG Classification Performance

In order to assess the EOG classification performance when using a selected set of electrodes and VP classification, the system was trained and tested with five subjects.

5.1 Procedure

The set up and procedure of 4.3.1 was once again utilized with a subset of 27 of 53 electrodes used at the hairline and around and on the ears (Figure 5-3). These electrodes were chosen based on their aesthetic appeal for an ALS patient as it would prevent wires from being draped over the face and avoid interfering with most BiPAP masks. By cutting down the total number of electrodes used from 53 to 27, significant time savings were achieved for computations. The 4 eye glance directions, along with the rest/center position, were used to train the algorithm resulting in centroids such as those shown in Figure 4-5. Following this, the user was then instructed to perform 2 trials of 2 different patterns, namely, ‘Pattern A’ and ‘Pattern B’ as shown in Table B-1. Each eye movement was sustained for roughly 2 seconds each, providing at least 100 samples post filtering for classification.

5.2 Processing

The data was once again put through a 15 Hz LPF and the DC offset removed by subtracting a “resting” eye glance from each capture.

The ‘S1’ signal portion of Figure 4-1 for each of the 4 glances, along with the rest, were used by the training algorithm to provide class centroids for each electrode that
were passed to the VP algorithm during operation. The training was performed once for each pattern/trial per subject.

Upon training completion, test data obtained for the two trials of both patterns was processed for each subject. As the VP algorithm can operate with any number of inputs/electrodes, it was decided to repeat each trial for each possible combination of 2 (351 combinations), 3(2925 combinations), 4(17550) and 5(80730) inputs/electrodes. This would allow a further analysis of the effect of multiple inputs on the overall accuracy.

The spread factor was set to 1 to increase the chances of proper classification if the eye movements were not exactly in the same direction as trained. The amplitude factor was set also set to 1 such that no scaling was done on the outputs and threshold factor to 0.3 to provide a small region where a rest class could be considered while still allowing the classifying of weaker eye movements than were trained. These settings as discussed are shown in Figure 4-7 (i).

The data was passed one sample at a time to the VP algorithm where its output value, from 0-1, was provided for each of the 4 trained eye movements (up, down, left, right). If all the output values were 0, the sample was classified as ‘rest’, otherwise, the sample was classified as the eye movement with the largest output value.

In order to determine output accuracy, it was necessary to time stamp each pattern as a certain eye movement based on the knowledge of what the subject was asked to do as shown in Figure 5-1. As the movement of the eye from one position to another is not instantaneous, there will always be a finite period of time in which these transitions will take place. One such transition can be seen in Figure 4-1 as ‘T1’ where the eye rotates
from a resting position to a left glance. To examine the effect of the algorithm with and without transitions, the data was also manually classified without the transitions as shown in Figure B-1.

If the output of the VP algorithm matched the manually classified data, that sample was assigned a value of 1. If the output of the VP algorithm did not match the manually classified data, a 0 was assigned. The overall accuracy was the average value of the whole sample length of data.

5.3 Results and Discussion

Figure 5-2 contains the Accuracy results for all possible combinations of 2 electrodes being used as inputs to the VP algorithm. The two trials of ‘Pattern A’ and two trials of ‘Pattern B’ are shown as the 4 sub-plots with the accuracy results for each of the 5
subjects contained within each one. The dashed lines represent the accuracy results when transitions were included whereas the solid lines represent the accuracy when the transitions were removed.

Figure 5-2 Classification Accuracy for All Possible Combinations of 2 Inputs/Electrodes
There is a wide spread of accuracy results not only from subject to subject, but also from trial to trial on the same subject. Subject 1 outperforms all other subjects in all but the second trial of 'Pattern A' achieving 100% accuracy for nearly 40% of all combos in the second trial of 'Pattern A' with no transitions. Subject 5 demonstrated poor results on 'Pattern B' but also indicated cognitive issues associated with fatigue, especially with 'Pattern B' being the last portion of the experiment and requiring multiple trials for the subject to complete it successfully according to their perception.

For every subject, the average of all trials were taken with the top 10 results for each subject for the 2 electrode combinations shown in Table 5-1. Adjacent to each subject's accuracy are the associated standard deviations, with both the accuracy and standard deviations results averaged and presented at the bottom of each column. The overall poor results of subject 5 are apparent with a top 10 average of 68.1% and a nearly 20% standard deviation. Subject 1 has an overall average of 90.5% but a rather large standard deviation of 17.6%. This standard deviation becomes apparent when observing the poor results of subject 1's second trial of pattern 'A'. The remaining subjects maintained a roughly 90% accuracy with close to 10% standard deviation.

As the number of electrode combinations increased to 5 (see Appendix B), subjects 2, 3 and 4 had a 7-10% increase in accuracy with standard deviations dropping to 2% or less. Subject 5 had an overall accuracy increase but still suffered from large standard deviations due to the multiple poor trials. Due to the nearly 100% accuracy for 3 of the 4 trials that subject 1 completed with the single bad trial, the results did not change from 2 to 4 electrodes.
These results demonstrate that as the number of electrodes used as inputs to the VP algorithm increases, there are more options for electrode choices to produce similar accuracy results. Therefore, if a subject, or patient, encounters varying difficulties from one trial/day to the next, it would be prudent to increase the number of electrodes used to provide a level of consistency in operation.

Although the addition of electrodes may provide an increase in consistency, there is still a need to understand what locations on the face can typically provide better results than the others. The results of all subjects and trials, with the exception of subject 5 who encountered consistently poor results due to the aforementioned issues, were pooled together. The average accuracy for each electrode combination across subjects 1 to 4 was taken and then sorted according to the top 10 accuracy results with the corresponding electrodes used as shown in Table 5-2 for the 2 electrode combination case.

The results suggest that a small subset of electrode combinations/locations can be selected which will give acceptable performance across the general client population.
The associated electrode positions can be found in Figure 5-3. The most common electrode, 12, is naturally one of the closest to the eyes. Electrodes 16 and 17, just beside the horizontal mid-line of the ear, also appear multiple times in all cases. The results demonstrate that non-traditional electrode positions are viable for eye position classification.

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Average 81.4%

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Average 87.3%

Table 5-2 Top 10 Electrode Results for All Subject/Trials Pooled (2 & 5 Electrodes)
6 Conclusion

6.1 Summary

A theoretical and experimental overview of many aspects for design of an EOG acquisition device for the disabled was performed and documented. The origins of the signal, the inherent problems associated with acquisition and the current techniques used were discussed.

Seven electrodes, all small enough to be potential candidates for a commercially viable device, were presented with several measures of performance experimentally assessed and provided for consideration. Of particular interest were the performances of the Zoflex and conductive rubber electrodes, both of which require no preparation compared to a conventional wet adhesive Ag-Ag/Cl electrode.

The similarities in the distribution of EOG signals from subject to subject were explored with strong results indicating that any given subject will be representative of another. The presence of EOG signals throughout the face was assessed with all locations proving viable, albeit providing variable classification performance. Analysis of the EOG signal was introduced with the application of a vector projection pattern recognition algorithm discussed. The algorithm was used to classify the movements of the eyes and the overall accuracy found to be near perfect for some combinations of electrodes, but dependent on the subject. Several locations were selected and shown to typically provide the best results, with the variation between trials decreasing as the number of electrodes used increased.
The contents of this thesis will enable an informed starting point for future development of a commercially viable EOG acquisition device for the disabled and prevent unnecessary experimentation and setbacks due to improper design choices.

6.2 Contributions

This work made the following contributions:

1. A full overview of EOG signal origin pertinent to acquisition for ATS use was not found in any of the literature reviewed by the author. The information is provided in this thesis.

2. A thorough comparison of seven electrodes, each of appropriate size and material for a commercial design, was provided. This overview offers the designer an informed decision on the pros and cons of each electrode and how it may affect the performance of their system in regard to impedance, settling time, and SNR.

3. The spatial distribution of EOG signal amplitude was mapped and found to be very consistent over 5 subjects, thus making for easier electrode location selection for a given patient.

4. The classification performance as a function of electrode number and location was investigated which provides information regarding configuration choice to achieve a required performance.

5. Finally, the work of Dr. Losier's VP algorithm was shown to transfer from the residual shoulder motion application, with which it was first demonstrated, to an electrode based EOG acquisition setup. In doing this, the performance of the algorithm was used to assess its viability for EOG use.
All of these contributions will aid in expediting the design process of any future developed EOG devices without the need for reproducing any of the extensive research provided in this thesis.

6.3 Future Work

All of the data gathered to date was done so with the use of existing hardware such as the TMSI Refa and general purpose signal acquisition instrumentation. A custom EOG acquisition device has been designed by the author, and is to be implemented.

The use of the vector projection algorithm was implemented on a PC with MATLAB and not performed in real-time. The VP algorithm has been implemented on a PIC microcontroller by the author and proven to operate in real-time, and will be tested in an EOG application.
Bibliography


[22] November 2013. [Online]. Available:
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[40] J. Behrman, "Electrodiagnostic tests in eye disease., 40, 725-6.," Postgraduate


[58] February 2014. [Online]. Available:


Appendix A Motion Artifact Plots

Figure A-1 Sample Motion Artifact Raw Data
Figure A.2: Average Motion Artifact Over Time with Standard Deviation and 95% Confidence Intervals.
Figure A.4 Motion Artifact Plots: All Trial
Appendix B Classification Supplementary Plots & Tables

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Table B-1  Experiment Movement Patterns
Figure B-1 Manually Segmented Pattern (Transitions Not Included)
Figure B-2 Normalized EOG Amplitude Subject Comparisons
Figure B.3: Classification Accuracy for All Possible Combinations of 3 Inputs/Electrodes.
Figure B.4: Classification Accuracy for All Possible Combinations of 4 Inputs/Electrodes
Figure B.5 Classification Accuracy for All Possible Combinations of 5 Inputs/Electrodes
Table B-2 Top 10 Results per Subject Over All Trials (3 Electrodes, 1 Standard Deviation)

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Table B-3 Top 10 Results All Subject/Trials Pooled with Corresponding Electrode Combos (3 Electrodes)

Table B-4 Top 10 Results per Subject Over All Trials (4 Electrodes, 1 Standard Deviation)

Average 90.8% 17.7% 98.1% 2.6% 96.9% 3.8% 98.7% 1.3% 79.9% 16.1%
Table B-5 Top 10 Results All Subject/Trials Pooled with Corresponding Electrode Combos (4 Electrodes)

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Table B-6 Top 10 Results per Subject Over All Trials (5 Electrodes, 1 Standard Deviation)

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96
Curriculum Vitae

Candidate’s full name: Joshua Edward Keys

Universities attended:

University of New Brunswick, Fredericton, NB, Canada
B.Sc.E (Electrical and Computer Engineering), 2006

Conference Presentations:
