Library Authorization

In presenting this thesis in partial fulfillment of the requirements for an advanced degree at the University of British Columbia, I agree that the Library shall make it freely available for reference and study. I further agree that permission for extensive copying of this thesis for scholarly purposes may be granted by the head of my department or by his or her representatives. It is understood that copying or publication of this thesis for financial gain shall not be allowed without my written permission.

Gregory G. Forrest
Name of Author (please print)

10/09/2004
Date (dd/mm/yyyy)

Title of Thesis: Investigation of an EMG Referenced Control Channel for Grasp Force Supplementation

Degree: Master of Applied Science

Year: 2004

Department of Mechanical Engineering

The University of British Columbia
Vancouver, BC Canada
Abstract

This work investigates the potential of using an air-muscle actuated orthosis controlled by an electromyographic (EMG) signal to reliably supplement the grasping force of the hand, thereby allowing the user to reduce the muscle activation required for a power-grasping task. In particular, the study reported herein tested the hypotheses that subjects could stably handle objects and learn to reduce both their grip force and muscle activation levels with force supplementation. In this study, a surface-mounted EMG sensor on the flexor digitorum provides the input to a proportional-integral-derivative controller governing the force generated by the orthosis. Although this approach presumes that the human motor system will stably adapt to the orthotic system, we designed the system to operate in an intuitive and predictable manner. Nine subjects performed a sequence of unassisted and assisted lifts of a weighted and instrumented cylinder. When using the orthotic system to lift the cylinder, subjects reliably reduced their mean grip force and mean percent maximum voluntary contraction (%MVC) (p<0.01). The grip force applied to the cylinder was reduced for seven of the nine subjects (p<0.01) and the %MVC was reduced for five of the nine subjects (p<0.01). None of the subjects exhibited any instability or reported any difficulties when using the orthosis. On average, the subjects reduced their %MVC and grasp force by 31% and 56% respectively.
Table of Contents

Abstract................................................................................................................. ii
Table of Contents................................................................................................. iii
List of Tables ......................................................................................................... vii
List of Figures ........................................................................................................ viii
List of Abbreviations ............................................................................................ xi
Glossary .................................................................................................................. xii
Acknowledgements ............................................................................................... xiii

1 Introduction ........................................................................................................ 1
  1.1 Background - Causes of Work-related Musculoskeletal Disorders.................. 1
     1.1.1 Repetition.............................................................................................. 3
     1.1.2 Awkward postures .............................................................................. 4
     1.1.3 Force ................................................................................................... 5
  1.2 Problem Formulation ..................................................................................... 7

2 Review of Existing Technology .......................................................................... 9
  2.1 Gripping Orthosis ....................................................................................... 9
  2.2 Air Muscle Technology ............................................................................. 14
     2.2.1 Air Muscle Modeling ......................................................................... 20
  2.3 Electromyogram Signals and Control.......................................................... 25
     2.3.1 Electromyogram Characteristics ....................................................... 26
     2.3.2 Electromyogram Filtering and Normalization ................................. 29
     2.3.3 General Biomechanical Use of the Electromyogram ...................... 31
     2.3.4 Specific Application of EMG to Handgrip Force Estimation ......... 33
  2.4 Summary ..................................................................................................... 34

3 Methodology and Experimental System Design and Simulation ..................... 36
  3.1 Overview .................................................................................................... 36
  3.2 Simulated EMG Generation ...................................................................... 37
  3.3 Human Grip Motoric and Reactive Models ............................................... 38
  3.4 Orthotic System ........................................................................................ 40
     3.4.1 Overview .......................................................................................... 40
     3.4.2 Electromyogram Signal to Percent Maximal Voluntary Contraction .. 41
     3.4.3 Thresholding Percent Maximal Voluntary Contraction ................. 43
     3.4.4 Hand Force Estimation Model ......................................................... 44
     3.4.5 Air Muscle Model, Valves, and Control System ......................... 44
3.4.6 Determination of the Air Muscle PID Controller Gains ........................................49

3.5 Simulated Results ........................................................................................................50
3.5.1 Verification of Simulated EMG Response .................................................................50
3.5.2 Simulated Unassisted Lift Results .............................................................................52
3.5.3 Simulated Supplemented Lift Results ........................................................................53

3.6 Summary .......................................................................................................................55

4 Experimental Setup ........................................................................................................56
4.1 Overview .........................................................................................................................56
4.2 Air Muscle and Air Muscle Controller Design ..............................................................58
4.3 Pressure Measurement Components .............................................................................61
4.3.1 Pressure Transducer ................................................................................................61
4.3.2 Pressure Transducer Calibration .............................................................................62
4.4 Force Measurement Components ..................................................................................62
4.4.1 Grip Force Transducer .............................................................................................62
4.4.2 Air Muscle Force Transducer ..................................................................................63
4.4.3 Force Transducer Calibration ..................................................................................63
4.5 Length Measurement Components ...............................................................................63
4.5.1 Encoder .....................................................................................................................63
4.5.2 Encoder Calibration ................................................................................................64
4.6 Electromyogram Measurement Components ...............................................................64
4.6.1 EMG Electrode .........................................................................................................64
4.7 Valve and Valve Driver .................................................................................................65
4.7.1 Valve and Valve Modifications .................................................................................65
4.8 Summary .........................................................................................................................66

5 Experiments .....................................................................................................................67
5.1 Initial Setup ......................................................................................................................67
5.2 EMG Electrode Placement and MVC Determination .....................................................67
5.3 Arm and Hand Positioning ............................................................................................68
5.4 Control Experiment – Unassisted Cylinder Lift .............................................................71
5.5 Assisted Lift Experiment ...............................................................................................71
5.6 Validation of the 100 Sample Estimate for MVC Calibration ........................................72
5.7 Preprocessing of Raw Experimental Data ....................................................................73
5.8 MVC Calibration to Maximum Squeeze in Assisted Lift ...............................................74
5.9 Control Experiment to the Normal Lift in Assisted Lift Experiment .............................77
5.10 Analysis of the Reduction in %MVC and Grasp Force between the Control Experiment and the Assisted Lift Experiment .................................................................79
5.11 Typical Experimental Results ......................................................................................81
5.11.1 Typical Control Lift Results .....................................................................................81
5.11.2 Typical Assisted Lift Experimental Results ...............................................................82
5.12 Learning Effect ............................................................................................................84
5.12.1 Immediate Adaptation without Stiffening of the Arm ..............................................84
5.12.2 No Adaptation, Subject Stiffens Arm ....................................................................86
5.12.3 Progressive Adaptation .........................................................................................87

6 Conclusions .....................................................................................................................89

7 Recommendations .........................................................................................................92
Appendix A. Experimental Setup Circuitry

A.01 Pressure Transducer Circuitry
A.02 Force Transducer Circuitry
A.03 Additional EMG Filtering Circuitry
A.04 Valve Driver Design and Circuitry

Appendix B. Open Real-Time System Implementation

B.01 Hardware
B.02 The Structure of ORTS
B.03 ORTS Functions – DSP Side

(a) Delay
(b) Determine Maximum and Minimum Voluntary Contraction and Force
(c) Digital Input/Output to Hexadecimal
(d) Emergency Shutdown
(e) Encoder to Length
(f) Filter
(g) Force and Length to Mass
(h) Impulse
(i) Kill
(j) Link Limit
(k) Mass to Duty Cycle
(l) Percent Maximum Voluntary Contraction to Force
(m) Polynomial
(n) Pressure and Length to Force
(o) Pulse Width Modulation
(p) Pressure Regulation
(q) Root Mean Squared to Duty Cycle
(r) Root Mean Squared to Percent Maximal Voluntary Contraction (MVC)
(s) Route Signal
(t) Sinusoid Generator
(u) Spring Force Calculator
(v) Step Generator
(w) Staircase Generator
(x) Single-Input-Single-Output (SISO) Proportional-Integral-Derivative (PID) Control
(y) Variable
(z) Voltage to Actual
(aa) Voltage to Force (obsolete, surpassed by Voltage to Actual)
(bb) Voltage to Pressure (obsolete, surpassed by Voltage to Actual)
(cc) Pseudo-White Noise Generator
(dd) Wait for Signal
(ee) Windowed Root Mean Squared (RMS)

B.04 Trial Class

(a) Constructors, Destructors, and Operators
## Appendix C. Detailed Experimental Procedure

- **C.01** Experimental Setup Checklist ........................................ 119
- **C.02** Pre-Experimental Setup ........................................ 119
- **C.03** Common Experimental Setup ........................................ 119
- **C.04** Experiment 1 Setup – Base lining Subject .......................... 120
- **C.05** Experiment 2 Setup ........................................ 120

## Appendix D. MVC Calibration Plots ........................................ 122

## Appendix E. Control Experiment Plots .................................... 131

## Appendix F. Assisted Lift Experiment Plots ............................... 146

## Appendix G. Ethics Form ................................................ 164

## Appendix H. ORTS Scripts ........................................ 165
- **H.01** DetMVC.spt ................................................ 165
- **H.02** Exper1.spt (Control Experiment) .................................. 166
- **H.03** EMGCtrl.spt (Assisted Experiment) ............................... 168
List of Tables

Table 2-1: Comparison of Different Actuator Types [28] ..............................................................18
Table 3-1: Average Unassisted Lift %MVC and Grasp Force (N) ..................................................53
Table 3-2: Average Assisted Lift %MVC, Air Muscle Force, and Grasp Force (N) ..................54
Table 4-1: Progression in determining air muscle controller PID gains using subject response ..................................................................................................................61
Table 5-1: Comparison of estimated maximum EMG and force levels to statistically generated levels ..................................................................................................................73
Table 5-2: Student t-test Results between the MVC Calibration and MVC consistency checks for the ratio of %MVC to the Mean Force ..................................................77
Table 5-3: Student t-test Results between the Control Experiment and Unassisted Lift consistency checks for the ratio of %MVC to the Mean Force ..........................78
Table 5-4: Scale factor obtained from the relation between the Control Experiment and Unassisted Lift consistency checks .................................................................79
Table B-1: Types of Filters and Code to Use in Function ................................................................109
List of Figures

Figure 1-1: Stress-Strain Relationship over Time [16] ................................................ 4
Figure 1-2: Incidents of carpal tunnel syndrome for Cycle time vs. Posture [19] .......... 5
Figure 2-1: Examples of Dynamic Splints [36-38] ......................................................... 10
Figure 2-2: Methods of Paraplegic Orthotic Control: (a) Puff control, (b) Bicep EMG
               Control, (c) Chin Control, (d) Back Control, (e) Foot Control [42] ................. 12
Figure 2-3: Shape Memory Alloy Actuated Hand Orthosis [45] ................................. 12
Figure 2-4: Artificial Muscle Actuator Orthosis [54] ........................................................ 13
Figure 2-5: Photo of Air Muscle [66] ............................................................................ 14
Figure 2-6: Construction of an Air Muscle [57] ............................................................... 15
Figure 2-7: Air muscle mesh angle [60] ........................................................................ 16
Figure 2-8: Force versus length relationship comparison for different types of
               muscles [29] ..................................................................................................... 17
Figure 2-9: Two air muscle configurations: Dual Air Muscle Configuration (left);
               Single Air Muscle and Spring Configuration (right) ........................................... 19
Figure 2-10: Air muscle model demonstrating the end effect [52] ............................... 21
Figure 2-11: Geometric explanation of the three derived characteristic parameters
               (thread length, number of turns on each thread, and interweave angle)
               of the air muscle [66] .................................................................................... 22
Figure 2-12: Relation between the number of circumferential trapezoids and
               lengthwise trapezoids and the thread length and number of turns on
               each thread [66] .............................................................................................. 23
Figure 2-13: Demonstration of the hysteresis present in the air muscle actuator with
               constant internal air mass [58] ........................................................................ 25
Figure 2-14: A schematic of the differential amplifier configuration. The EMG signal
               is represented by 'm' and the noise signals by 'n'. [88] ................................. 27
Figure 2-15: Schematic diagram of the factors that affect the EMG signal [93] .......... 27
Figure 2-16: Placement of the EMG sensor on the muscle [93] ................................. 29
Figure 3-1: Simulation Overview .................................................................................... 37
Figure 3-2: Activation signal and simulated EMG generator ........................................ 38
Figure 3-3: Orthosis system overview ......................................................................... 41
Figure 3-4: Comparison between an AM radio signal and an EMG signal ............... 42
Figure 3-5: Fourier Analysis of EMG and Force Spectrums ........................................ 43
Figure 3-6: Air muscle control loop ............................................................................ 44
Figure 3-7: Pressure and length to force model ............................................................ 45
Figure 3-8: Force and length to mass model ................................................................. 46
Figure 3-9: Force and pressure to length model ............................................................ 46
Figure 3-10: Valve and PWM model .......................................................................... 47
Figure 3-11: PWM valve implementation .................................................................... 48
Figure 3-12: PWM for inlet/outlet valve model ............................................................. 49
Figure 3-13: Air muscle model .................................................................................... 49
Figure 3-14: Verification of Simulated EMG and Windowed EMG for a single lift
               with an activation signal of 1 ....................................................................... 51
Figure 3-15: Verification of Simulated EMG and Windowed EMG for multiple lifts
               with varying activation signals .................................................................. 52
Figure 3-16: Unassisted Lift Simulation Results for a single lift with an activation
               signal of 1 .................................................................................................. 53
Figure 3-17: Unassisted Lift Simulation Results for multiple lifts with varying
activation signals .................................................... 53
Figure 3-18: Assisted Lift Simulation Results for a single lift with an activation
signal of 1 .......................................................... 54
Figure 3-19: Assisted Lift Simulation Results for multiple lifts with varying
activation signals .................................................... 55
Figure 4-1: Experimental Setup ..................................... 57
Figure 4-2: The force instrumented cylinder ..................... 57
Figure 4-3: Air Muscle and Force Sensor used in Experimental Setup .... 58
Figure 4-4: Calibration of Air Muscle ................................ 59
Figure 4-5: Polynomial Correction of Air Muscle Force Estimate ............ 60
Figure 4-6: AutoTran Model 250 Pressure Transducer [105] ............. 61
Figure 4-7: Precision Transducers PT4000 [106] .................... 62
Figure 4-8: US Digital S1 Encoder [107] .......................... 64
Figure 4-9: Simon Fraser University Neuromotor Control Lab EMG Sensor [108] ............................. 65
Figure 4-10: Matrix Valve [109] ..................................... 65
Figure 4-11: Valve Restriction Plates (units in cm) .................... 66
Figure 5-1: Flexor Digitorum Profundus adapted from [110] .......... 68
Figure 5-2: Location of the radial styloid process and the ulnar styloid process on the
right hand [111] ...................................................... 69
Figure 5-3: Sensitivity to Wrist Angles given a 50% MVC ...................... 70
Figure 5-4: Comparison of Estimated EMG Level and Maximum Force to Mean
EMG Level and Maximum Force ..................................... 73
Figure 5-5: Example of Anomalous Data for Subject 3 during the Control
Experiment Trial 2 .................................................... 74
Figure 5-6: Comparison of MVC Calibration Data to Maximum Squeeze in Assisted
Lift Experiment ....................................................... 75
Figure 5-7: Comparison of Control Experiment Lift to Normal Lift in Assisted Lift
Experiment .......................................................... 78
Figure 5-8: Percent Reduction in %MVC and Grasp Force ...................... 81
Figure 5-9: Control Lift data for Subject 6 ............................ 82
Figure 5-10: Assisted Lift Experiment data for Subject 6 (excluding consistency
checks) ............................................................... 83
Figure 5-11: Changes in the measured contraction levels (top) and applied forces
(bottom) when the air muscle actuator is activated .................... 84
Figure 5-12: Assisted Lift Experiment for Subject 6 Trial 2 demonstrating immediate
adaptation without subject stiffening arm .......................... 85
Figure 5-13: Assisted Lift Experiment for Subject 8 Trial 1 demonstrating activation
peak in %MVC signal ................................................. 86
Figure 5-14: Assisted Lift Experiment for Subject 5 Trial 3 demonstrating no
adaptation with subject stiffening arm ................................ 87
Figure 5-15: Assisted Lift Experiment for Subject 4 Trial 1 demonstrating
progressive adaptation ............................................... 88
Figure 5-16: Assisted Lift Experiment for Subject 4 Trial 2 demonstrating
progressive adaptation ............................................... 88
Figure A-1: Pressure transducer circuitry ................................ 100
Figure A-2: Transducer Techniques TMO-1 Amplifier/Conditioning Module [117] .... 101
Figure A-3: Additional EMG Conditioning Circuitry ......................... 102
Figure A-4: Physical Implementation of EMG Conditioning Circuitry and Valve Driver Circuitry .......................................................... 102
Figure A-5: Valve Driver Circuitry ........................................................................... 104
Figure A-6: Minimum duty cycle for a given frequency ............................................ 105
Figure B-1: Loughborough Sound Images' PC/C32 DSP Board [120] ......................... 106
Figure D-1: MVC Calibrations for Subject 1 ............................................................. 122
Figure D-2: MVC Calibrations for Subject 2 ............................................................. 123
Figure D-3: MVC Calibrations for Subject 3 ............................................................. 124
Figure D-4: MVC Calibrations for Subject 4 ............................................................. 125
Figure D-5: MVC Calibrations for Subject 5 ............................................................. 126
Figure D-6: MVC Calibrations for Subject 6 ............................................................. 127
Figure D-7: MVC Calibrations for Subject 7 ............................................................. 128
Figure D-8: MVC Calibrations for Subject 8 ............................................................. 129
Figure D-9: MVC Calibrations for Subject 9 ............................................................. 130
Figure E-1: Control Experiments for Subject 1 ......................................................... 132
Figure E-2: Control Experiments for Subject 2 ......................................................... 133
Figure E-3: Control Experiments for Subject 3 ......................................................... 135
Figure E-4: Control Experiments for Subject 4 ......................................................... 136
Figure E-5: Control Experiments for Subject 5 ......................................................... 138
Figure E-6: Control Experiments for Subject 6 ......................................................... 139
Figure E-7: Control Experiments for Subject 7 ......................................................... 140
Figure E-8: Control Experiments for Subject 8 ......................................................... 142
Figure E-9: Control Experiments for Subject 9 ......................................................... 145
Figure F-1: Assisted Experiments for Subject 1 ....................................................... 147
Figure F-2: Assisted Experiments for Subject 2 ....................................................... 151
Figure F-3: Assisted Experiments for Subject 3 ....................................................... 153
Figure F-4: Assisted Experiments for Subject 4 ....................................................... 155
Figure F-5: Assisted Experiments for Subject 5 ....................................................... 156
Figure F-6: Assisted Experiments for Subject 6 ....................................................... 158
Figure F-7: Assisted Experiments for Subject 7 ....................................................... 160
Figure F-8: Assisted Experiments for Subject 8 ....................................................... 162
Figure F-9: Assisted Experiments for Subject 9 ....................................................... 163
### List of Abbreviations

<table>
<thead>
<tr>
<th>Abbreviation</th>
<th>Full Form</th>
</tr>
</thead>
<tbody>
<tr>
<td>ADC</td>
<td>Analog-to-Digital Converter</td>
</tr>
<tr>
<td>CTD</td>
<td>Cumulative Trauma Disorder</td>
</tr>
<tr>
<td>DAC</td>
<td>Digital-to-Analog Converter</td>
</tr>
<tr>
<td>EMG</td>
<td>Electromyogram</td>
</tr>
<tr>
<td>FSO</td>
<td>Full Scale Output</td>
</tr>
<tr>
<td>LED</td>
<td>Light Emitting Diode</td>
</tr>
<tr>
<td>MVC</td>
<td>Maximum Voluntary Contraction</td>
</tr>
<tr>
<td>ORTS</td>
<td>Open Source Real-Time System</td>
</tr>
<tr>
<td>PID</td>
<td>Proportional-Integral-Derivative</td>
</tr>
<tr>
<td>PSI</td>
<td>Pounds per Square Inch</td>
</tr>
<tr>
<td>PWM</td>
<td>Pulse Width Modulation</td>
</tr>
<tr>
<td>RMS</td>
<td>Root-Mean-Squared</td>
</tr>
<tr>
<td>RSI</td>
<td>Repetitive Strain Injury</td>
</tr>
<tr>
<td>SMA</td>
<td>Shape Memory Alloy</td>
</tr>
<tr>
<td>WMSD</td>
<td>Work-related Musculoskeletal Disorder</td>
</tr>
</tbody>
</table>
## Glossary

<table>
<thead>
<tr>
<th>Term</th>
<th>Definition</th>
</tr>
</thead>
<tbody>
<tr>
<td>Air muscle, McKibben muscle, artificial muscle</td>
<td>A nylon braid wrapped over a rubber inner tube that when inflated, contracts and has similar characteristics to biological muscle</td>
</tr>
<tr>
<td>Cumulative Trauma Disorder (CTD), Work-related Musculoskeletal Disorder (WMSD), Repetitive Strain Injury (RSI)</td>
<td>Injury to muscles and tendons that occurs due to excessive straining the musculoskeletal system repeatedly over time</td>
</tr>
<tr>
<td>Electromyogram (EMG)</td>
<td>The electrical signal measured from muscles during activation</td>
</tr>
<tr>
<td>Force Supplementation</td>
<td>Force sharing between the person and the device not exceeding the person’s own natural capacity</td>
</tr>
<tr>
<td>Orthosis, Orthotic</td>
<td>An assistive device used to supplement or provide function</td>
</tr>
<tr>
<td>Percent of Maximal Voluntary Contraction (%MVC)</td>
<td>A normalization method for electromyogram signals where the signal is compared to the maximum of which the muscle is capable</td>
</tr>
</tbody>
</table>
Acknowledgements

I thank Dr. Antony Hodgson for giving me insight and expanding my ideas beyond a narrow focus. Conversely, I thank Dr. Elizabeth Croft for providing direction and focus to my work. Without either of your support, I would never have been able to finally complete this thesis.

I thank Mike Dirk for all his help in designing and constructing the experimental setup. I also thank Navid Boostani and Amir Jahbari for keeping the IAL relatively organized during my stay. I also acknowledge Gord Wright, Glenn Jolly, Doug Yuen, and the rest of the machine shop staff for their help. Dr. Ted Milner for supplying the EMG sensors on extremely short notice. I thank the Advanced Systems Institute (ASI), the University of British Columbia Top Up, and Natural Science and Engineering Research Council (NSERC) for all their financial support.

I thank the professors and friends who helped me through my courses in both my undergrad and during my masters. Explicitly, Dana Kulic for her help with robotics and controls as well as helping me tune my system. My best man, Luke Deutscher, for his vast mathematical and physics knowledge. Dr. Ron Palmer, Klaus Ottenbreit, and David Duguid for their ability to teach with clarity and humor.

I thank everyone in the IAL and NCL who helped with and were willing participants in my experimentation. Specifically, Damien Clapa for all his air muscle knowledge and willingness to share complaints when thesis frustration was getting the better of us. We are finally both done! David Langlois for helping everyone with projects without asking for anything in return.

I thank Dr. Broadway for helping me maintain some of my sanity during my thesis. I also acknowledge the support of my friends, Nancy Price and Sherry Lamb, who introduced me to the humor of MST3K that allowed me to keep the rest of my sanity. I would also thank Brendan Pyle for being a longtime friend; sorry the NDP did not win more seats.

I thank the support of my family. My sisters, Pam and Cindy, for teasing and kidding. My niece, Sasha, for being silly. My brother-in-law and soon-to-be brother-in-law, Mike and Maurizio, for treating my sisters right. My Mother and Father for their love, support, and inspiration throughout the years; listening to the frustration, drying the tears.

Finally, I thank my wife, Dana Forrest. I know this thesis has been just as hard on you as it was for me. I am truly sorry.
1 Introduction

The motivation of this work stems from the importance and potential to reduce chronic workplace injury. Many workers, including my own mother, unknowingly experience progressive injury over the course of their employment, which results in financial losses to the corporation and eventually society. Furthermore, the worker and their families experience significant hardship through the loss of ability.

My mother experiences significant pain in simple tasks around the home such as opening a jar. The potential cause of her problem stems from the fact that, during her employment, she was required to lift stacks of cheques from a cheque-sorting machine. The key to the repetitive strain injury was force. The force required to lift the cheques was too large and over time caused injury to the finger tendons. It is true that her condition could have been prevented or its severity reduced if she was trained to lift fewer cheques; however, productivity is more important than training and this training is likely not performed in many other similar lifting jobs.

The potential to reduce injury using electromyogram (EMG) controlled compliant actuation to assist workers has not been investigated. This work intends to address this deficiency and provide direction for further research into force supplement devices for the potential of reducing work-related musculoskeletal disorders (WMSD) in the future.

1.1 Background - Causes of Work-related Musculoskeletal Disorders

According to the Canadian Centre for Occupational Health and Safety (CCOHS), work-related musculoskeletal disorders account for over half the industrial disease claims in British Columbia [1]. WMSD is a generalized term for many similar ailments such as repetitive strain injuries, cumulative trauma disorders, regional musculoskeletal disorders, and soft tissue disorders. Since
most work requires the use of the arms and hands, most of the WMSDs affect the upper extremities as well as the neck. A definition of wrist disorders given by Kroemer: “a collective term for syndromes characterized by discomfort, impairment, disability or persistent pain in joints, muscles, tendons and other soft tissues, with or without physical manifestations…” [2]. Specific WMSDs that affect the wrist include: tendonitis, a form of tendon inflammation occurring when a tendon is repeatedly tensed; tenosynovitis, a repetitive-induced tendon injury involving the synovial sheath; and carpal tunnel syndrome, in which swelling of the tendons and tendon sheaths compress the median nerve, resulting in pain and numbness in the hands [3].

However, most of the current research in this area is focused on rehabilitation once an injury has occurred, often too late to correct the damage. Research on preventative measures considers methods to improve job design, workplace design, and ergonomic tool design. In some cases, almost half of the physiological cost could be spared if the workplaces were redesigned [4]. Despite such preventative measures, many workers still complain about musculoskeletal pain after prolonged periods of work [4]. During a long-term study at an industrial plant where ergonomic interventions were being implemented, no statistically significant improvements were observed for people suffering from cumulative trauma disorders (CTDs) [5]. In the same study, it was also found that 31% of workers with CTDs had changed jobs, and, more specifically 40% of workers with hand-wrist CTDs had changed jobs. This result implies that the preventative measures were insufficient to return many employees back to the job that lead to the injury. In fact, the number of years on the job was negatively correlated with CTDs, suggesting a survivor effect [5].

According to the American National Institute for Occupational Safety and Health (NIOSH), the causes of WMSDs can be attributed to a combination of repetitive and forceful exertions of the muscles [6]. In particular, carpal tunnel syndrome as well as other WMSDs of the hand and wrist can be attributed to: repetitive hand motions; awkward postures; mechanical stress at the base of the palm; vibration; and force [7-11]. The factors were classified by risk; it was found that the order of risk factors from highest to lowest is: the forces exerted by the wrist and hand, the velocity of movement in flexion-extension, and the level of repetition [12]. Hand and wrist disorders are typical in industries with heavy work such as mining, construction, agriculture, manufacturing, and forestry where seemingly minor movements cause injury due to the repetitive nature of tasks involved [13]. Furthermore, WMSDs have also been found in
workplaces where repetitive handling of lightweights occurs such as in supermarket checkouts and in assembly lines [4].

1.1.1 Repetition

The majority of musculoskeletal injuries are caused by repetition of a hazardous task over time that causes successive micro traumas [3]. Although it is clear that repetition does cause injury, it has also been found that repetition is highly correlated to the forces involved and posture. Nevertheless, the definition of repetitiveness in terms of a physiologic viewpoint is the number of times that a tendon is stretched or squeezed per unit of time [14]. Thus, repetitiveness is a function of the force that the hand applies and the geometry of the tendon.

When designing job parameters, employers and job designers focus on work quantities per unit of time [14]. However, it is also important that the employer consider the types of exertions to ensure that the same tendons are not stressed repetitively. Since the employers are focused on work quantities, the number of exertions per unit of time to each tendon becomes a function of this quantity. Silverstein defines the levels of repetition, 1986, to be:

- a highly repetitive job is defined as one with a cycle time of less than 30 seconds or one that occupies greater than 50% of the cycle time performing the same fundamental cycles
- a low repetitive job is one where the cycle time is more than 30 seconds and less than 50% of the cycle time performing the same fundamental cycles [15].

When the physical properties of the tendons are considered, it becomes apparent that repetition of tasks is a concern. A tendon is a viscoelastic material that exhibits a time-dependent stress-strain relationship shown in Figure 1-1 [16]. Therefore, when a stress is applied to the tendon, the strain will increase when applied over time (creep) and when the stress is removed, the tendon returns to zero strain only after a finite length of time (relaxation). In experiments, tendon strain was found to be reversible up to an approximate limit of 4% strain when the tendon is not repeatedly stressed [17]. When the tendon is repeatedly stressed however, the strain is only reversible in a range of 2% to 4% strain [18]. This result shows that repetition alters the mechanics of the tendon’s collagen. Tendons are able to resist stress up to a critical strain without causing micro trauma [16]. Since micro trauma can be easily recovered from when given
sufficient time to heal, exceeding the critical strain temporarily is acceptable. However, given an insufficient recovery time, the worker is at high risk of cumulative trauma [15].

![Stress-Strain Relationship over Time](image)

**Figure 1-1: Stress-Strain Relationship over Time [16]**

1.1.2 **Awkward postures**

The properties of tendons play an important role in the analysis of awkward postures as a cause of WMSDs. It has been found that as the wrist deviates from the normal position, the tendons also become displaced from their normal position as well. This displacement is responsible for the tendons interacting with the carpal bones during extension and the transverse carpal ligament during flexion [8].

The contact between the tendons and the other tissues is responsible for slightly higher rates of WMSDs. For jobs with higher incidences of carpal tunnel syndrome, it was found that there was greater ulnar deviation and pinching required (Figure 1-2); however, these differences were not significant [19]. Nevertheless, in another study, WMSDs were shown to correlate with jobs that require great force and demand awkward postures [14]. Finally, Malchaire found that the most significant correlations with injury were the mean relative EMG value and the duration for which the velocity in flexion-extension is above 50 degrees/s [12]. Furthermore, he found that, in addition to force, the second independent factor was the mean relative angle in radial or ulnar deviation [12].
Of particular interest is a predictive model based on tendon geometry and properties constructed by Miller [16]. This model predicted that for the same angular deviation, a greater probability of failure exists for wrist extension than for wrist flexion. Moreover, the expected values of tendon stress increased linearly for increasing grasp forces; however, the stress increased dramatically for increasing deviations of wrist angles for both flexion and extension [16]. Finally, the model predicted that pinch grips are riskier than grasp grips since in a pinch grasp the entire load is carried by a single finger and thumb.

**1.1.3 Force**

Although repetition and awkward postures play an important role in WMSDs, in most of the studies conducted, repetition and awkward postures were closely associated with force and in some studies, force was found to predominate over repetitiveness. The mean force was strongly correlated to the parameters of repetitiveness and velocity [12]. For example, when Silverstein considered force as the only exposure measure in her model, the odds ratio was 4.4 (p < .0001); however, when only repetitiveness was considered, the odds ratio was 2.8 (p < .005) [15]. Moreover, in a later study it was found that the odds of having a hand-wrist CTD was approximately ten times greater in a high force, low repetitive jobs and nine times greater in the low force, high repetitive jobs when compared with low force, low repetitive jobs [5]. It is possible that the threshold between the low force and high force was set too high, which resulted in the higher incidence rates for the low force, high repetitive jobs. Therefore, the main factor on
which to act in order to reduce the risk of wrist disorders is the force exerted by the hand [12]. Furthermore, epidemiological data showed that the risk of hand tendonitis for highly repetitive and forceful jobs was 29.4 times greater than in jobs with high repetitiveness and low force requirements [14].

Studies indicate that the combination of high force and high repetitiveness contained much more risk than when either was considered alone. Silverstein's model exhibited a substantial increase in the incidence of CTDs than when each was considered singularly [15] exhibiting a risk of more than 5 times that of either factor alone [19].

The above review clearly indicates that force and repetition play a role in the cause of musculoskeletal disorders. However, methods to measure the force vary depending on the study. For example, Silverstein uses a measurement of the average force exerted by the worker, measured in “kg of force” \(^1\) [5, 15, 19], to characterize different types of jobs. Her studies then used an adjusted force (variance/mean force + mean force) to account for jobs with long cycle times and periodic high peak forces. This adjusted force formula was chosen to provide the greatest numerical difference between her low-force, high repetitive jobs and the high force, low repetitive jobs [19]. Another method used to estimate the force is to actually measure the grip force of the worker measured in Newtons [20]. Finally, the most consistent and perhaps reliable measure is to determine risk of musculoskeletal disorders is the force exerted by the wrist and the hand as estimated by EMG recordings [12].

A high force job is defined by Silverstein as one with an estimated average hand force of more than 39.2 N and a low force job is defined as one with an estimated average hand force below 9.8 N [5, 15, 19]. Her studies also state that an adjusted force (either right or left hand) greater than 58.86 N was designated “high force”. Mean adjusted forces of 31.4 ± 15.7 N and 142 ± 60.8 N were found for the low force jobs and the high force jobs respectively \(^2\) [19]. Similar to Silverstein, Armstrong states that high-force jobs are those with an estimated average hand force of more than 40 N and low-force jobs less than 10 N [14]. Alternatively, a maximum grip force of 90 N based on the 95\(^{th}\) percentile value is suggested as a high force for jobs [20].

---

\(^1\) In Silverstein's work, “kg of force” is more accurately specified as Newtons (force) divided by gravity, (i.e. weight / 9.81). Herein, all further “kg of force” units will be converted to Newtons.

\(^2\) Several examples of the types of jobs where high force was identified were: core press, adjusted force of 72.6 N, wax injection, adjusted force of 93.2 N, and cut off, adjusted force of 240.3 N [5].
Although the force measurement at the hand is important, the amount that the tendon is actually stressed cannot clearly be determined from this measurement. Physiological variations from person to person result in different stress levels in each person’s tendon for the same grip force. The measurement of percent maximal voluntary contraction (MVC) provides a more reliable measurement of tendon stress. It has also been concluded by many researchers that the mean acceptable contraction intensity for work over extended periods is 15% of MVC [21-24]. In fact, it has been shown that the predicted prevalence of wrist disorders increased by 18.7% for an increase of the mean relative EMG amplitude by 10% [25].

The threshold of 15% of MVC is of interest since, for relatively simple tasks, people can readily exceed this value. Strasser showed that when moving a mass of 1 kg, the electromyographic activity values reached values of up to 30% of MVC [4]. During his experiment, he asked supermarket employees to move a 1-kg object in a plane while measuring the EMG from various muscles involved in the motion. It was found that several muscles, including the flexor digitorum, approached or exceeded the 15% MVC threshold. Therefore, it is proposed that reducing the muscular activity to below the 15% MVC during grasping tasks could be beneficial in reducing repetitive strain injury (RSI) in workers performing handling and manipulating tasks. A method of reducing cumulative strains is to minimize grasp forces in tasks [26].

Another example of a seemingly low force job where WMSDs occur is in pipetting tasks. In a Swedish study, it was found that female subjects who pipette have hand-ailment incidence rates of 44% in their dominant hand, whereas the incidence in other female employees was 24% [27]. In this task, it was found that transferring the fluid required a force of 4.2 N. During the task, EMG was taken and found that significantly higher EMG on the extensor pollicis brevis (23.6% MVC) and the extensor digitorum (24.5%) were present [27]. When three different types of pipetting devices were analyzed, none of them exhibited a percentage MVC of less than 19.1% for the extensor digitorum [27].

### 1.2 Problem Formulation

From the evidence presented in Section 1.1, it is clear that the use of over 15% MVC can increase the risk of a worker developing musculoskeletal disorders. The crux of the problem is how to reduce the force required and subsequently the %MVC required to accomplish the worker’s task. It is hypothesized that repetitive strain injury risk can be decreased with a force
supplementing orthosis actuated by an air muscle and activated by EMG in an example power-grasping task. A power-grasping task was selected because of its relative simplicity of motion as well as its usefulness in the workplace.

Several problems exist in designing the actuated orthosis and controller. The first problem is providing a safe and compliant method of actuation. Compliance can be supplied to the orthosis either mechanically or through closed loop control. The main problem with supplying compliance through a control method is the typically higher processor requirements to maintain appropriate compliance. Although cost of the higher processor requirements may not be a factor, the higher power requirements of faster processors could make the orthosis hot and uncomfortable, more costly to run, and restrict movement to a workstation. For the orthosis to be useful, the power requirements and cost requirements should be minimized. Herein, a McKibben air muscle actuator was proposed to introduce mechanical compliance into the system. The air muscle is light, safe, and inexpensive [28] and commercially available; furthermore, the air muscle has the additional advantage of having a force response similar to biological muscle [29]. In this context, this work will study the effectiveness of a McKibben air muscle actuator as a compliant actuator for orthotic force reduction.

The second problem considered in the context of assistive orthotics is how to share stable control between the actuated orthosis and the person. Problems can occur in the stability of the control if the delay between the person’s response and the actuator’s response is too large. This situation results in the person becoming frustrated at their perceived inability to control the device. To allow for a minimal delay, a surface-mounted EMG sensor on the flexor digitorum is proposed to provide a signal for the controller, which regulates the force output by the air muscle. The user interface to the system should be as simple to the person as moving his or her own hand. Thus, as part of this work, the control loop of person – EMG – Controller – McKibben Actuator is studied in the context of an assistive orthotic.

The overall goal of this work is concerned with the potential of using an air muscle actuated orthosis controlled by EMG to reliably supplement the grasping forces of the hand. Thus, the main objective of this work is to gauge the feasibility of using an actuated orthosis to reduce the %MVC for a power-grasping task. A secondary objective is to observe the persons’ %MVC response to the force supplementation process. Finally, a better understanding of the person-device control system in a force supplementation orthotic is a tertiary objective.
2 Review of Existing Technology

Powered assistive technology has been developed over the last 100 years as a remedy for paralysis and amputation. This technology has focused on providing function to an otherwise nonfunctional appendage. Recently, other researchers have applied this technology for force amplification where much greater forces than the person is capable of are generated, such as in military applications [30]. In this work, we consider the issues surrounding a force supplementation device that would share the load with the person. In fact, the device would use the person’s biologically generated force as a controlling parameter for the force generated by the device. In the following section, a review of existing power orthotic technology, specific to the example-grasping task, is provided. This review is then followed by a review of the air muscle technology proposed as the candidate actuator. Finally, the section is completed with a review of EMG processing and its biomechanical use.

2.1 Gripping Orthosis

An orthosis is any medical device (such as a splint) applied to, or around, a bodily segment in the case of physical impairment or disability. This definition can be extended to include functional supports that assist or supplement a person’s movement [31]. The purposes of an orthosis are to substitute for absent motor power, to assist weak segments, to support segments, to provide traction, to enforce specific directional control, or for the attachment of devices [32]. For the purpose of this document, only orthotics that apply to grasping will be discussed.

Orthoses are classified into two main categories: static and dynamic. Static splints support joints when only a few muscles function inadequately, to prevent deformity, or to immobilize. Static
splints cannot generate dynamic forces, limit the range of motion, and can lead to muscle degeneration if overused [33, 34]. In the following, only dynamic splints are considered.

Dynamic splints support one limb while allowing for voluntary motion of another limb provide motive force for motions lost because of paralysis, and supply a force to help correct a deformity [34]. Dynamic splints can use elastics, electric, transmitted power, or pneumatic actuators to store, release, or inject power into the system [32]. Elastics can include springs, elastic bands, or flexible plastics. This type of device is typically used for aiding motion in one direction; however, it has the problem of inhibiting motion in the opposite direction. Electric powered systems use battery powered (usually from an electric wheelchair) DC motors and have the advantage of being easily controlled; nevertheless, it has the disadvantage of having a low power-to-weight ratio, which inhibits mobility. Transmitted power refers to the process of transmitting power from a non-paralyzed area to a paralyzed area of the body [32]. The main problem with this arrangement is that it is difficult for the person to adjust one part of the body to control a different part of the body. Finally, pneumatic power uses compressed air to actuate pistons, or artificial muscles. A common attribute of most dynamic splints is that they shift the control, strength, or stability from one part of the person’s body to another resulting in an awkward posture and discomfort [35]. Examples of dynamic splints are shown in Figure 2-1.
Shape memory alloy actuators use wires, or thin metal strips or coils that, when heated, transition to an alternate state with a different crystal structure. When electrical current is passed through an SMA wire, it becomes heated and shrinks in length as it transforms from Martensite to Austenite. As the wire cools, it will return to the longer Martensic state and a restoring force is produced. One of the main problems with SMA actuators is the electric current that passes through the actuator. The current is drawn from batteries that are very heavy and relatively inefficient [35]. The heavy batteries significantly increase the weight (the relationship between battery weight and power is constant [35]), counteracting the advantages of the lightweight SMA actuator. Furthermore, if the SMA wire is overheated, the SMA actuator will fail. Since the heating time is proportional to the diameter, using thinner SMA strands would result in faster opening and closing times; however, the likelihood of the actuator breaking is higher [45]. The SMA wire must be sized appropriately to provide the necessary forces to grasp an object while allowing for a fast response time. The power consumption by these actuators also makes them prohibitive, current ones use a rotary ratchet system to minimize power consumption as shown in Figure 2-3 [45].

![Figure 2-3: Shape Memory Alloy Actuated Hand Orthosis [45]](image)

The McKibben artificial muscle actuator consists of a rubber inner tube covered with a shell braided according to helical weaving to provide smooth motion [47, 48]. The muscle inflates and contracts to about 30% of its original length when the rubber tube was pressurized [49]. These actuators were originally designed to actuate pneumatic arm orthotics as shown in Figure 2-4 and
were successful until the mid 1960s. Once electric wheelchairs, nickel cadmium batteries, and small solid-state motors were invented, the majority of the people using these orthoses switched to battery-powered orthoses since the wheelchairs were equipped with a "whole barrel of power" [28, 40, 50-53]. It was decided that it was inefficient to carry two different energy sources and a simplification would result if the pneumatic source could be eliminated [42, 50]. In addition, electric motors had the additional advantage of being easier to control.

Figure 2-4: Artificial Muscle Actuator Orthosis [54]

Although it is difficult to achieve smooth movement with externally powered orthotics, electric systems had the advantages of more reliability, better response to linear control, joints are bi-directional driven, and electrical energy is readily available [42]. However, pneumatic systems had the advantage of being safe when pressures of 50 to 90 pounds per square inch (PSI) were used [42], high pressure compressed gas is light, the system has a very high power-to-weight ratio, and it has a fast response [35, 41]. Nevertheless, the system has the disadvantages of requiring an airtight system, a supply of pressurized air, and depends on the rubber bladder and braiding characteristics [48].
2.2 Air Muscle Technology

Artificial Muscles\(^3\), like the one shown in Figure 2-5, were patented by Gaylord in 1958; however, they did not become popular until applied by McKibben and the Rancho Los Amigos hospital to orthotics for patients in the 1960’s [28, 52, 53, 55-58]. An air muscle is composed of a longitudinal gas-tight inflatable bladder covered by a hollow double helical braid as shown in Figure 2-6. The muscle contracts lengthwise with expanded radially using compressed gas [27, 53, 56-64]. Although high tensile forces are generated by this contraction, the actuator is very compliant [53, 65]. The main drawback is the actuator is highly nonlinear, exhibits significant hysteresis, and some delay, which limits its ability to make precision tasks [53, 64, 65].

Figure 2-5: Photo of Air Muscle [66]

As mentioned above, air muscles were abandoned by orthotic practitioners when electrically powered wheelchairs became available. Recent renewed interest is in part a result of collaboration, based on the work by Uno & Sakaguchi, between Bridgestone Corporation and Hitachi, culminating in the release of their ‘Rubbertuator’ [53, 57, 67]. This design was not commercially exploited by Bridgestone [53]; nevertheless, it has led the way for other new designs.

The two-layered construction of an air muscle actuator is shown in Figure 2-6. When inflated, the thin inner rubber tube acts as a pressurized containment vessel. The wall thickness of the rubber tubing is determined by minimizing the viscoelastic dissipation and the maximal allowable air pressure supplied to the muscle. The rubber diameter should be greater than the at
Although powered orthotics are useful, these orthoses have to withstand the three classic problems of mobile robotics: weight, power, and endurance [35]. When the wearer’s concerns are considered, additional problems are raised. Practical evidence has shown that if the person does not find the orthotic attractive or useful, then it will not be used [39]. The typical user desires a sturdy, comfortable, reliable, easily removed (in less than 5 minutes), and easily controllable orthotic [40, 41]. A grasping orthotic must be able to accommodate natural motions of the hand without stressing the joints [41, 42]. Furthermore, since people have different hand sizes, a universal device cannot be constructed [36]. To accomplish all these requirements is not a simple task. In general, there has been a lack of progress in rehabilitation robotics due to high cost, poor interfaces, and the social stigma around disabilities [43, 44].

Gripping orthotics use an actuator to increase or create force for handgrip. These orthotics are applied to people with full or partial paralysis of the hand. They are activated using a push button (or mechanical system from other working muscles), puff switches, tongue switches, or EMG signals from non-paralyzed muscles as shown in Figure 2-2 [42]. The methods of actuation involve Bowden cables, pneumatic cylinders, electric motors, and linear ratchets. These actuators tend to have the disadvantages of being bulky and unsightly [45]. Recent methods of actuation include the use of shape memory alloys (SMA) and McKibben artificial (air) muscles. These actuators provide a lightweight, comfortable, inexpensive orthotic while still generating adequate forces [45]. In the near future, actuators based on Electrochemically-driven conducting polymers and carbon nanotube sheets may provide even better properties [46] as strain and strain rates of these materials improve; however, they are currently not commercially available.
rest diameter of the flexible wall to prevent energy losses due to the initial expansion of the rubber liner. This inner layer presses against a sleeve made of a strong flexible inelastic double helix cordage braid\(^4\) that forms a network of pantographs. The flexibility of the braiding is necessary to allow it to be stretched or compressed without damage while preventing the rubber liner from over inflating and rupturing. One end of the inner tube ends is sealed using a termination plug; the other end is sealed to an air inlet/outlet. Both ends of the muscle are fixed to the linkages by fixtures across the joint to be actuated. This configuration allows the air muscle to have nonlinear passive elasticity, physical flexibility, lightweight, and variable-stiffness spring-like characteristics. Details of air muscle design can be found in [28, 57, 60, 62, 63, 68-72].

![Figure 2-6: Construction of an Air Muscle [57]](image)

While the air muscle design provides an inexpensive actuator with a high power-to-weight ratio and other advantages listed above, it also has some disadvantages. As mentioned previously, the air muscle exhibits hysteresis due to the braiding contacting the rubber liner and binding. The hysteresis has been estimated by Chou [73] as a constant loss term of ±2.5 N where it is positive for shortening and negative for lengthening [70]. Although this loss could be reduced or eliminated using pleated pneumatic muscles, their bulkiness provides significant challenges for their use in orthoses [74]. Another disadvantage is the air muscle is quite slow when compared to servomotor actuators; however, this disadvantage is not a concern when using these actuators.

\(^3\) In addition, artificial muscles are known as McKibben muscles, air muscles, rubbertuators, flexator, ROMAC (RObotic Muscle ACltuator), and pneumatic muscle actuators. For the purposes of this document, the muscle will be referred to as an artificial or air muscle.

\(^4\) The braiding is typically made of nylon. Steel and polyester are also used.
with people [65]. Finally, concern exists for the miniaturization of the artificial muscle because its contraction force is proportional to its radius squared [27].

Functionally, the air muscle is a power-conversion mechanism that converts pneumatic energy into mechanical energy [60]. Any change from the rest state of the muscle will cause an increase in pressure. This pressure increase induces a force to return the muscle to its minimum energy state [28]. Consequently, the air muscle always attempts to minimize pressure within the muscle, and rising pressure inside the muscle causes the radial forces from the expanding inner tube to be translated into axial contraction forces along the fiber mesh [27, 60, 62, 68, 69, 72]. The main parameters that limit the elongation and compression are the width and length of the nylon fibers, and the pressure within the rubber tubing. The range that the air muscle can contract is between 15% and 75% of the relative unstrained dimension [57]. The amount that the muscle can contract depends on the initial mesh angle compared to the equilibrium angle. The mesh angle can be seen in Figure 2-7. It has been shown experimentally and theoretically that an equilibrium angle, where energy is at its minimum and volume is at its maximum, is 54.6° [57]. However, when the angle is too small, mesh stability problems occur. Mesh instability results in a bulge in the rubber bladder causing it to rupture [63].

![Figure 2-7: Air muscle mesh angle [60]](image)

The power conversion translation mechanism for the air muscle is similar to biological muscle’s converting chemical energy into mechanical energy where the output force is a function of length, velocity, and level of activation [29]. As a result, the air muscle has gross force and length properties that are similar to biological muscle shown in Figure 2-8 [29, 75]. Moreover, air muscles are similar to biological muscle by allowing control of joint passive tunable stiffness through the changing of the pressure [29, 52, 55, 58, 69]. An important difference between air
2.2 - Review of Existing Technology - Air Muscle Technology

muscles and biological muscles is that biological muscle can be stretched far beyond its in-vivo resting length [29].

![Figure 2-8: Force versus length relationship comparison for different types of muscles [29]](image)

The pressure range for the air muscle ranges from 200 kPa to 750 kPa; however, 200 kPa to 450 kPa is the usual operating pressure [28, 57, 68, 76]. The pressure in the muscles is regulated using electrically driven, low-power piezoelectric valves [68]. Since these are on/off valves, pulse width modulation (PWM) must be used with a switching frequency from 40 to 60 Hz depending on the type of piezoelectric valve [28, 68]. The rapid response of the valves, with a period of approximately 100 ms, makes it possible for computer control of the system. The efficiency rating can be anywhere from 25% to 94% of theoretical values depending on the characteristics of the air muscle [28, 57, 69]. The efficiency is decreased due to: line leakage, friction, rubber elasticity, expansion energy for the liner, valve losses, and dead spaces. The power output is 1-1.5 kW/kg [28, 57, 77]. The maximum power is limited by the maximum pressure of the supply, and the valve and gas characteristics. To minimize the system deficiencies, the piping length from the valves to the air muscles must be kept to a minimum and a lubricant can be used to minimize the frictional effects [28, 68]. The highest efficiency that has been reported is 94% using a polyester braiding and a latex tube [69].

The desirable properties for a powered orthotic are: high power density, high power-to-weight ratio, rapid response, accurate repeatable control, cleanliness, high efficiency, and low cost [52, 57, 70]. Two of the main hindrances to the development of powered orthotics are safety and
mobility. Currently, actuators such as those used in industrial robots are inherently unsafe for interactive use with humans because they are not compliant, they can become unstable if severe disturbances occur due to system nonlinearity, and they have powerful actuators capable of harming people [52, 78]. The actuators spend a major portion of their power to overcome gravity and as a result, these actuators are heavy, overpowered, and inefficient [69, 70]. Furthermore, industrial actuators often have controllers that produce jerky movements, which people tend to find uncomfortable [78]. A summary of the advantages and disadvantages are given in Table 2-1 [52, 62].

Table 2-1: Comparison of Different Actuator Types [28]

<table>
<thead>
<tr>
<th>Actuator</th>
<th>Advantages</th>
<th>Disadvantages</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pneumatic</td>
<td>Cheap, quick response time, simple ‘bang bang’ control</td>
<td>Position control difficult, fluid compressible, noisy</td>
</tr>
<tr>
<td>Hydraulic</td>
<td>High power/weight ratio, Low backlash, very strong, direct drive possible</td>
<td>Less reliable, expense servo-control complex, noisy</td>
</tr>
<tr>
<td>Electric</td>
<td>Accurate position and velocity control, quiet, relatively cheap</td>
<td>Low power and torque/weight ratio, possible sparking</td>
</tr>
</tbody>
</table>

Kiessling stated that the essential requirements for any external power source are: universal availability, low cost, non-toxic, safe in use, ease in handling, portable, and high power to weight ratio [52, 72, 79]. Pneumatic systems satisfy a significant amount of these requirements being lightweight and self-contained [80]. Pneumatic cylinders are cheap, reliable, quick action, safe, high power to weight ratio. Furthermore, pneumatic canisters are readily available, non-toxic, easily handled if below 400 kPa gauge [72]. The two major disadvantages are poor positional and velocity control due to the high compliance, which means that variations in loading effect the position due to the compressibility of air [28, 52, 57, 68, 70]. Partial solutions have been attempted such as precisely controlling the venting of the cylinders [57]. Air muscles attempt to correct these disadvantages by providing a larger drive area [28]. Any pneumatic system’s driving force is determined by the pressure difference and the area over which a distortion pressure is applied [70] according to the equation:

\[
dW_m = \int_L (P - P_0) dl_i \cdot ds_i = (P - P_0) \int_{s_i} dl_i \cdot ds_i = P'dV
\]  

(1) [71]

5 Typically DC or AC motors and hydraulics [28]
where $p$ is the absolute internal gas pressure, $P_0$ is the environment pressure, $P'$ is the relative pressure, $S_i$ is the total inner surface, $ds_i$ is the area vector, $dl_i$ is the displacement, and $dV$ is the volume change [71]. From this equation, using virtual work, the output force is given by

$$F = P' \frac{dV}{dL}$$

Since the drive area for the air muscle is the entire surface area of the inner tube whereas in a conventional cylinder it is only the piston head [70], the air muscle is capable of producing more work.

Air muscles have the same advantages as pneumatic cylinders even though they are more difficult to control [53, 57, 64, 81]. Their high power-to-weight ratio allows the actuators to be significantly smaller and lighter while providing comparable forces, resulting in compactness and energy savings [58, 78, 82]. An air muscle weighing less than ½ kg can lift over 1,100 N at more than 20% strain [58]. Moreover, the muscle has force-limiting output, is physically flexible, and exhibits spring-like characteristics [27, 55, 82].

Natural compliance can be produced using either flexible links or joint compliance [67]. Flexible links exhibit approximately a constant compliance. The compliance of an air muscle actuated arm can be varied by two antagonistic muscles as shown in Figure 2-9 [61]. By increasing the pressure in these muscles, while maintaining the pressure ratio, the compliance will decrease and the angular position will remain constant [53]. Passive joint stiffness provides energy efficiency [55]. Colbrunn has produced a leg capable of walking with the control valve off 90% of the time [55]. However, a pressure difference between the two antagonistic muscles produces asymmetric stiffness. Mills proposed using DC servomotors to maintain tension in the cables [61, 83]. With this method, the stiffness can be maintained equal between the two actuators [61].

![Figure 2-9: Two air muscle configurations: Dual Air Muscle Configuration (left); Single Air Muscle and Spring Configuration (right)](image-url)
2.2 - Review of Existing Technology - Air Muscle Technology

Variable compliance is a significant advantage of the actuator in terms of safety when encountering people, such as in rehabilitation robotics and orthotics, while still allowing for high stiffness applications [65, 72, 82]. If a low maximum joint stiffness of an air muscle actuated arm is designed, then the arm will remain stable in every position in the workspace, even when a disturbance occurs from system nonlinearities such as saturation of inputs or input dependent delays [78]. A low joint stiffness will offer no resistance when a person touches it, or when it touches a person [78]. Moreover, the frequency spectrum of an actuated arm exhibits no inherent high frequency oscillations [78].

2.2.1 Air Muscle Modeling

The air muscle is frequently compared to biological muscle due to the similarity between their force-length properties [29]. Nevertheless, the air muscle has very little damping and as a result, it has a different force-velocity profile than biological muscle. Moreover, the dominant Coulomb friction causes a velocity-independent hysteresis in the muscle since the velocity-dependent viscous friction is immeasurable [71]. It has been suggested by Klute et al. [29] that adding a damper could possibly create an air muscle with more biological muscle-like force-velocity profile. However, when adding a damper in parallel, this configuration requires that the orifice diameter to change instantaneously due to the conservation of energy and using Bernoulli’s equation [29]. Chou and Hannaford suggests lubricating the muscle to reduce the Coulomb friction and adding viscous material in order to fit the biological tension-velocity relationship more accurately [71].

To design a control system for the air muscle, a sufficient model of its performance is necessary [52]. Complex models exist for the air muscle that includes elastic effects, end effects (shown in Figure 2-10), radial expansion forces, and friction [52] (predominantly Coulomb friction [71]); however, these models typically require the measurement of the braid angle of the muscle. This measurement requirement is unrealistic in most applications. Moreover, the equations are computationally complex when all of the effects are considered making it difficult to provide a quick response time when implemented in a controller.
Most models simplify the differential equation by using the lumped-parameter models of the muscle that are based on the physical characteristics of the air muscle [27, 71]. The differential equation is given as:

\[ F = P \frac{dV}{dL} - V_b \frac{dW}{dL} \]  

(3) [71]

where \( P \) is the input actuation pressure, \( dV \) is the change in the actuator's interior volume, \( dL \) is the change in the actuator's length, \( V_b \) is the volume occupied by the bladder, and \( dW \) is the change in stored energy on a per volume basis [56]. The non-cylindrical end effects of the air muscle are ignored, and the mesh material is assumed as inextensible [66]. The majority of the models begin with the assumption that the air muscle is a cylinder with a wall thickness of zero, resulting in the second term on the right hand side of equation (3) to become zero [59, 71, 84], and dimensions of length, L, and diameter, D [29, 55, 58, 66, 71]. These simplifications are necessary to provide a suitable equation for a control; however, they do result in some inaccuracies in the model.

In this work, the air muscle controller will use the model devised by Colbrunn [66]. In Colbrunn’s model, the force and translation of the air muscle depend on four main parameters:

1. the weave characteristics of the braided sleeve,
2. the material properties of the elastic tube,
3. the actuation pressure,
4. and the muscle’s length [27].

Nevertheless, the parameters of length and diameter are not constant and therefore, these parameters must be expressed in terms of two constant parameters. The length and diameter are
related to the constant parameters of the thread length, \( b \), and the number of turns for a single thread, \( n \), [66] as shown in Figure 2-11. It is the interweave angle, \( \theta \), that determines the instantaneous length and the diameter of the cylinder. The interweave angle, \( \theta \), is defined as the angle between the thread and the long axis of the cylinder [58, 66].

![Figure 2-11: Geometric explanation of the three derived characteristic parameters (thread length, number of turns on each thread, and interweave angle) of the air muscle [66]](image)

Although the three parameters determine the characteristics of the air muscle, they are difficult to measure, especially the interweave angle since it must be measured during muscle operation [58, 66]. Caldwell, Medrano-Cerda, and Goodwin solved the problem by determining that the constant parameters for the thread length and number of turns on a single thread are:

\[
b = 2A\ell \tag{4}
\]

\[
n = \frac{A}{B} \tag{5}
\]

where \( A \) is the number of lengthwise trapezoids, \( B \) is the number of circumferential trapezoids, and \( \ell \) is length of a trapezoid side [66, 68]. The trapezoid they define is the trapezoids formed by the mesh when it is opened and laid flat such as in Figure 2-12.
The final equations for volume and force are then given as:

\[ V = \frac{L(b^2 - L^2)}{4\pi n^2} \]  

(6)

\[ F = \frac{P_g b^2}{4\pi n^2} \left( \frac{3L^2}{b^2} - 1 \right) \]  

(7)

where, \( P_g \) is the gauge pressure, and \( L \) is the muscle length [66]. The equation for stiffness, \( k \), is derived from the force equation as:

\[ k = \frac{dF}{dL} = \left( \frac{b^2}{4\pi n^2} \right) \frac{dP_g}{dL} + \frac{3P_g L}{2\pi n^2} \]  

(8)

and since \( \frac{dP_g}{dL} \) is difficult to formulate, and the pressure change as a function of length is small, it is neglected giving:

\[ k = \frac{3P_g L}{2\pi n^2} = \frac{6F}{3L - \frac{b^2}{L}} \]  

(9)

A detailed derivation and verification of the accuracy of these equations is given in [66]. Other researchers have derived variations of this basic equation [56, 67, 69, 78, 85]. Colbrunn further refined the equation by including an effectiveness term:
2.2 - Review of Existing Technology - Air Muscle Technology

\[ F = \begin{cases} \frac{P_b}{4\pi n^2} \left( \frac{3L^2}{b^2} - 1 \right) \cdot \text{Eff}(P_g) + F_{\text{max,lin}} & \text{if } (L > L_{\text{min}}) \\ 0 & \text{if } (L < L_{\text{min}}) \end{cases} \]  

(10)

where:

\[ F_{\text{max,lin}} = \begin{cases} K_{\text{braid}} (L - L_{\text{max}}) & \text{if } (L > L_{\text{max}}) \\ 0 & \text{if } (L < L_{\text{max}}) \end{cases} \]  

(11)

Here, \( K_{\text{braid}} \) is the braid material stiffness, \((L_{\text{min}}, L_{\text{max}})\) are the minimum and maximum lengths of the air muscle, and \( \text{Eff}(P_g) \) is an “effectiveness term” that is determined empirically based on the theoretical and actual force at each nominal pressure [58, 66]. It is given by the equation:

\[ F_{\text{measured}} = \text{Eff}(P_g) \cdot F_{\text{theoretical}} \]  

(12)  [58, 66]

The effectiveness term is necessary to compensate for error of up to 20% when the muscle is stretched to higher extensions resulting in a force that is overestimated [52]. These errors are a result of energy losses due to line leakage, valve losses, dead spaces, rubber elasticity, expansion energy for the inner tubing, friction between the bladder and the braid, and the friction between braiding threads [56, 58, 70]. All air muscle systems exhibit some, if not all, of these properties. If the errors were modeled, the damping would increase and the mass flow rate into the actuator would decrease [58]. These nonlinear factors are all functions of the braiding, bladder, pressure, and actuator length [56]. Nevertheless, the errors are more likely near the extreme lengths where the system stiffness changes drastically. Most applications of air muscle tend to avoid these regions of operation where, according to [71], the tension reaches zero when the length is shorter than \( 0.75L_0 \) and becomes very stiff when the length is longer than \( 1.1L_0 \) (where \( L_0 \) is the muscle rest length) [71]. Furthermore, errors are caused by the end effects, which violate the theoretical cylindrical assumption. It is suggested by Klute that the length to diameter ratio for the actuator should be at least 14 for the assumption to maintain reasonable accuracy [56]. As opposed to the aforementioned errors that increase at the muscle length extremes, hysteresis is an error that is exhibited over the entire range of the air muscle and decreases toward the extreme lengths as shown in Figure 2-13.
2.3 Electromyogram Signals and Control

The human muscle movement process is complex. In this thesis, only the information necessary to this thesis will be presented. For information on how the brain produces a motor command or on the transmission of a motor command to the muscle the reader is referred to Basmajian and De Luca [86].

Once the motor command issued instructs a muscle to contract, the nerve cells associated with that muscle release a neurotransmitter into a small gap (synapse) between the nerve cell and the muscle membrane. The neurotransmitter crosses the gap and binds to a protein receptor on the muscle membrane resulting in an action potential in the muscle cell. This action potential causes the muscle to contract. Since muscles are made of many muscle cells, there are tens of thousands of muscle cells (depending on the muscle size) activating randomly during muscle activation [86].

An EMG signal is the measurement of these action potentials activating during the muscle activation. It is measured either internally through a needle, intramuscular, or externally through a surface mounted sensor [87]. The EMG signal is a stochastic signal reasonable represented by a Gaussian distribution function [88]. For a surface EMG sensor, the signal levels are in the range of microvolts and require heavy amplification. The frequency range for such a signal is in the
range of 5 to 500 Hz with the dominant energy being in the 10 to 150 Hz range [88]. The preferred filter range for surface EMG recordings should be 10 to 350 Hz [87]. For intramuscular recording, the range is from 10 to 450 Hz [86]. Finally, needle recording should have a bandwidth of 10 to 1500 Hz. This work uses surface EMG.

EMG has been used in a wide variety of areas; however, research has focused on several interrelated areas:

1. Clinical use for diagnosis of neurological disorders
2. Feature identification, decomposition, and pattern recognition
3. Biomechanics
4. Prosthetic/orthotic control

The primary focus will be placed on biomechanics and with secondary focus on prosthetic control. In particular, the ability to approximately determine the force of a muscle from the EMG signal is important for the proposed controller used for this work. The secondary focus on prosthetic control considers the use of EMG use as a proportional/analog controller. However, topics in which the EMG signal is used as more than one control command are not discussed herein [89]. The experiments in this thesis only use single-channel EMG control applied to able-bodied individuals. For more information on the clinical use of EMG and decomposition see Gutmann [90] and Fang [91] respectively.

2.3.1 Electromyogram Characteristics

The pseudorandom noise nature of the EMG signal makes it difficult to use as a control signal due to its erratic nature. Moreover, the small signal level is susceptible to electrical noise from the components, the environment, cross-talk from other muscles, and motion artifacts [87, 88, 92]. The electrical noise from the components can be reduced by using high quality components; however, the environmental noise presents more of a problem. The environmental noise can have an amplitude that is one to three orders of magnitude larger than the EMG signal [88]. Therefore, it is important to amplify the EMG signal as close to the source of the signal as possible to prevent the ambient noise from overwhelming the EMG signal before it is recorded. Primarily, differential amplification (Figure 2-14) is used in order to reduce common mode noise such as 50 Hz or 60 Hz power line noise [88, 92]. Motion artifact noise is created from the interface
between the sensor and the skin as well as the motion of the cable connecting the electrode to the amplifier [88].

![Figure 2-14: A schematic of the differential amplifier configuration. The EMG signal is represented by 'm' and the noise signals by 'n'. [88]](image_url)

Furthermore, the EMG signal is composed of many different factors that affect the quality of the signal received and can be broken down into three categories: causative, intermediate, and deterministic (Figure 2-15) [93]. However, only partial models of these factors have been developed and therefore most work focuses on the use of simplistic models and analysis while further research is performed on developing more complete models [93].

![Figure 2-15: Schematic diagram of the factors that affect the EMG signal [93]](image_url)
The electrodes for the differential amplifier typically have a separation of 1 cm and are approximately 1 mm in diameter; however, this configuration is not standardized and varies based on the application. The distance between the electrodes affects the bandwidth and amplitude of the EMG signal where a smaller distance shifts the bandwidth to higher frequencies and lowers the amplitude of the signal [88]. The size of the electrodes affects primarily the amplitude of the EMG signal detected with larger electrodes eliciting greater amplitudes [88]. Nevertheless, the increased size of electrodes results in a larger sensor that may not be appropriate for smaller sized muscles. Selecting the appropriate electrode size, inter-electrode distance and location of recording over the muscle must be carefully planned in areas where many narrow muscles are tightly gathered or when working with superficial/thin muscles [87]. The work of Basmajian and De Luca can be consulted for more specific recommendations [86, 87].

The placement and orientation of the electrode is critical in providing a good signal from the muscle of interest. The electrode should be placed between a motor point and the tendon insertion or between two motor points as shown in Figure 2-16. Application of electrical stimulation to the skin over the muscle may help determine the location of the innervation zones; however, this procedure is not convenient in many circumstances [88]. Therefore, it is recommended to place the electrode in the middle of the muscle between the origin and the insertion point [86, 88].
The reference electrode grounds the differential amplifier in the EMG sensor. It is suggested that the placement of this electrode should be as far as possible away from the EMG sensor on electrically neutral tissue such as a bony prominence [88]. The reference electrode is typically much larger than the EMG sensor's electrodes to allow for good electrical contact with the skin.

### 2.3.2 Electromyogram Filtering and Normalization

Once the EMG sensor and reference electrode are appropriately placed, one must consider how the EMG signal should be filtered prior to recording it with a computer. Following initial amplification, the signal is filtered with a band pass filter with typical cutoff frequencies of 5 to 500 Hz depending on the application. The lower frequency cutoff removes the gross motion of the appendage and the high frequency cutoff is used to remove the noise caused from the signal passing through the skin and other extraneous sources [86, 88]. A notch filter may be used to remove power line noise from the signal [92]; however, the use of such a filter can be inappropriate as removal of power line noise is possible with a differential amplifier and the notch filter removes significant EMG signal content. A notch filter should only be used when a differential amplifier cannot be used for the application.
Following the band pass filtering, the signal is rectified using either full wave or half wave rectification if the signal power is required [87]. Performing the rectification prior to sampling the signal allows for reduction in quantization noise as the full range of the analog-to-digital converter (ADC) can be spread over half the signal level. However, if the signal is to be used to decompose the EMG signal into individual action potentials, this process should not be performed.

Finally, if the signal envelope is required, one or more of three methods in the time domain are typically used: smooth filtering, root-mean-square (RMS), or integrated EMG. Although smooth filtering will obtain the envelope of the signal, the power level of the EMG signal will not be provided. RMS is preferred for most application in biomechanical engineering as it provides a power measurement of the signal and thus has a physical meaning. Finally, integrated EMG is unfiltered and provides an integration of the EMG signal over time.

The signal in smooth filtering is filtered in the frequency range of 0.5 to 4 Hz (or normally 50 to 250 ms time constant [87]). The smoothing filter can be performed in either hardware or software and both methods have different advantages. Filtering in hardware allows for sampling the EMG signal at a significantly lower rate resulting in inexpensive sampling hardware and less power consumption. However, filtering in hardware creates difficulties in changing future filter cutoff frequencies. On the other hand, filtering in software provides flexibility in changing the filter cutoff frequencies relatively easily. Nevertheless, it requires relatively more expensive sampling hardware (over 1 kHz sampling rate) and more power consumption. These trade-offs should be considered during the design of the proposed system. For example, if the system were to be portable, filtering in hardware would be preferred due to the low power consumption.

The RMS value for the signal is calculated by recording windows of the EMG signal, squaring the result, averaging the squared values, and finally performing the square root on the result. This operation is performed in software and therefore requires relatively more expensive sampling hardware and more power consumption. Due to the averaging, a period over which the RMS value is calculated must be provided [87, 94].

Finally, integrated EMG performs the integration of the EMG signal over time rather than simply smoothing the signal. This procedure provides the ability to observe the accumulated EMG
activity over time. However, it has the disadvantages of providing a signal that does not have a physical meaning and requires appropriate reset parameters of voltage or time for the integrator [87, 88].

Regardless of the procedure used to smooth the EMG data, the EMG data typically is normalized relative to the MVC in investigations where force or torque are correlated to the EMG [87]. Nevertheless, to obtain a true MVC from the subjects requires some preliminary training. Without training the MVC could be between 20 to 40% less than the true value [87]. However, most papers neglect presenting a procedure for eliciting the MVC. In fact, De Luca states that a consensus is required for a procedure for determining the MVC in studies [93].

2.3.3 General Biomechanical Use of the Electromyogram

The use of surface EMG in biomechanics is applied mainly to three applications [93]:

1. Indication of muscle activation initiation
2. Proportional relationship to force produced by a muscle
3. An index of fatigue processes occurring in a muscle.

The indication of muscle activation initialization is important in providing the timing sequence of one or more muscles involved in performing a task, such as during gait. To determine muscle activation initialization, more complicated measurement equipment is required due to crosstalk. During initialization, the EMG signal registered is initially low and near the noise level. Consequently, an adjacent muscle may be detected erroneously as the activation of the muscle of interest [86, 93].

To reduce false activation detection, several different sensor placement schemes can be attempted. The first and simplest method is using a double differential technique with an EMG sensor with 3 equally spaced detection surfaces. A differential signal is created from two other differential signals (surfaces 1, 2 and 2, 3) [95]. A second, more impractical technique is to measure EMG from all adjacent muscles to monitor them for lack of activity [93]. Finally, the third technique is monitoring the frequency spectra of the signal and removing lower frequency signals. Since the crosstalk is from farther away muscles, the result is a shift to the lower frequency bands as it passes through tissue [96].
A window size of 10 ms is suggested to determine the activation state of the muscle; however, the muscle activation exhibits a delay between the registered activation by the EMG sensor and the resultant force measured through a force sensor. This delay depends on several factors such as the fiber type composition of the muscle, the firing rate dynamics of the muscle, and the viscoeelastic properties of the muscle and tendon tissues [93]. Logically, slow twitch muscles have a slower rise-time and fast twitch muscles have a shorter rise-time between EMG signal and force.

Similar to determining muscle activation, determining a quantitative relationship between an EMG signal and the muscle's force (or torque at a joint) depends on the muscle being measured in combination with many other factors (see section 2.3.1). As a result, only relatively simplistic models are implemented in most of these types of studies [93]. Although only simplistic models are used, they are usually sufficient in determining the contribution of a muscle to the torque experienced at the joint.

The main advantage of using surface EMG in muscle force studies is to provide information about the force contribution of individual muscles to a motion and to develop muscle models to describe this behaviour. By measuring the EMG from all the muscles involved in an action, the individual contribution of each muscle can be monitored non-invasively. This ability is important since the moment action on a joint is rarely due to a single muscle group [86, 93]. The interosseous muscles of the hand are one instance where only one muscle acts on a joint at a time and is an additional reason why a grasp was chosen as the example task.

For indicating muscle fatigue, EMG has a significant advantage over traditional measures of fatigue. Traditionally, the fatigue point has been defined as the point at which contraction can no longer be maintained [93]. This definition considers the fatigue point as a specific point in time; however, fatigue occurs over the course of the action. Additionally, this definition has the disadvantage that it only detects fatigue once failure has occurred and does not account for multiple muscles action on a joint with the possibility of some muscles fatiguing before others [93].

The primary method suggested by De Luca is the principle of median frequency shifting, which is less sensitive to noise and signal aliasing; however, it is sensitive to the biochemical and
physiological factors that occur during sustained contractions [93, 97, 98]. During a sustained contraction, the median frequency shifts to lower frequencies due to the firing behaviour of the motor units and the change in shape of the motor units' action potentials.

2.3.4 Specific Application of EMG to Handgrip Force Estimation

A mathematical model describing the relationship between handgrip force and EMG measurements was developed by Duque et al. The technique was developed to reliably determine the handgrip force during studies of workers in the assessment of grasping tasks and closely relates to the requirements for this thesis. Since the technique was to be used in the field, it had the requirements of being simple, safe, and comfortable for the worker while maintaining accuracy. As a consequence of these requirements, surface EMG electrodes and wrist angles had to be used [99].

During the experiment, the subject was instructed to increase the force progressively over 1 second until the desired was reached and maintain it for 4 to 5 seconds before progressively releasing. This exertion was followed by a 60 second period for an effort of less than 50% MVC and at least 2 minutes for an effort of greater than 50% MVC. Electrodes were placed over the flexor carpi radialis lying superficially above the flexor digitorum superficialis. Since wrist angles affect the amount of mechanical force produced at the fingers during grasp, the wrist angles were calculated using the maximum angle values recorded at the beginning of the experiment in full voluntary flexion, extension, and deviations without exerting any force [99].

The hand force estimation model uses empirical data to produce a model of how the percent grip force varies with the %MVC and the relative wrist angles in the extension-flexion and ulnar-radial direction [99]. This empirical model was created to provide a simple method of evaluating handgrip force in the workplace. Previous methods of estimating the force were too cumbersome to use for work related tasks [99].

The empirical equation for hand force estimation based on %MVC and wrist angles proposed in [99] is:

\[
\ln(Force\%) = 1.13 + 0.714 \ln(EMG\%) + 3.19 \times 10^{-2} \text{Flex} + 0.83 \times 10^{-2} \text{Dev} - 26.9 \times 10^{-6} \text{Flex}^2 + 15.7 \times 10^{-6} \text{Dev Flex}
\]
where $ln(\text{Force\%})$ is the natural logarithm of the relative force; $ln(\text{EMG\%})$ is the natural logarithm of relative EMG; $\text{Flex}$ is the relative angle in extension-flexion (%); $\text{Dev}$ is the relative angle in the ulnar-radial deviation (%). When this equation is used, the exponential of both sides is taken to produce the percent force. Multiplying the percent force by the maximum grip force yields the estimated grip force.

Duque’s model produced extremely high correlation coefficients ($R=0.895$). Nevertheless, some flaws exist with the model. When extreme wrist angles are used, the model predicts forces greater than 100 percent. However, it was found during Duque’s experiments that at full flexion, the subjects were unable to exceed levels of force greater than approximately 50% MVC. Furthermore, in many working conditions, the flexion of the fingers is accompanied by wrist flexion. This situation may cause the wrist flexion muscles to predominate leading to a force overestimation. Duque’s paper notes that the model does not aim to depict the physiology of biomechanical behaviour of the hand-force system; it simply provides an order of magnitude estimate of the force produced [99]. Finally, the model does not predict fatigue or temperature changes in the muscles.

Many similarities exist between the force reducing orthosis system and the setup described in [99]. The EMG data was taken using surface electrodes attached to the forearm and filtered with a bandwidth of 20 Hz to 600 Hz. However, the system in [99] is used for a grasp in which an object is picked up from the side. Since the subjects in our study will lift objects from the top, the type of grasp is significantly different. Consequently, the model may not accurately represent the type of grasp used in our experiments. Nevertheless, since the wrist angles used in this thesis to be kept fixed and at a neutral position, it is expected that the different grasp position’s effects will be minimal. The primary difference expected would be due to the different contact surfaces of the fingers on the object grasped.

2.4 Summary

As can be seen from the review of orthotic technology, the design of an orthotic system is a complex task. Therefore, the decision was made to choose an example task where the design of a full orthotic system was not required. This simplification reduced the amount of mechanical design without diminishing the quality of the results. Furthermore, due to the simplicity and
relative accuracy of Colbrunn's equation, his equation was chosen to model the air muscle. Finally, the method derived by Duque was chosen to determine the force for the air muscle actuator to generate. In the following section, these elements will be combined in a simulation to aid in the development of the experimental test system.
3 Methodology and Experimental System Design and Simulation

In this chapter, the design of the experimental EMG controlled air muscle force supplementation system is developed through modeling and simulation of each system component. The design culminates in a set of simulated experiments used to predict the behaviour of the system, which guides the hardware design and the physical experimental setup as described in Chapter 4.

3.1 Overview

A key objective of the simulation is to design the experimental system controller. The simulation provides a tool to quantify the expected system response to different possible EMG inputs and different human reactions to the system. Moreover, in combination with experimental results, the simulation allows the quantification and evaluation of the human (i.e. the subject’s physiological, and neurological) response to the system.

The simulation models the physical components of the system as shown in Figure 3-1: the brain (descending command and simulated EMG generation), the periphery behaviour model (human response), and the orthotic system (actuated air muscle orthotic). The simulated EMG generation and the periphery behaviour model attempt to approximately simulate the subject’s response to the orthotic system. The simulated EMG generator produces pseudorandom noise that is modulated by an input that can vary between zero and one. The periphery behaviour model contains the empirical motoric and sensory models of the human grasping system. The motoric model simulates the response of the person’s nerves and muscles to an input signal and the sensory model simulates the response of the proprioceptors and muscle reaction to a given feedback signal.
3.2 Methodology and Experimental System Design and Simulation - Simulated EMG Generation

The orthotic system models the EMG signal processing algorithms required by the air muscle controller as well as the pneumatic muscle system. The EMG algorithm consists of filtering and rectifying the EMG signal and converting it to a %MVC. The %MVC is used in an empirical model to compute a percent muscle force. The result of this processing is a set point force for the air muscle controller. The pneumatic muscle system models the valve response, the equalizing spring, and the air muscle’s pressure, length, and force. Figure 3-1 shows an overview of the simulation model for an assisted lift. The unassisted experiment simulation is similar; however, the systems involved with the assisted lift have been removed. The removed systems are the Control Algorithm, Valve and Air Muscle Model, and the Orthosis Mechanics.

The simulation is implemented in Simulink® [100] and simultaneously simulates the unaided and aided lift experiments. The output from both simulated experiments can then be observed and compared for identical input signals into the simulation.

3.2 Simulated EMG Generation

The simulated EMG generation model is shown in Figure 3-2. The subject creates a descending command signal that is used as a feedforward signal to the flexor digitorum muscles, which flex the proximal interphalangeal joints of the fingers. This feedforward signal is compared to the sensory feedback signal and passed through a proportional-integral-derivative (PID) automatic controller, which modifies the muscle activity based on the sensory feedback. The maximum
force that the person is capable of generating scales the sensory feedback to a value between zero and one for comparison between the descending command signal and the received force.

The PID controlled error signal is added to the feedforward command (for simplicity the descending command is used) to generate the $\alpha$-signal. The $\alpha$-signal is sent to the biological muscle and to a simulated EMG generator. The simulated EMG is generated by a modulation of the $\alpha$-signal with a band-limited white noise generator with levels between 5 and $-5 \text{ V}$.

Fagergren’s method for determining the models for the motoric and reactive systems was considered in selecting the gains for the PID controllers. Since the Fagergren models assume that the voluntary activation gain is 1, the models should incorporate any significant gain factors that would be present in the PID controller [101]. Therefore, it was decided that the gain factors for this controller should only contain a proportional gain of 1. This assumption was necessary to allow for the determination of the air muscle controller’s gain factors determined in section 3.4.6.

![Activation signal and simulated EMG generator](image)

**Figure 3-2: Activation signal and simulated EMG generator**

### 3.3 Human Grip Motoric and Reactive Models

Detailed modeling of upper limb dynamics is extremely complex. For a precision grip, the model requires a complete set of 15 parameters for all muscles involved, as well as parameters for joint locations and moment arms. Such a model is overly complex for the purposes of this work. Therefore, a black box model derived by Fagergren and others is used [101]. Fagergren
identified the motoric model and the reactive model for the precision grip system. Since Fagergren's technique identifies the whole system from the brain signal to the finger's response, this technique identifies many of the same muscle groups and nerves that a power grasp would use. Moreover, the response of their systems when excited by an activation signal exhibit delays similar to those observed in practice.

Fagergren's approach involves *common subsystem identification*. This identification technique attempts to identify common poles and zeros associated with two different experiments that contain some common elements. Fagergren used an active and a reactive experiment to determine the linear transfer function of the precision grip system [101]. Although the human muscular system is nonlinear, it can often be approximated as linear if the signal is limited in range or if only a general estimate of the response is necessary. Genadry suggests that a 2nd order approximation is capable of accounting for 95% of the variance in the ankle [102].

Fagergren's active experiment involved asking the subjects to initiate a "step response" in their precision grip. The reactive experiment involved dropping a weight and having the subject grasp it as it was falling. For both these experiments, the force output was measured and a perfect step response was assumed as the input. From this, various transfer functions of different magnitudes were constructed and cross-validation is used to determine an adequate order for each system. It was found that for active data, a second- or third-order transfer function with no zeros described the data. For the reactive data, a third- or fourth-order transfer function with one zero was adequate.

After identifying the poles and zeros from these two experiments, the common and differing elements of each experiment were determined:

\[
H_{motoric}(s) = H_{active}(s) \cap H_{reactive}(s) \tag{14}
\]

\[
H_{sensory}(s) = H_{reactive}(s) \setminus H_{active}(s) \tag{15}
\]

\[
H_{voluntary}(s) = H_{active}(s) \setminus H_{reactive}(s) \tag{16}
\]

where \(\setminus\) denotes the set difference, and \(\cap\) the intersection.
The result of Fagergren’s experiment [101] yielded:

\[
H_{\text{motoric}}(s) = \frac{280}{s^2 + 22s + 280} \text{ (N)} \tag{17}
\]

\[
H_{\text{sensory}}(s) = \frac{2.8(s + 1.2)}{s + 3.4} \tag{18}
\]

\[
H_{\text{voluntary}} = 1 \tag{19}
\]

Although these transfer functions are for the precision grip, it is assumed that the transfer functions will be similar to a power grasp. The distance the activation signal has to travel for a precision grasp is the same distance for a power grasp. Furthermore, similar muscle groups are used for the other fingers during a power grasp. Therefore, these transfer functions give a good first approximation of the power grasp behaviour. Further work to more closely identify the transfer functions for a power grasp could be considered to fine tune the simulation proposed herein. However, at this investigative stage, the results provided by Fagergren are considered a suitable approximation.

3.4 Orthotic System

3.4.1 Overview

The orthotic system models the complete air muscle actuated orthotic as well as the signal conditioning required by its controller. It consists of four main components shown in Figure 3-3: (i) EMG to %MVC, (ii) %MVC thresholding, (iii) hand force estimation model, and (iv) the air muscle control loop. The “keep off” or switch block prevents EMG input into the air muscle control loop during its settling time.
3.4.2 Electromyogram Signal to Percent Maximal Voluntary Contraction

The EMG to %MVC block performs filtering and rectification on the EMG signal and performs a scaling to produce a single output of %MVC. This block has two inputs: (i) the MVC that is to be recorded at the beginning of the experiment, and (ii) the raw EMG signal from the EMG sensor placed on the subject.

The main purpose of the filtering and rectification is to estimate the output grip force from the hand. Since EMG is an amplitude modulated (AM) noise signal, it has many attributes in common with an AM radio signal, as shown in Figure 3-4. In an AM radio system, a low frequency signal such as a person's voice is used to modulate a carrier wave of a higher frequency. To receive the signal, the signal is rectified and envelope filtered to obtain the baseband signal. In an EMG system, the person generates a desired force signal and it is transferred to the muscle via motor neurons where the EMG sensor picks it up. The transfer through the motor neurons effectively modulates the desired signal to a higher frequency. To obtain the person's baseband signal, the EMG is filtered to remove noise from the signal, rectified, and low pass filtered. The main difference between the AM system and the EMG system is that the AM radio system typically has a fixed carrier frequency while the EMG system has an indeterminate carrier frequency between 20 Hz and 500 Hz.
Figure 3-4: Comparison between an AM radio signal and an EMG signal

The EMG is filtered by a 17 Hz to 500 Hz Butterworth band pass filter. From the literature, it was found that this pass band is used commonly in the practice of recording EMG [87]. The 17 Hz cutoff is to prevent movement artifacts from entering the system and the 500 Hz cutoff is to prevent high frequency noise from the environment from overpowering the EMG signal.

The frequency spectrum of both the EMG signal and the force signal were taken from some preliminary data sampled at 1000 Hz to determine a sufficient cutoff frequency for the EMG signal. As shown in Figure 3-5, the EMG signal ranges from DC to the Nyquist frequency of 500 Hz although very little is above 100 Hz whereas the force signal ranges only from DC to 5 Hz. Since only the force envelope is required from the EMG signal, a 7 Hz Butterworth low-pass filter is applied to the signal following rectification. This filter smoothes the EMG signal to provide an envelope for the signal that will be more stable for control of the air muscle. A gain of 10 is applied to scale the signal back to the input amplitude whereby compensating for the filter’s attenuation in their pass bands. It is likely that the gain for the real system will have to be adjusted depending on the implemented filters characteristics. The filtered output is then reduced
to between 0 and 100 by dividing the signal by the MVC input into the block and multiplying by a gain of 100.

Figure 3-5: Fourier Analysis of EMG and Force Spectrums

3.4.3 Thresholding Percent Maximal Voluntary Contraction

For the purposes of the implemented system, the MVC thresholding block performs no operation on its inputs. The result is that the air muscle system will match the desired force generated by the subject. In the current simulation, the air muscle is programmed to match the subject’s grasp force, thus sharing the load equally. However, this block was provided as a hook for future algorithms in adjusting the method for reducing the %MVC of the subject. In future work, it is suggested that thresholding the %MVC will be necessary to allow the air muscle to compensate for a larger portion of the %MVC and subsequently a larger portion of the force. Several methods could be attempted to compensate for larger portions of the %MVC such as simple gains, non-linear gains, neural networks, or fuzzy logic. Using a gain to scale the %MVC to a value larger than the person would cause the air muscle system to take a larger portion of the force; however, this method does not limit the person’s %MVC thereby allowing the person to still generate unsafe %MVC levels. Moreover, the person may cause the system to generate
exceptionally high forces resulting in injury and stability problems between the person and the system. It is suggested that the use of a non-linear gain would be a satisfactory solution for this problem. By providing a scaling function, the system could provide little assistance in the 0 to 15 %MVC range followed by a higher gain in the 15 to 50% range, and then scale back the gain above 50 %MVC to prevent injury and oscillatory behaviour between the subject and the system.

3.4.4 **Hand Force Estimation Model**

The hand force estimation model used in the simulation is based on the work by Duque et al [99]. The simulation implements equation (13); however, since the wrist angles were fixed at a neutral position, the equation reduces to:

\[
\ln(\text{Force}) = 1.13 + 0.714 \ln(\text{EMG})
\]  

(20)

where \(\ln(\text{Force})\) is the natural logarithm of the relative force; \(\ln(\text{EMG})\) is the natural logarithm of relative EMG.

3.4.5 **Air Muscle Model, Valves, and Control System**

The air muscle control loop consists of six main subsystems as shown in Figure 3-6: equalizing spring, pressure & length to force, force & length to mass, force & pressure to length, the valve model, and the air muscle model.

![Figure 3-6: Air muscle control loop](image-url)
Since the air muscle must be pressurized to approximately 10 PSI to maintain inflation, an equalizing spring is used to approximately balance this force. The block takes the initial, uninflated length of the air muscle and subtracts the current length of the air muscle. This distance should represent the spring extension. The force is estimated using Hookes law for a linear spring, \( F = k \cdot x \).

The pressure and length to force block shown in Figure 3-7 is used to provide an estimate of the air muscle’s force and is an implementation of equations (4) to (7) given in Chapter 2 [66].

\[
F = \frac{P \cdot g \cdot (3L^2 - b^2)}{4\pi n^2}
\]

Figure 3-7: Pressure and length to force model

The force and length to mass model is used to determine the change in mass required given the change in force and length. The formula is derived from the volume formula for the air muscle given in [66] and from the Ideal Gas Law. The model’s formula, implemented in Figure 3-8, is:

\[
M = \frac{F_{est} L \cdot M_w (b^2 - L^2)}{RT (3L^2 - b^2)} \quad (21)
\]

where \( b \) = thread length, \( L \) = muscle length, \( M \) = mass, \( F_{est} \) = the integrated delta force, \( R = 8.3145 \), \( M_w \) = molecular weight of the gas, and \( T \) = temperature in Kelvin.
As shown in Figure 3-9, the force and pressure to length model is used to update the estimate of the air muscle length as a function of the force and pressure change in the muscle:

\[ L = \sqrt{\frac{4\pi n^2 F}{3 P_L} + \frac{b^2}{3}} \]  

(22)

The valve model, shown in Figure 3-10, takes the estimated change in mass required and determines if the inlet valve or outlet valve is to be activated. For both cases, the volumetric flow rate must be converted to the mass flow rate. For the inlet case, the change in mass is divided by the multiplication of the air source's air density, inlet volume flow rate, and the inlet PWM period. The result is the percent duty cycle that the PWM valve must be on to provide that
amount of air mass to the muscle. For the outlet case, the change in air mass is divided by the multiplication of the air density in the air muscle, outlet volume flow rate, and the outlet PWM period. If the mass flow, $\frac{dm}{dt}$, is positive, then the inlet valve is executed and if the mass flow is negative, then the exhaust valve is executed.

$$\text{DM} / \text{dt} (\text{g/s})$$

**Tank**

Air Density (g/m$^3$)

Inlet Volume Flow Rate through Valve (L/s)

Outlet Volume Flow Rate through Valve (L/s)

Neg/Pos Flow

Zero

Duty cycle (%)

Duty cycle (%)

Pos Flow

Duty cycle (%)

Neg Flow

PWM Valve Exhaust Port

Outlet Pressure (Pa)

Outlet PWM Period (ms)

Outlet Volume Flow Rate through Valve (L/s)

Outlet Pressure (Pa)

Outlet PWM Period (ms)

PWM Valve Port Model

Figure 3-10: Valve and PWM model

The PWM valve port model uses the downstream to upstream pressure ratio to determine if there is choked or unchoked flow. If the pressure ratio is below 0.528, then the flow is sonic (choked) otherwise the flow is subsonic (unchoked). For choked flow, the mass flow is simply a function of the temperature, orifice area and upstream pressure and is given by the equation:

$$M = \frac{4.04 \times 10^{-2} \cdot A_{orifice} \cdot P_{up}}{\sqrt{T}}$$ (23)

For unchoked flow, the mass flow is:

$$M = 3.48 \times 10^{-3} \cdot A_{orifice} \cdot \left[2009 \cdot \left(\frac{P_{down}}{P_{up}}\right)^{2/k} \cdot \left(1 - \frac{P_{down}}{P_{up}}\right)^{(k-1)/k}\right]$$ (24)
where \( M \) is the mass flow, \( A_{orifice} \) is the orifice area, \( P_{up} \) is the upstream pressure, \( P_{down} \) is the downstream pressure, and \( k=1.4 \) for dry air. An implementation of these formulae in Simulink® is shown in Figure 3-11.

![Diagram of PWM valve implementation](image)

**Figure 3-11: PWM valve implementation**

The PWM model takes a duty cycle (%) and a PWM period in milliseconds and creates the PWM waveform that turns the valve on and off. The timer is the implementation of a digital timer that takes an analog clock input (such as the Simulink sample time) and time interval. When the timer reaches or exceeds the timer interval, the timer issues an alarm signal and resets the current clock output. The current clock output will count from zero to the time interval entered into the timer block. Both the PWM inlet valve and the PWM outlet valve are structurally identical. The mass output from the valve block provides the actual mass flow from the desired mass flow.
The air muscle model based on equations (6) and (7) and is implemented in Figure 3-13. The gauge pressure is calculated from the current muscle length, mass of air, pre-calculated fibre length, and number of turns of each fibre. These parameters are used to calculate the current muscle volume. The Ideal Gas Law is then used to determine the absolute pressure. The atmospheric pressure is subtracted from this value to give the gauge pressure. This result is then used in equation (7) along with the current muscle length to determine the output force from the air muscle. The output force from this equation is passed through a low pass RC filter that approximates (25) the air muscle's time delay as a first order response with a time constant of approximately 79 ms.

\[ H(s) = \frac{2nf}{s + 2nf} = \frac{1}{1 + \tau s} \quad (25) \]

The PID controller gains were determined through a combination of preliminary experimentation and simulation. The PID gains determined solely on simulation were inadequate in predicting the actual reaction of the subject and system dynamics. For example, the reaction time for the air
muscle in the simulation was typically more responsive than the physical air muscle. Consequently, the subject noticed a significant delay in the air muscle activation. As a result, it was decided that the PID gains used in the experimental setup had to be determined through experimentation. To allow comparison between the simulation and the experimental setup, the experimentally determined gains of 0.8333 and 0.75, for proportional and integral respectively, were implemented as the gains in the simulation. However, the experimentally determined derivative gain could not be implemented in the simulation due to extraordinary small simulation step sizes required to prevent simulation errors. As a result, the simulation would take considerable time with little increase in simulation accuracy.

3.5 Simulated Results

The simulation was performed using a variable step solver with a maximum step size of 0.0001 and an initial step size of 0.000005 (ode113) [100]. The small step size was a requirement of the full simulation of the PWM valve system. The solver is a variable-order Adams-Bashforth-Moulton PECE solver, which is more efficient than a Runge-Kutta at stringent tolerances [100]. To gauge the simulation’s reaction to variable system inputs, the simulation receives descending command signals of 1, 0.8, 0.6, 0.4, 0.2, and 0.1. These descending command signal levels correspond to %MVC levels of 100%, 80%, 60%, 40%, 20%, and 10% respectively. Changing the %MVC levels allows observation of the system for varying masses of the cylinder lifted. The maximum grasp force for this simulation was set to 70 N.

3.5.1 Verification of Simulated EMG Response

The activation signal, simulated EMG, and windowed EMG are shown in Figure 3-14, where the windowed EMG is the RMS value over the previous 200 ms. The activation signal quickly rises to the descending command level with an initial sharp spike. The spike results from several different aspects of the simulation. The first cause is the step signal for descending command, which results in extreme changes that normally would not be exhibited by a person. The second cause is the lack of initial feedback from the neuromuscular models. The lack of feedback results in an error signal as large as the descending command signal and when passed through the PID automatic response controller and added to the feedforward command signal, results in a large peak during descending command signals changes. This peak is not a problem for the simulation
on a whole since it is quickly eliminated when it passes through the simulated neuromuscular system.

The simulated EMG appears similar to measured EMG data; however, it does not reproduce the noise on the real EMG sensor that would be observed in the intervals between lifts. Finally, the windowed EMG was rectified and a root-mean-squared value was computed over intervals of 200 ms and display typical results that were taken during preliminary experiments performed to confirm the validity of the simulation. Similar results are observed for varying levels of activation input as shown in Figure 3-15.

Figure 3-14: Verification of Simulated EMG and Windowed EMG for a single lift with an activation signal of 1
3.5.2 Simulated Unassisted Lift Results

The simulation results for the unassisted lift system (Figure 3-16) show that the simulated system design performs as desired. The average %MVC and grasp force are given in Table 3-1. The %MVC is close to the expected behaviour, and the grasp force is slightly lower than the expected values. For example, the hand force for an activation signal of 1 should theoretically be 70 N; however, in the simulation the result is 66.2 N. The discrepancy is due to the force generated by the simulated motoric system passing through the simulated reactive system. Since the reactive system has a gain of approximately 2.8 for relatively low frequency signals, the sensed force is slightly higher than the force measured at the grasp. The sensed force causes the neuromotor PID controller to adjust the descending command signal to be lower resulting in a lower grasp force. These results are similar for varying levels of activation signal input as is observed in Figure 3-17.
3.5 Methodology and Experimental System Design and Simulation - Simulated Results

3.5.3 Simulated Supplemented Lift Results

In response to the force supplementation, the %MVC and hand force were expected to start at high levels and decrease to half their original values as the air muscle became more active. Although the %MVC decreased by approximately 50% (as observed in Table 3-2), the reduction in this value is nearly immediate and instead of starting at a higher level and decreasing (as observed in Figure 3-18). The %MVC started at approximately 50% and initially decreased in response to the air muscle’s activation to approximately 25 %MVC then increasing to 50% over
time. Moreover, the reduction in the grasp force is lower than was anticipated. This result can be understood as an artifact of the linear models used to model the human neuromotor control behaviour.

<table>
<thead>
<tr>
<th>Activation Signal</th>
<th>Average %MVC</th>
<th>Air Muscle Force (N)</th>
<th>Human Grip Force (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.0</td>
<td>49.0</td>
<td>34.3</td>
<td>42.9</td>
</tr>
<tr>
<td>0.8</td>
<td>38.3</td>
<td>28.9</td>
<td>33.3</td>
</tr>
<tr>
<td>0.6</td>
<td>27.9</td>
<td>22.9</td>
<td>24.0</td>
</tr>
<tr>
<td>0.4</td>
<td>17.4</td>
<td>16.4</td>
<td>15.3</td>
</tr>
<tr>
<td>0.2</td>
<td>7.8</td>
<td>9.4</td>
<td>6.9</td>
</tr>
<tr>
<td>0.1</td>
<td>3.5</td>
<td>5.1</td>
<td>3.2</td>
</tr>
</tbody>
</table>

Figure 3-18: Assisted Lift Simulation Results for a single lift with an activation signal of 1

In this simulation, the neuromotor system is modeled as a simple PID automatic response system with Fagergren's linear models [103] as the plant and feedback path, which is insufficient to correctly model the adaptive behaviour of the neuromotor system. The PID automatic response system, in comparison to a person, has a faster response, resulting in the very quick reduction exhibited in the %MVC signal; however, it is unclear whether the fast reduction was due to the air muscle controller or the PID automatic response system controller.
3.6 - Methodology and Experimental System Design and Simulation - Summary

Although the major physical components of the experimental setup were determined from previous work by Clapa [104], the simulation helped assess the control structure that should be applied to the system. By using Colbrunn’s air muscle force estimation equation [66], control of the air muscle system only required the pressure and length of the air muscle. Finally, Duque’s model [99] for determining grasp force based on %MVC allowed a simple method for accurately determining the required force for the air muscle controller.

The simulation allowed the assessment of the safety of the control system for these components in their use with force supplementation of a person’s grasp force. It allowed this assessment by providing a potential human response through Fagergren’s models [103] of the motoric and reactive systems.

![Figure 3-19: Assisted Lift Simulation Results for multiple lifts with varying activation signals](image-url)
4 Experimental Setup

From the successful system simulation results, we can be reasonably confident that the force supplementation system will operate within the specified parameters. This chapter provides a description of each physical system component and the overall operation of the force supplementation system. Once the experimental system components have been sufficiently explained, the experimental procedures are developed and the results are presented in the following chapter.

4.1 Overview

The experimental setup is shown in Figure 4-1 with all the major components identified. It is designed to prevent the subject from moving any other muscles apart from their grasping muscles during the experiment. Immobilization is important to prevent interference in the EMG reading from the other muscles. The armrest is designed to fix the subject’s arm at a typical angle for grasping an object on a table at waist height. The subject is requested to adjust the height of their chair to the appropriate height for the grasping position. Arm straps are used to keep the subject’s arm stationary during the experiment and are adjustable to different arm lengths. The hand strap is used to apply the force generated by the air muscle to the subject’s hand while allowing the subject to have the same gripping surface for both the assisted and unassisted trials. The thumb spacer is used to maintain an ulnar-radial angle of 0°. Since the subject’s arm is stationary, the cylinder is placed in their hand and the subject is instructed to lift the cylinder using only a finger grasp.

The spring is used to balance the force at the hand strap generated by the elasticity of the rubber tubing present inside the air muscle. The spring force acts against the air muscle force through a timing belt attached to an encoder. The encoder is used to measure the muscle length.
The force-instrumented cylinder, Figure 4-2, contains a force sensor calibrated in compression. The sandblasted area is used to keep the subject’s fingers in the same position for each trial and over different subjects. The area where the subject's fingers were placed was smoothed to create a worst-case grasping surface. The aluminum sections of the cylinder were of insufficient mass to elicit sufficient EMG from the subjects and therefore, the steel plate in the cylinder was added to provide additional mass.
4.2 Air Muscle and Air Muscle Controller Design

The air muscle shown in Figure 4-3 is designed to provide a force range of 0 N to 90 N for a length of 28 mm to 35 mm and pressure of 0 to 100 PSI. This force range is typical for the grasp force of an average person lifting a heavy object; however, it is low enough to ensure no permanent injury to the person would result. The stiffness, in combination with the spring, is designed to provide a “spongy” feeling to the subject. This “sponginess” was considered more comfortable to the subject and since precise position is not controlled, it was unnecessary to create an overly stiff system. The measured values for the number of circumferential trapezoids (B) and lengthwise trapezoids (A) were 36 and 140 respectively and the length of a trapezoid side (ℓ) was 0.00196 m. These values yielded b and n from (4) and (5) provided in Chapter 2 [66] of 0.5488 m and 3.89 respectively. Finally, the force of the air muscle is measured for experimental purposes using a force sensor attached to the air muscle and calibrated in tension.

Like Colbrunn, the experimental setup's air muscle did not exhibit ideal behaviour and required a correction factor determined through calibration. The air muscle, at a fixed length, was inflated with a setpoint force increasing in 1 N/s increments from 0 to 65 N followed by decrementing 1 N/s back to 0 N as observed in Figure 4-4. To reduce the effects of initial inflation problems, the air muscle was initially inflated to 10 N and reduced to 1 N. This data was not included in the calibration of the air muscle. To ensure that the air muscle force matched the setpoint force, the force measurements from the air muscle were used as the feedback signal for a PID controller controlling the inflation.

The estimated force calculated from Colbrunn’s equation was compared to the measured force and a series of mapping polynomials were developed [66]. These polynomials mapped the
estimated force to the measured force and the error was tabulated (Figure 4-4). It was important to map the estimated force to the measured force since during the experiment, only pressure and length measurements would be used to control the air muscle's force. It was determined that a cubic polynomial (Figure 4-5) of equation (26) was sufficient as it reduced the RMS\(^6\) error to below 1 N.

\[
F_{\text{mapped}} = -1.3399 \times 10^{-4} F_{\text{est}}^3 + 2.4102 \times 10^{-2} F_{\text{est}}^2 \\
\ldots - 3.8705 \times 10^{-1} F_{\text{est}} - 1.2212
\]

(26)

![Before Mapping Estimated Force](image1)

![After Mapping Estimated Force](image2)

Figure 4-4: Calibration of Air Muscle

\(^6\) Root-Mean-Square
The PID gains for the air muscle controller were determined partially to obtain the best response from the air muscle and partially through experimentation with a subject. Initially, the air muscle was positioned at fixed length to be expected during the experiment. The air muscle was subjected to typical inputs to determine its response (step, sinusoid, staircase). The proportional and integral gains were adjusted to 0.6 and 0.833 respectively until the air muscle exhibited satisfactory accuracy and rise time. The experimental setup was determined to be safe enough to begin preliminary testing on a subject.

The subject, with training in controls, was asked to gauge the performance of the system and recommend adjustments to the PID gains. Table 4-1 shows the progression of the gain adjustments made in order to reduce system, without significant force overshoot or system oscillation. The last row of the table shows the finalized gains.
4.3 Pressure Measurement Components

4.3.1 Pressure Transducer

The selected pressure transducer is an AutoTran, Inc Model 250 and is shown in Figure 4-6 [105]. The sensor has a gauge pressure range from 0 to 100 PSI with 1% full-scale output (FSO) accuracy. The transducer is relatively inexpensive, required no user adjustments, and supplied a satisfactory accuracy in the required pressure range. The transducer outputs a 4-20 mA output that is commonly used in industrial environments. The 4-20 mA output was chosen due to noise problems experienced with 1-5 V output sensors of the same model. The 4-20 mA sensor had the additional advantage of being powered using the 4-20 mA current loop and therefore only required a 2-wire connection. The 1-5 V sensor required the use of a 3-wire connection. Since the orthotic system was a prototype, no consideration on the transducer size was taken into account. In an actual orthotic system, the pressure transducer would have to be much smaller. Details of the transducer and the transducer circuitry are provided in Appendix A.01.

![AutoTran Model 250 Pressure Transducer](image_url)
4.3.2 Pressure Transducer Calibration

The pressure transducer was calibrated using an existing calibrated sensor. The pressure was systematically increased in discrete levels and measurements from both pressure sensors were taken during this time. The voltage measured on the uncalibrated pressure sensor was plotted against the pressure measurement from the calibrated sensor and the following linear relationship was derived:

\[ P_{\text{PSI}} = 17V + \text{offset} \]  

(27)

However, the offset was found to shift every time the system was started and therefore the offset is adjusted automatically as an average of the first 100 ms following a 2 second wait to allow the DSP system to settle.

4.4 Force Measurement Components

4.4.1 Grip Force Transducer

The grip force transducer that was selected is a Precision Transducers, Inc PT4000 and is shown in Figure 4-7 [106]. The sensor has a force range from 0 to 50 kg of force (490 N) with a combined error of 0.02%. This larger range was selected since the force transducer must measure the air muscle’s force combined with the subject’s grip force. The transducer supplied a satisfactory accuracy in the required force range. The transducer outputs 3 mV for every Volt of excitation voltage. The grip force sensor is the sensor mounted inside the cylinder as shown in Figure 4-2. Since this force sensor is used only to observe the experimental data and not for control purposes, no consideration on the transducer size was taken into account. Details of the force transducer and circuitry are given in Appendix A.02.

![Figure 4-7: Precision Transducers PT4000 [106]](image-url)
4.4.2 **Air Muscle Force Transducer**

The selected air muscle force transducer is a Precision Transducers, Inc PST20 and is similar to the transducer shown in Figure 4-7. The sensor has a force range from 0 to 20 kg of force (196 N) with a combined error of 0.01%. Since this force transducer has to measure only the subject’s grip force, a lower force transducer was selected. The transducer outputs 2 mV for every Volt of excitation voltage. Since the air muscle only required a force transducer to verify the force being calculated from the pressure and length were accurate, the transducer size was not a consideration. For more detailed technical information on the force transducer and related circuitry consult Appendix A.02.

4.4.3 **Force Transducer Calibration**

The grip force transducer was calibrated in compression and the air muscle force transducer was calibrated in tension. The grip force transducer was connected to the floor and a bucket was placed on the sensor. The force was increased by weighing and then placing objects into the bucket. The following linear relationship was derived from the calibration.

\[ F = 29.3V + 28.0 \quad (28) \]

The air muscle force transducer was calibrated in tension by attaching the sensor to the air muscle-mounting rig and attaching a cable to the bucket over a pulley. The force was then increased by weighing objects and placing them into the bucket. The following linear relationship was found:

\[ F = 19.611V - 1.6174 \quad (29) \]

4.5 **Length Measurement Components**

4.5.1 **Encoder**

The encoder is a US Digital S1-360-IB shown in Figure 4-8, which is a 360 count per revolution free spinning encoder [107]. The encoder has the advantages of having a small size and low cost and provides a 2-channel quadrature, TTL square wave output as well as a 3\(^{rd}\) channel index. Since the MFIO3B data capture board has all the required circuitry to decode the encoder’s three channels, the encoder requires no additional circuitry other than a single 5 V power supply.
4.5.2 Encoder Calibration

The encoder’s theoretical calibration was determined from the mechanics of the experimental setup. Since the encoder is turned using a timing wheel with a 14.7 mm diameter and belt, the distance per degree was calculated as 0.128 mm/°. This distance is slightly inaccurate since the timing belt tends to travel with a slight angle over its full range. To determine the error due to this angle, the timing belt was pulled specific distances along a ruler and the output distance was compared to the number of degrees turned. The result was a distance of 0.133 mm/°, which differs from the theoretical distance by approximately 4%. Therefore, over the typical operating range of approximately 4 cm, the error will be 16 mm. The sensitivity of the air muscle equation to 16 mm is within the tolerance range of error for the experimental system and this error should be corrected by the controller.

4.6 Electromyogram Measurement Components

4.6.1 EMG Electrode

The EMG electrode, shown in Figure 4-9, is a miniature self-amplified sensor designed at Simon Fraser University (SFU) in the Neuromuscular Control Lab (NCL). The sensor provides an output range of −5 V to 5 V before clipping and requires dual supply voltages of −5 V and 5 V. The electrode also contains band pass filtering circuitry with cutoff frequencies of 30 Hz and 450 Hz, which is typical for EMG applications. More information about the EMG electrode and its additional circuitry can be found in Appendix A.03.
4.7 Valve and Valve Driver

4.7.1 Valve and Valve Modifications

The valve used is a Matrix GNK821.203C3KK (shown in Figure 4-10), which is a PWM valve with a response time 3 ms and 1 ms for opening and closing respectively. The valve requires a speed up voltage of 24 V for the first 2.5 ms and a tension voltage of 5 V.

Since the flow rate for the valve at 90 nominal liters per minute (NL/min) is excessive for the air muscle application, valve plates with smaller orifice sizes must be inserted into the valve to restrict flow. An example valve plates is shown in Figure 4-11. Two plates were necessary: one for the outlet, and one for the inlet. Each plate required two holes. The smaller center hole is the hole of interest. The second hole is the same size as the corresponding hole inside the valve. The orifice diameters are 0.508 mm and 0.7874 mm for the inlet and outlet plates respectively. The orifice diameters were determined by calibration and sized to provide a similar fill and deflate time. Since the ratio of source pressure to air muscle pressure is greater than the ratio of air muscle pressure to atmospheric pressure, the inlet orifice diameter must be smaller to provide the same fill and deflate times. Details of the valve driver circuitry are found in Appendix A.04.
4.8 Summary

Each physical system component and the overall operation of the force supplementation system were described in this chapter. Upon assembly of these components, experimental procedures and system calibration can be discussed in the following section.
5 Experiments

The experiments were designed to determine if muscle activation and grasp force decrease when force supplementation is applied. Therefore, a control experiment was devised where the subject lifted the cylinder unassisted followed by an experiment where the subject lifted the cylinder assisted. This chapter describes the experimental system calibration procedures, the experimental procedures, and, finally, the experimental results. The experiments begin with placement of the EMG electrode on the subject’s arm, placement of the arm in the apparatus, and then determination of the MVC. Then the subject goes through a series of procedures for the control experiment, where the lift of the cylinder is unassisted, and then for the assisted lift experiment, where the lift of the cylinder is assisted. Statistical analysis of the data at the end of the chapter shows the comparative effectiveness of the apparatus in the assisted lift experiments. The subjects were recruited from graduate laboratories and consisted of 7 men and 2 women, ages 19 to 27.

5.1 Initial Setup

The subject was required to read the ethics form shown in Appendix G and ask any questions about the experiment. The ethics form was either signed or not depending on the subject’s willingness to participate. If the subject agreed to participate, he was seated next to the experimental setup on an adjustable lift chair. The subject was asked to adjust the chair to a position where the can would be picked up at waist height.

5.2 EMG Electrode Placement and MVC Determination

Since the position of the EMG electrode was on the flexor digitorum profundus muscle, which is on the underside of the arm, the electrode had to be placed before positioning the arm in the
experimental setup. The subject was instructed to stand in front of a table and grasp the top of the cylinder. This position was similar to the position used during the experiment. The subject was instructed to grasp the cylinder with a light grip force. During the subject’s grasp, the stiffest forearm muscle was found in the area corresponding to the flexor digitorum profundus muscle as shown in Figure 5-1 and a small amount of electrode gel was applied to the EMG electrode if it was necessary. The electrode was placed at a spot slightly higher than the bulge of the muscle. Using an oscilloscope to monitor the signal, the sensor was moved around the muscle area until a satisfactory signal is received. The subject was then asked to exert a large grip force. If the EMG signal was satisfactory, then the EMG sensor was then taped using medical tape to the subject’s arm. The subject was again asked to exert a large grip force while the experimenter observed the oscilloscope. If the EMG signal was not using the full range of the ADC, -5 V to 5 V, then the gain was increased on the EMG board until nearly the full range was used. If the EMG signal was cutoff indicating the railing of one of the amplifiers, then the gain was decreased until railing no longer occurred.

![Flexor Digitorum Profundus muscle](image)

Figure 5-1: Flexor Digitorum Profundus adapted from [110]

5.3 Arm and Hand Positioning

The position of the subject’s arm was important in maintaining consistency between all the subjects’ results. Two main marking points were identified and marked to aid in the positioning the arm. The marking points were the ulnar styloid process (tip of the ulnar bone) and the radial styloid process (tip of the radius) shown in Figure 5-2. The locations of these two points were selected based on the ease of locating these points on all subjects.
The reference points marked earlier were lined up with calibrated marking points on the experimental setup (see section 4). Once the marking points were in position, the subject’s arm was strapped down using the Velcro straps. It was important to ensure that the straps hold the subject’s arm stationary while maintaining the subject’s comfort.

The effectiveness of the subject’s muscles to generate force was determined by %MVC in conjunction with the wrist angles as shown in Figure 5-3. The wrist angle was maintained stationary and the same for all the subjects. Both the flexion-extension and the ulnar-radial angle were chosen to be approximately $0^\circ$ to maintain the hand and forearm in the same plane. The wrist brace placed on the arm previously maintained the correct wrist angle. Once the subject was in position, an appropriately sized thumb spacer was used to support the subject’s thumb at the consistent distance compared to other subjects. The thumb spacer was kept in place for the entire experiment to maintain the wrist position. The cylinder was placed on an adjustable platform to allow the finger positions to be in the same place for all subjects given the $0^\circ$ wrist angle.
The ORTS system was setup to begin capturing data from the force sensor and the EMG electrode using the *DetMVC* script given in Appendix H. During this part of the experiment, the first 10 seconds were used for calibration of the cylinder force sensor. The script took digitally filtered (at 7 Hz) root-mean-square windows of the EMG signal over 100 ms and continuous measurements from the cylinder’s force transducer. From these two signals, the script determined the average maximum and average minimum values for both the EMG signal and the force signal. The subject was instructed to grasp the cylinder with a maximum squeeze while the light emitting diode (LED) was active. The LED was programmed to be on for a 2 second duration followed by a 6 second rest period and was repeated for a 60 second duration. This step was performed three times to obtain a statistical average for both the maximum force and the MVC. A rest period of 2 minutes was taken after each exertion set to prevent muscle fatigue from affecting the results [99]. Due to limitations of experiment length, an estimate of the maximum and minimum EMG signal level and grip force was taken from a sample of 100 of the highest and lowest measured values respectively. This estimate allowed for a faster calibration of the subject to the experimental system. A statistical analysis of the validity of this estimate is given in section 5.6.
5.4 Control Experiment – Unassisted Cylinder Lift

Experiment 1 had two main purposes. The first purpose was to allow the subject to become comfortable with the weight of the cylinder. It was hypothesized that the subject would use excessive force at the beginning of the experiment and this force would decrease with the number of times that the cylinder was picked up. The second purpose of the experiment was to determine the baseline for the subject’s %MVC and grasp force during pick up and put down of the cylinder. The baseline was used as a comparison for the following two experiments.

The complete procedure, summarized here, for this experiment is given in Appendix C and the ORTS script is given in Appendix H. For the first 20 seconds, the system is in calibration mode and the subject was instructed to relax their muscles and to not squeeze the cylinder. Next, the subject was instructed to lift the cylinder when the LED was on (4 seconds) and put down the cylinder when the LED shut off (4 seconds). The subject was also instructed to pick up the cylinder with the least amount of force possible without dropping it using only a finger grasp (i.e. minimal lifting of the arm or bending of the wrist). During this experiment, measurements from the force sensor in the cylinder and the EMG electrode were taken. Following the first iteration of this experiment, the data was checked to ensure that both the EMG signal and grasp force were measured correct (i.e. the EMG did not saturate). Five lifts were performed per trial followed by a 1-minute rest break between each trial. Three trials were taken for each experiment.

5.5 Assisted Lift Experiment

The subject was allowed to rest, without removal from the setup, while the air muscle was being set up. The air muscle was attached to the strap’s bar and the bar’s height was adjusted to be in the same plane as the subject’s fingertips. The subject’s fingertips were placed in the air muscle’s strap and held in place using elastic fabric. The elastic fabric did not cover the person’s fingertips to maintain the same grip surface and proprioception on the cylinder between the Control Experiment and the Assisted Lift Experiment.

The subject was instructed to relax his grip as much as possible to prevent sudden increases in force when the air muscle system was started. The air muscle was disconnected during the initial 20-second calibration period to allow for the calibration of the air muscle’s force sensor. The air
muscle’s initial force from the elasticity of the rubber tubing in the air muscle was compensated by using a spring in opposition to the air muscle. The subject was instructed to lift the cylinder when the LED was on and place the cylinder back on the table when the LED was off, except for the first two periods. During the first two periods, the air muscle was disconnected and the subject was instructed to perform a maximum squeeze of the cylinder, without lift, followed by a normal unaided lift. These periods were used to confirm that the data taken during the Assisted Lift Experiment were consistent with the %MVC calibration as well as the Control Experiment.

A complete, itemized list of the experimental procedure is given in Appendix C and the ORTS script is provided in Appendix H.

5.6 Validation of the 100 Sample Estimate for MVC Calibration

Ideally, a complete statistical analysis of the subject’s maximum grip force and EMG levels would be performed before the experiments are performed. For this analysis, the mean and standard deviation would be performed for each maximum squeeze and the statistical error would be calculated between calibration times. Unfortunately, performing this analysis with the supplied technology could not be sufficiently automated to allow completion of the calibration in a satisfactory experiment length. Therefore, a more expedient method was devised where the subject would perform a maximum squeeze three times over a period of 30 seconds.

During the 30 seconds, the data acquisition system, ORTS, tracked the top 100 maximum grip force levels and EMG levels registered on the sensors for three maximum squeezes. ORTS then obtained the mean of these 100 values and presented these means to the experimenter, which entered them into the experiments’ script files. This procedure was performed two or more times depending on the consistency of the means for each trial. The purpose of this section is to add statistical validity to this method of determining the maximum grip force and EMG levels.

The actual and estimated levels are shown in Figure 5-4. The method used for determining the maximum grasp force and contraction levels provided an additional safety measure. Since the algorithm to determine the estimated levels only retains the highest 100 samples, it is susceptible to selecting values that are on the higher end of the probability density spectrum. As shown, for all subjects, the estimated levels are higher than the actual calculated mean levels. Thus, when the EMG signal is scaled to a %MVC, a slightly smaller signal is used to modulate the air muscle’s force output resulting in a smaller force generated by the air muscle.
One can note that when comparing the estimated levels to the actual levels, the estimated levels are all within three standard deviations of the mean actual levels as summarized in Table 2-1. When a paired Student t-test was performed across all subjects, the estimated levels represented the actual mean levels (p < 0.999). This result indicates that estimated levels provide a sufficient indicator of the actual mean levels and should be sufficient for estimation.

Table 5-1: Comparison of estimated maximum EMG and force levels to statistically generated levels

<table>
<thead>
<tr>
<th>Subject Number</th>
<th>Actual Levels</th>
<th>Estimated Levels</th>
<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Mean EMG Level</td>
<td>Mean Maximum Force</td>
<td>EMG Level</td>
<td>Maximum Force</td>
<td>EMG Δsd</td>
<td>Force Δsd</td>
</tr>
<tr>
<td>1</td>
<td>4.67</td>
<td>122.20</td>
<td>5.70</td>
<td>134.86</td>
<td>0.94</td>
<td>0.81</td>
</tr>
<tr>
<td>2</td>
<td>0.87</td>
<td>47.98</td>
<td>1.27</td>
<td>61.84</td>
<td>1.49</td>
<td>2.17</td>
</tr>
<tr>
<td>3</td>
<td>3.91</td>
<td>71.63</td>
<td>4.55</td>
<td>81.20</td>
<td>0.56</td>
<td>1.06</td>
</tr>
<tr>
<td>4</td>
<td>3.42</td>
<td>78.88</td>
<td>4.45</td>
<td>96.38</td>
<td>1.09</td>
<td>1.38</td>
</tr>
<tr>
<td>5</td>
<td>4.48</td>
<td>60.87</td>
<td>5.25</td>
<td>80.00</td>
<td>0.73</td>
<td>2.66</td>
</tr>
<tr>
<td>6</td>
<td>3.68</td>
<td>70.80</td>
<td>4.38</td>
<td>102.60</td>
<td>0.93</td>
<td>2.87</td>
</tr>
<tr>
<td>7</td>
<td>0.93</td>
<td>23.30</td>
<td>1.20</td>
<td>31.00</td>
<td>0.78</td>
<td>2.20</td>
</tr>
<tr>
<td>8</td>
<td>2.05</td>
<td>113.01</td>
<td>2.65</td>
<td>140.80</td>
<td>1.12</td>
<td>1.60</td>
</tr>
<tr>
<td>9</td>
<td>3.96</td>
<td>71.25</td>
<td>4.90</td>
<td>93.00</td>
<td>1.14</td>
<td>2.77</td>
</tr>
</tbody>
</table>

5.7 Preprocessing of Raw Experimental Data

For each subject, the data was plotted and observed for any anomalies and inconsistencies between the force data and the %MVC data. For each lift, data was eliminated if the force data and %MVC data did not exhibit the same (coordinated) lift features (increase at the onset of the lift, sustained period, followed by a decrease at the end of the lift). This step prevented a loose
EMG sensor from affecting the resultant data. Following this elimination procedure, the mean and standard deviation were calculated for the %MVC and forces for each lift. These means and standard deviations are named lift mean and lift standard deviations in this document. Finally, the lift means and lift standard deviations were used to calculate an overall mean and standard deviation for each subject called a subject mean and subject standard deviation. The subject means and subject standard deviations are used for sections 5.8, 5.9, and 5.10 whereas, the lift means and lift standard deviations are used in section 5.12.

For example (Figure 5-5), subject 3 has seven lifts in the control experiment. However, the %MVC and grasp force exhibited inconsistencies in the first lift and this lift had to be eliminated. The %MVC for the eliminated lift stayed high during the immediate period following the lift and therefore some ambiguity exists as to the %MVC level used during the lift. The lift means and lift standard deviations for the %MVC and the grasp force were obtained for the remaining six-lifts. These lift means were then used to calculate a subject mean and subject standard deviation for this trial.

![Figure 5-5: Example of Anomalous Data for Subject 3 during the Control Experiment Trial](image)

### 5.8 MVC Calibration to Maximum Squeeze in Assisted Lift

As described in section 5.5, the subject was instructed to perform a maximum squeeze as a consistency check before each trial for the Assisted Lift experiment. This maximum squeeze was required to ensure that the MVC calibration was consistent with the Assisted Lift experiment data. In this section, these maximum squeezes are compared to the MVC calibration to determine
the validity of the EMG used in the calibration for the data recorded during the Assisted Lift experiment. All plots of the MVC calibration data are presented in Appendix D.

As shown in Figure 5-6, significant discrepancies exist between the MVC calibration and the Assisted Lift Maximum grip for subject 4, 5, and 7. However, this result does not invalidate the results or require scaling of the Assisted Lift data for these subjects. When the grasping force is compared between the MVC calibration and the maximum squeeze consistency checks, they are consistently lower. This result indicates that the subject may have become fatigued preventing them from being able to exert a maximum squeeze force on the cylinder. However, the ratio of grasping force to %MVC should remain the same if no other discrepancies are in effect. Since both Windowed EMG and Mean Force for subjects 4 and 5 were lower, it is likely due to fatigue of the subject. The fatigue indicates that the experiment should have allocated more rest time for these subjects. As for subject 7, it is likely that the sensor has shifted; however, this hypothesis is investigated in Section 5.9 where the normal lift consistency checks from the Assisted Lift experiment are compared to the Control Experiment data.

![Figure 5-6: Comparison of MVC Calibration Data to Maximum Squeeze in Assisted Lift Experiment](image_url)
A change analysis was performed to determine the validity of the fatigue assumption. For example, if the percentage change in the Windowed EMG was not significantly different, using a Student t-test, from the percentage change in the Mean Force, then it was determined that the MVC Calibration data was valid for the Assisted Lift experiment. This rule assumes that if the change in the Windowed EMG was due to fatigue, the result would be a corresponding change in the force generated by the hand. These percentage change formulas are given by equations (32) and (33) where $EMG$ is the Windowed EMG and $MF$ is the Mean Force. The subscripts $calib$ and $Assisted$ denote the MVC Calibration and Assisted Lift experiments respectively.

$$\Delta EMG = (EMG_{calib} - EMG_{Assisted})$$ (30)
$$\Delta MF = (MF_{calib} - MF_{Assisted})$$ (31)

$$\Delta% EMG = \frac{\Delta EMG}{EMG_{calib}}$$ (32)
$$\Delta% MF = \frac{\Delta MF}{MF_{calib}}$$ (33)

The standard deviations were adjusted according to equations (34) and (35) where $std_{EMG}$ is the standard deviation for the Windowed EMG and $std_{MF}$ is the standard deviation for the Mean Force. The subscripts $calib$ and $Assisted$ denote the MVC Calibration and Assisted Lift experiments respectively.

$$std_{EMG} = \Delta% EMG \sqrt{\frac{std_{EMG_{calib}}^2 + std_{EMG_{Assisted}}^2}{\Delta EMG^2} + \frac{std_{EMG_{calib}}^2}{EMG_{calib}^2}}$$ (34)
$$std_{MF} = \Delta% MF \sqrt{\frac{std_{MF_{calib}}^2 + std_{MF_{Assisted}}^2}{\Delta MF^2} + \frac{std_{MF_{calib}}^2}{MF_{calib}^2}}$$ (35)

To further confirm the fatigue assumption, the ratio of %MVC to grasp force was compared between the calibration and the maximum squeeze consistency checks using a Student t-test. If fatigue was present, the ratio of %MVC to grasp force should remain the same between the MVC calibration and the maximum squeeze consistency checks. The results for both these analyses are presented in Table 5-2. For the change analysis, 3 of 9 subjects were statistically different ($p<0.1$); however, 4 of 9 subjects were inconclusive as to whether the change was significant. The ratio of grasp force to %MVC was then compared and the result was 3 of the 9
subjects were statistically different (p<0.05) and 2 of the 9 subjects were statistically equivalent (p>0.90) with the rest of the subjects being inconclusive. Therefore, it was decided that only the ratio of %MVC to grasp force would be used for further analysis due to its simpler algorithm and less variance added to the resulting outcome with similar results. It is suspected that the ratio of grasping force to %MVC is a nonlinear relationship and therefore, at the maximum squeezes, it exhibits more sensitivity to small changes in the relationship. Finally, due to these inconclusive results for most subjects and the fact that the lift data was more consistent, it was decided to use the control experiment and the unassisted lifts consistency checks as a true measure to determine the validity across the data.

Table 5-2: Student t-test Results between the MVC Calibration and MVC consistency checks for the ratio of %MVC to the Mean Force

<table>
<thead>
<tr>
<th>Subject</th>
<th>p-value for Δ %MVC to Δ Grasp Force</th>
<th>p-value for Grasp Force to %MVC ratio</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.087900</td>
<td>0.154142</td>
</tr>
<tr>
<td>2</td>
<td>0.220905</td>
<td>0.263091</td>
</tr>
<tr>
<td>3</td>
<td>0.142384</td>
<td>0.089353</td>
</tr>
<tr>
<td>4</td>
<td>0.092153</td>
<td>0.002381</td>
</tr>
<tr>
<td>5</td>
<td>0.147790</td>
<td>0.008500</td>
</tr>
<tr>
<td>6</td>
<td>0.593602</td>
<td>0.544921</td>
</tr>
<tr>
<td>7</td>
<td>0.045377</td>
<td>0.045252</td>
</tr>
<tr>
<td>8</td>
<td>0.953418</td>
<td>0.942687</td>
</tr>
<tr>
<td>9</td>
<td>0.906183</td>
<td>0.918571</td>
</tr>
</tbody>
</table>

5.9 Control Experiment to the Normal Lift in Assisted Lift Experiment

When the control experiment data was compared to the unassisted lift consistency checks (Figure 5-7), the result observed was also quite different than expected. It was expected that the %MVC levels and hand grasp forces would remain the same between the control experiment and the consistency checks, which is not the case for most subjects. However, the result was more accurate than the comparison of the MVC Calibration to the Maximum Squeeze consistency checks and it is common for some variation in lifts [112]. It was found that the %MVC and grasp force was statistically different in 3 of the 9 subjects (p<0.05). However, the %MVC exhibited no evidence that the distributions were statistically different in 3 of 9 subjects (p>0.90). Additionally, the force demonstrated no evidence that the distributions were statistically different in 1 of 9 subjects. The remaining subjects were inconclusive.
Again, the comparison of the ratio of grasping force to %MVC (Table 5-3) was made to determine if the relationship between the grasp force and %MVC was still valid. Similar results were obtained with 3 of 9 subjects being statistically different (p<0.01), 1 of 9 subjects being statistically the same (p>0.95), and inconclusive in the remaining 5 subjects. Therefore, it was decided to scale the assisted lift experiment data according to the unassisted lift consistency checks in an attempt to correct the statistically different results.

Table 5-3: Student t-test Results between the Control Experiment and Unassisted Lift consistency checks for the ratio of %MVC to the Mean Force

<table>
<thead>
<tr>
<th>Subject</th>
<th>%MVC ratio p</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.000156</td>
</tr>
<tr>
<td>2</td>
<td>0.643342</td>
</tr>
<tr>
<td>3</td>
<td>0.839206</td>
</tr>
<tr>
<td>4</td>
<td>0.000011</td>
</tr>
<tr>
<td>5</td>
<td>0.507793</td>
</tr>
<tr>
<td>6</td>
<td>0.960285</td>
</tr>
<tr>
<td>7</td>
<td>0.548353</td>
</tr>
<tr>
<td>8</td>
<td>0.000011</td>
</tr>
<tr>
<td>9</td>
<td>0.321578</td>
</tr>
</tbody>
</table>
The scaling was to be performed on the data if there was a statistical difference in the grasp force to %MVC ratio and hence only applied to subjects 1, 4, and 8. This rule was implemented since scaling data with inconclusive statistical differences would result in increasing the standard deviation of those subjects while accomplishing little in reducing the discrepancies between the means. The scale factor only applied to the %MVC data since the force data was more reliable and stable; it was less susceptible to sensor error. The force sensors did not exhibit any wide variations like the EMG sensor. The EMG sensor was susceptible to variations due to the sensor shifting around on or losing contact with the arm and changes in skin conductance. To ensure that this scaling factor did not skew the results significantly, the analysis between the control experiment and the assisted lift experiment was performed before and after any scaling of the data. To ensure statistical accuracy the standard deviation of the scaling factor was taken into account in all calculations. The scaling factor and associated standard deviations are presented in Table 5-4.

<table>
<thead>
<tr>
<th>Subject</th>
<th>Scale Factor</th>
<th>Scale Factor Std Dev</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.664770</td>
<td>0.355953</td>
</tr>
<tr>
<td>4</td>
<td>2.364254</td>
<td>0.586860</td>
</tr>
<tr>
<td>8</td>
<td>1.832318</td>
<td>0.728817</td>
</tr>
</tbody>
</table>

5.10 Analysis of the Reduction in %MVC and Grasp Force between the Control Experiment and the Assisted Lift Experiment

For analyzing the reduction in %MVC and grasp force, Student t-tests were performed on individual subjects and as a group. To account for the scaling factors' influence on the results, the analysis was performed before and after scaling the data. The group analysis is comprised of the matched-pairs Student t-test equations (36) to (40) performed on both %MVC and grasp force.

\[
d_i = \bar{X}_{ctrl} - \bar{X}_{assist}
\]

\[
\bar{d} = \frac{\sum_{i=1}^{N} d_i}{N}
\]
5.10 - Experiments - Analysis of the Reduction in %MVC and Grasp Force between the 
Control Experiment and the Assisted Lift Experiment

\[
 s_d = \sqrt{\frac{\sum_{i=1}^{N}(d_i - \bar{d})^2}{N-1}} 
\]

\[
 t = \frac{\bar{d}}{s_d / \sqrt{N}} 
\]

\[
 dof = N - 1 
\]

where \( \bar{X}_{ctrl} \) and \( \bar{X}_{assist} \) are the control and assisted lift experiments’ subject means respectively, \( dof \) is the degrees of freedom for the t-statistic, and \( N \) is the number of subjects [113]. The individual analysis was performed with the t-statistic for small independent samples using the following equations:

\[
 D_i = \bar{X}_{ctrl} - \bar{X}_{assist} 
\]

\[
 (s_{D_i}) = \sqrt{\frac{(n_{ctrl} - 1)s_{ctrl}^2 + (n_{assist} - 1)s_{assist}^2}{n_{ctrl} + n_{assist} - 2}} \sqrt{\frac{1}{n_{ctrl}} + \frac{1}{n_{assist}}} 
\]

\[
 t_i = \frac{D_i}{(s_{D_i})_i} 
\]

\[
 dof_i = n_{ctrl} + n_{assist} - 2 
\]

where \( \bar{X}_{ctrl} \) and \( \bar{X}_{assist} \) are the control and assisted lift experiments’ subject means respectively, \( dof \) is the degrees of freedom for the t-statistic, and \( i \) denotes the subject number [113].

The aggregate Student t-test for all subjects concluded that the aided lift effort was statistically less than the unaided lift effort for both the %MVC (p<0.01) and grasp force (p<0.01). This result was the same between the scaled and unscaled data. The individual analysis found them to be statistically different in 6 of 9 subjects (p<0.05) and 8 of 9 subjects (p<0.05) respectively. Following scaling, the result was 7 of 9 subjects (p<0.05) and 8 of 9 subjects (p<0.05) respectively.

The subjects experienced an average reduction in the assisted lift experiment from the control experiment of 31% and 56% for %MVC and grasp force respectively. During the assisted lift experiment, the control method for the air muscle should track the subject’s desired grasp force resulting in a balance between the subject’s own muscles and the air muscle actuator. Ideally, the subject will adapt until approximately half the exertion is taken by the air muscle and as a result
the reduction of the %MVC and grasp force will be 50% that of the control experiment. For a majority of subjects, this behaviour is not present (Figure 5-8).

![Figure 5-8: Percent Reduction in %MVC and Grasp Force](image)

Subjects 1 and 6 follow the expected behaviour by reducing both the %MVC and grasp force by between 50 to 60%. Subjects 5, 7, and 8 follow the ideal behaviour for grasp force with a 50 to 60% reduction; however, the %MVC reduction exhibits no reduction in subject 5 and only a 35% reduction for subject 8. Although subjects 3 and 4 exhibit a more significant reduction in grasp force with an 87% reduction for subject 4 and a 114% reduction for subject 3, their %MVC reduction is only 11 and 45% respectively. The greater than 100% reduction in grasp force is due to the air muscle exerting more force on the cylinder than the subject exerted during the control experiment. This result suggests that the relative gain of the air muscle’s controller was possibly too sensitive for this subject. Finally, subjects 2 and 9 show a larger reduction in %MVC than grasp force with 12% and 1% reduction for grasp force and 16% and 41% reduction for %MVC. The larger reduction in %MVC implies that the EMG sensor was receiving cross-interference from other muscles, other than the flexor digitorum muscles, during the control experiment. When the force supplementation was applied, it aided these other muscles resulting in a larger reduction.

5.11 Typical Experimental Results

5.11.1 Typical Control Lift Results

Typical control lift results (Figure 5-9) have a very good correlation of the %MVC signal with the grasp force signal. Moreover, both signals display minimal noise between lifts. The results indicate a reliable and stable %MVC signal was available to control the air muscle actuator.
Although the %MVC signal should theoretically lead the grasp force data, this result is not the case. The delay of filter and processing caused a reduction in any leading of the %MVC signal over the grasp force. A complete collection of control lift data for all subjects is contained in Appendix E.

![Control Lift data for Subject 6](image)

**Figure 5-9: Control Lift data for Subject 6**

### 5.11.2 Typical Assisted Lift Experimental Results

The assisted lift data allows evaluation of the stability of the supplemented lift (Figure 5-10). The measurement of the hand force was not direct and the subtraction of the air muscle force from the total force measured at the cylinder provided this force. The set point force, derived from the %MVC signal, provides a relatively stable signal to the PID controller for the air muscle; however, the air muscle force tracks this set point poorly. This result is not surprising as the air muscle is a nonlinear device controlled by linear techniques. More complex control methods should result in an improved tracking ability. Nevertheless, the tracking is sufficient for the purposes of the experiment. The PWM valve's inflation and deflation are responsible for the air muscle force exhibiting a significant amount of vibration. The damping characteristics of the air muscle appear to sufficiently reduce these vibrations. The lack of high frequency oscillations in the force measured at the cylinder as well as subjects' lack of complaint when asked about the vibration supports this conclusion.
An initial peak in the %MVC and hand force is present in many of the lifts. This peak is a result of the air muscle’s time to inflation delay. The subject gives the air muscle controller an initial peak to quicken the air muscle’s response to the subject and once the muscle begins taking the load, the subject relaxes. The subject initiates another peak to increase the force of the air muscle when the cylinder is slipping (peak 4). Additionally, some subjects demonstrate a decrease in %MVC and hand force through the 5 lifts. The first lift will have a larger average %MVC and hand force than the last lift. Interestingly, in the second trial, the subject again begins with a larger lift and decreases through the five lifts.

The comparison between the average %MVC and grasp forces for the control experiment and the assisted lift experiment results in several interesting conclusions (Figure 5-11). Since the %MVC is relative to the amount of exertion the subject uses to generate the required grasp force, the %MVC levels are quite variable between the subjects; less that 15% for subject 1 and more than 100% for subject 7. Conversely, the grasping force is relatively constant at 25 to 35 N. The primary reason that the grasp force stays relatively constant linked to the mass of the cylinder. To maintain a lift, the subject must maintain a friction force between the cylinder and the fingertips greater than the weight of the cylinder at 25 N. Nevertheless, due to the strength of each subject’s forearm muscles being different, the %MVC required to generate this force is also
different. It was informally noted that subjects with more forearm musculature were capable of generating more grasp force with less %MVC than the less muscular subjects.

During the assisted lift experiment, the cylinder force (the force of the air muscle and the subject’s grasp) is significantly higher than is necessary to lift the cylinder. The PID controller caused the air muscle to frequently overshoot the set point force. Moreover, the force overshoot is not present in all subjects indicating that tuning the PID control parameters for each subject would result in improved performance for the system.

Figure 5-11: Changes in the measured contraction levels (top) and applied forces (bottom) when the air muscle actuator is activated.

5.12 Learning Effect

5.12.1 Immediate Adaptation without Stiffening of the Arm

The %MVC for subject 6 in trial 2 or subject 8 in trial 1 is significantly lower in the assisted lift than the unassisted lift. Subject 6 did not exhibit a significant peak to get the air muscle activated; however, they did exhibit a small negative force when the cylinder was released (Figure 5-12). The negative force is likely due to a “snap-back” effect caused by a fast deflation of the air muscle. This effect is not present for all subjects since it depends on individual variability in how quickly the subject is able to deactivate the air muscle. Because both the %MVC and hand force are lower for this subject and do not experience the same large peaks, it
is likely that this subject did not stiffen the arm to lift the cylinder. Nevertheless, Subject 8 experienced a distinct peak in the hand force lift and did not experience a negative peak when the cylinder was released (Figure 5-13). This result is very similar to the simulated results. Since no pattern in muscle activation and grasp force was associated with the lift, neither of these two subjects showed any evidence of a learning effect.

![Graphs showing various forces and percentages over time.](image)

**Figure 5-12:** Assisted Lift Experiment for Subject 6 Trial 2 demonstrating immediate adaptation without subject stiffening arm
5.12.2 No Adaptation, Subject Stiffens Arm

A demonstration where the subject presented no adaptation and simply stiffened the arm is shown in Figure 5-14. For subject 5 in trial 3, the subject’s hand force decreases; however, the %MVC stays relatively constant. The force on the cylinder is approximately double the required force to lift the cylinder and the primary source of this force is from the air muscle actuator. The hand force exhibits a peak of force at the beginning of the lift to activate the air muscle. The force gradually decreases during the lift until the point where the cylinder is released. Before the cylinder is released, a negative peak is present. Since the %MVC stays relatively constant with the hand force fluctuating, it can be concluded that the EMG sensor was receiving signal from both the extensor and flexor muscles. The use of both flexors and extensors throughout the lift indicate that the subject stiffened their arm to control the air muscle rather than simply lifting the cylinder normally.
5.12 - Experiments - Learning Effect

5.12.3 Progressive Adaptation

An adaptation is evident in subject 4 for trials 1 and 2 (Figure 5-15, Figure 5-16). Subject 4’s %MVC begins less than for a normal, unassisted lift; however, unlike many of the other subjects, this subject’s %MVC is progressively lower for each lift of the cylinder. The adaptation is not limited to the %MVC with the grasp force decreasing for each lift. In fact, the hand force approaches approximately zero with the air muscle taking all the force required to lift the cylinder. The air muscle did not fully deflate between lifts and stayed at approximately 20 N. This 20 N offset was due to the subject not fully relaxing between lifts resulting in a non-zero %MVC. Although this anomaly was present in this subject’s results, the most important aspect of the results is that the subject clearly adapted to the air muscle by activating it instead of using his own muscles to lift the cylinder.
Figure 5-15: Assisted Lift Experiment for Subject 4 Trial 1 demonstrating progressive adaptation

Figure 5-16: Assisted Lift Experiment for Subject 4 Trial 2 demonstrating progressive adaptation
6 Conclusions

This study showed a clear effect on both muscle activation levels and grasping force applied by the subjects. However, we did not explicitly test the range of gains possible for a human subject. Moreover, we did not have any means to invasively verify an actual reduction in tendon stress. Nevertheless, since muscle activation is associated with the amount of stress on the tendon, we can surmise that a reduction has indeed occurred. Consequently, we can only draw the conclusion that muscle activation and grasp force were reduced.

We hypothesize that a significant portion of the reduction's variation was due to different adaptation strategies implemented by the subjects. Most subjects had an immediate adaptation response where their %MVC and grasp force both reduced. Subjects classified as having no adaptation strategy did not adapt to the force supplementation system by reducing their %MVC, only reducing their grasp force through co-contraction. In order to provide a lower grasp force with the same %MVC, the subject would have had to stiffen the arm. The stiffening was accomplished by isometric contractions where both the flexor and extensor muscles are contracted. Since the %MVC did not reduce, it is likely that injury would still result.

Finally, a few subjects exhibited a progressive adaptation to the system by reducing both their %MVC and grasp force gradually through the course of the experiments. These three different adaptation strategies of instantaneous reduction, stiffening, and progressive learning indicate that this system could be very useful in studying neuromotor learning and planning strategies.

The supplementation of grasping force in a reliable manner was successful. Subjects were able to reduce their average grasp force by 56%; however, the PID controller was incapable of controlling the supplemented air muscle force with high precision for all subjects. Tuning of the PID controller parameters for each subject could potentially improve these results and the use of
an adaptive method to adjust the PID gains (within stable limits) should be considered. It is important to note that although the PID controller tracked the set point force with less precision than desired, the system never became unstable and was safe for all the subjects, in part due to the air muscle actuator’s compliant nature.

The results of this work support our suggestion that an active orthosis can reduce the %MVC needed to perform selected grasping tasks, and may therefore be a useful strategy to decrease WMSDs through reduction in grasping force. Although not all the subjects responded with the same reduction, the aggregate t-test concluded that the aided lift effort was statistically less than the unaided lift effort for %MVC (p<0.01) with an average reduction of 31%. Moreover, when considered individually, 7 of 9 subjects exhibited less %MVC over the course of the experiment.

An important observation is that the average grasp force decreased by more than the %MVC. This result indicates that the subjects may be overall co-contracting the muscles. Consequently, this system may be shifting the strain from the flexor muscles to the extensor muscles. However, a better assessment of this difference can be obtained with a larger sample size and actual arm mounted orthosis.

From the aggregate results, the proposed actuated orthosis, controlled by an EMG signal, reliably supplemented the grasping force of the hand. Overall, the subject reduced the muscle activation required for a power-grasping task. The subjects steadily handled objects and reduced both their grip force and muscle activation levels with force supplementation. Finally, the air-muscle actuation device provided compliant actuation that could be controlled, safely and intuitively, by the subjects.

Based on the literature review herein, it is apparent that this study is unique in its analysis of the reduction of %MVC during human force supplementation for the purposes of reducing WMSDs. Much other work focuses on correcting paralysis, providing integrated haptic interfaces, or creating man-amplifiers [30, 114]. The closest study for a grasping task, that of Rodriguez [37], involved the reduction of force required to accomplish a grasping task for arthritic patients; however, the system does not use EMG as the control method, but relies on the use of a switch that is turned on and off by the other hand.

Similar work has been performed by Ferris *et al.*, where EMG is used to control an air muscle actuated ankle orthosis. The primary focus of their work is on using the orthotic as a tool for
rehabilitation and studying human locomotor adaptation. Although they did not observe a reduction in all EMG levels such as for the gastocnemius muscle, it is interesting to note that they did observe a reduction in EMG for the tibialis anterior and soleus muscles by 41% and 25% respectively [115]. This reduction could be improved with a different control strategy. Their control strategy uses low-pass filtered EMG to proportionally regulate the pressure within the air muscle. A more effective reduction should be observed if the EMG was used to estimate the biological muscle’s force exertion and using the air muscles to supplement this force.

Our study demonstrates that the concept of using an EMG-modulated compliant actuator to supplement the natural grip is a viable method of reducing the %MVC and grasp force required of workers in a potentially inexpensive manner. Two examples of such work-related tasks are as a checkout clerk at the supermarket, or lifting objects off an assembly line. The air muscle was capable of interacting reliably and safely with a person even with straightforward standard linear control methods.
7 **Recommendations**

Several limitations exist in this study. This study was limited to assessing the %MVC and force reduction for the flexor digitorum muscles, although evidence from Ferris' studies of ankle extensors suggests that similar results could be expected from other muscle groups [115]. Moreover, a small sample size and single session study prohibited a quantitative analysis of a learning effect. Furthermore, although we allowed subjects a period to become acclimated to the setup, we did not explore explicit training strategies. Longer experiments might have aided the two subjects who did not spontaneously reduce their muscle activations or grasping forces by learning different strategies. Additionally, varying between assisted lifts and unassisted lifts could reduce patterns associated with a repetitive lift such as the reduction in %MVC as the subject becomes accustomed to lifting cylinder. Nonetheless, the concept of using EMG signals to control force supplementation of the same muscles these signals activate has been verified.

Another limitation is the experimental setup. The slow response time of the air muscle and the data acquisition system prevented a matching of the force generation from the subjects' biological muscles and the air muscles. Through the use of co-contracting air muscles, the system response will become significantly faster due to the active tension placed on the supplementing hardware. The arm immobilization setup prohibited analysis of more dynamic motions. However, due to the preliminary state of research, this limitation is considered acceptable. Future work in this area using an arm mounted orthotic and performing a series of tasks should be studied. Considerable work would be necessary to develop a practical orthosis which workers at risk would be willing to wear. Once a practical orthosis has been developed, the degree of force supplementation by the air muscle needs to be assessed.
In broader terms, the reduction of %MVC provides a non-military oriented justification for previous military-oriented work, such as that by Kazerooni [30]. For example, the supplementation of soldiers has been focused on the ability to carry larger amounts of equipment; however, this research shows that the supplementation could provide the additional advantage of reducing incidences of musculoskeletal disorders among military personnel, and this work could be extended to workplace applications.

Finally, development of orthoses for different tasks susceptible to WMSDs should be performed. This development would allow analysis of the effect of force supplementation on other muscle groups. Furthermore, these studies would help assess if all muscles require the same degree of supplementation to prevent injury or if they have different requirements that possibly depend on their location, force magnitude, or reaction time. In all ergonomic and orthotic preventative measures, the effect on other muscles groups because of the intervention must be considered to prevent transference of damage.
8 Bibliography


100. Mathworks, MatLab 6.5. 2002, Mathworks.


Appendix A. Experimental Setup Circuitry

A.01 Pressure Transducer Circuitry

The pressure transducer and related circuitry were powered using a 24 V linear power supply. To minimize quantization error, it was desirable to have a 0-5 V output rather than a 1-5 V output using a 250 Ω resistor. The 4-20 mA signal from the transducer was converted to a 0-5 V signal using a precision 4-20 mA current loop receiver integrated circuit (IC) called RCV420 from Texas Instruments/Burr-Brown (TI). The complete datasheet for the RCV420 can be obtained from Texas Instruments (TI) [116]. The RCV420 was operated in a single supply configuration (see application bulletin for RCV420 from TI [116]) to accommodate the dual supply filtering circuitry.

Following the conversion to 0-5 V, the signal was filtered using a TI’s UAF42 [116], which is a generic filter IC. A 2nd-order low-pass Butterworth filter with a cutoff frequency of 90 Hz was implemented to prevent aliasing. As mentioned previously, the UAF42 requires a dual power supply; therefore, a pseudo ground was created for the filter circuitry to allow the RCV420’s single 24 V supply to act as a dual supply. The filter’s ground was referenced to the RCV420’s 10 V precision reference and the positive and negative supplies were referenced to 24 V and ground respectively. This configuration allowed the filter to have asymmetrical supply voltages of +14 V and -10 V. Since the RCV420 will only output in the range of 0 V to 5 V, this configuration is satisfactory and provides the UAF42 with supplies for correct operation. A circuit diagram of the pressure circuitry is shown in Figure A-1.
A.02 Force Transducer Circuitry

The force transducer's conditioning and filtering circuitry used a TMO-1 Amplifier/Conditioner Module from Transducer Techniques and is shown in Figure A-2. The TMO-1 has an input sensitivity of 1 mV for each Volt of excitation voltage, which is adequate for both force sensors. The filter is a low pass 2-pole Butterworth filter with a cutoff frequency of 220 Hz. The direct current (DC) supply required is a 12 V, 65 mA for an output from 0 V to 8 V; however, the output was adjusted by a -3 V offset to provide output from -3 V to 5 V to allow maximum use of the ADC, which has an input range of -5 V to 5V. This configuration provides an input sensitivity of 8 mV.

![Figure A-2: Transducer Techniques TMO-1 Amplifier/Conditioning Module](image)

A.03 Additional EMG Filtering Circuitry

The additional EMG filtering circuitry has four main components: power regulation, input level conversion, filtering, and rectification. The complete circuit diagram is shown in Figure A-3 with a physical implementation shown in Figure A-4 and additional technical data can be obtained from TI [116]. The power regulation was used to reduce the power supplies switching noise from affecting the EMG circuitry. Filtering capacitors were used as well as LM7812 and LM7912 voltage regulators, which regulate the 12 V and the -12 V sources respectively. These voltage regulators are generic and are available from many IC manufacturers.
Figure A-3: Additional EMG Conditioning Circuitry

Figure A-4: Physical Implementation of EMG Conditioning Circuitry and Valve Driver Circuitry
It was found that the filtering circuitry caused unstable filtering of the EMG signal when it rose above 4 V or went below -4 V. Therefore, an input level conversion circuit was built to attenuate the input signal from the range of ±5 V to the range ±4 V. This attenuation is accomplished with a resistor divider with a voltage follower for buffering. The voltage follower used a generic TL082 op-amp IC that could have been accomplished with virtually any op-amp IC.

Since the datasheets for the EMG electrode were not discovered until near completion of the EMG processing board, additional filtering circuitry was designed and built. It was decided that the additional filtering of the EMG signal would have no negative impact on the experimental setup due to large delays in the pneumatic system. The additional filtering circuitry is a 4\textsuperscript{th}-order Butterworth filter with cutoff frequencies of 17 Hz and 450 Hz respectively. Similar to the pressure filtering circuitry, the EMG filtering circuitry uses the TI UAF42 generic filter IC. The EMG filtering required the use of two UAF42s to accomplish the 4\textsuperscript{th}-order Butterworth filtering.

The precision full-wave rectification design is relatively simple and uses only two op-amps and two diodes. This design was selected, because of its simplicity and robustness, from EDN Access Design Ideas [118]. The rectifier does not require matching of diodes or adjusting of resistors while providing a minimal delay time of only one op-amp and one diode from the input to the output. Moreover, the delay is similar for both the negative and positive inputs since both the negative and positive signals pass through the same number of circuit elements. The rectifier was implemented using a TL082 op-amp IC, which contains 2 op-amps within its package, and two generic high-speed diodes. Following the rectifier is a single-pole passive anti-alias filter with a cutoff frequency of

\[
    f_{\text{cutoff}} = \frac{1}{2\pi RC} = \frac{1}{2\pi(4.3\times10^3)(0.1\times10^{-6})} = 370 \text{ Hz}
\]

yielding an absolute minimum sampling frequency of 740 Hz.

**A.04 Valve Driver Design and Circuitry**

The valve driver circuitry uses two Texas Instruments DRV101 PWM Solenoid/Valve Driver ICs. The circuit diagram is shown in Figure A-5 with a physical implementation shown in Figure A-4. Additional technical information can be obtained from TI [116].
To save power, the DRV101 uses a 24 kHz PWM signal during the tension operation of the valve. It is important to clarify that the PWM signal that will control the airflow is not the same as the 24 kHz PWM signal that is used by the DRV101 to save power. The duty cycle adjust pin changes the duty cycle of the 24 kHz PWM signal to the required power for the application. Since the valve requires the same power as a direct current (DC) 5 V to maintain tension operation, the duty cycle adjust pin on the DRV101 is adjusted to produce a duty cycle of 20%. The valve also requires 24 V for speed up operation and the DRV101 can only provide a maximum of 90% duty cycle. To compensate for this deficiency, the DRV101 provides a delay adjust that regulates how much time the DRV101's output is to be at the maximum supply voltage or 100% duty cycle during initial activation. The delay adjust was modified to provide the required 2.5 ms 100% output using a 2.2 nF capacitor.

The driving PWM signal for the valve by the controller was determined using Figure A-6. Since the valve has a minimum open-close cycle of 3.5 ms, a maximum driving frequency of 200 Hz, and the DRV101 has a duty cycle range of 10% to 90% during operation (excluding off and speed up operation), the effective duty cycle range for control is limited. The minimum duty cycle mainly depends on the base frequency of
the driving PWM signal, while the DRV101's range applies the absolute limits to the duty cycle. In Figure A-6, the two dotted red horizontal lines specify the absolute limits for the duty cycle. The angled blue solid line from 10% to 70% specifies the minimum limit for that particular frequency. For example, if 170 Hz is selected, the duty cycle will only have a range of 60% to 90%. Therefore, to maximize the duty cycle range that will allow for better control, a frequency of 30 Hz was selected.

![Duty cycle vs Frequency](image)

**Figure A-6: Minimum duty cycle for a given frequency**

For a 30 Hz PWM driving signal, the amount of time the valve is open varies from 3.5 ms to 30 ms during controlled operation. Controlled operation excludes the off and speed up operation states for which the time will be 0 ms and 33 ms respectively. The amount of time the valve is open for allows the theoretical calculation of the airflow through the valve for a specified duty cycle.
Appendix B. Open Real-Time System Implementation

B.01 Hardware

The controller code is implemented on Loughborough Sound Images’ (LSI’s) PC/C32 board shown in Figure B-1 using a Texas Instruments TMS320C32 floating-point digital signal processor (DSP). Input/output is communicated with the DSP board through a DSP~Link2 interface. The DSP~Link2 is connected to two MFIO-3B DSPLink Motion Control Interface Cards developed by Precision MicroDynamics Inc. The MFIO-3B is a multifunction I/O card for DSP control applications with three 16-bit digital-to-analog converters (DAC); three 12-bit ADC; three quadrature encoder inputs; and three 8-bit ports labeled A, B, and C [119]. Therefore, the two cards provide six DAC channels, six ADC channels, and six quadrature encoder inputs as well as six 8-bit ports.

Three of the ADCs on the first card are used to measure the EMG signal, the squeezing force on the cylinder (measured by a force sensor mounted in the cylinder), and the air muscle pressure. The air muscle force is measured for the experiment using an ADC on the other card. Only one encoder channel is used to measure the current length of the air muscle. Finally, the digital I/O mapped as follows:

- Port A monitors the system shutdown switch and the valve overheating flags.
- Port B outputs the PWM signals for the inlet and outlet of the valve.
- Port C is unused.

All data is logged to a Pentium 120-MHz personal computer (PC).
B.02 The Structure of ORTS

The Open Real-Time System (ORTS) is a general-purpose real-time operating system running on DSP boards that are installed in a host PC. It is developed by the Manufacturing Automation Laboratory (MAL) at the University of British Columbia for control of mechatronic devices, processes, and machine tools [121].

ORTS has two major components: a real-time microkernel (ORTS-DSP) and a PC-based scripting environment (ORTS-PC/NT) [121]. The microkernel runs on the DSP and provides the appropriate control structures, such as semaphores and interprocess communication (including to the PC), for the programmer. The programmer codes processes in ANSI C using ORTS specific data structures and function decomposition.

The programmer’s processes are registered in the kernel’s list of processes to enable compilation into the kernel. Each function must have data structure and an initialization function where data structures are initialized, parameters are set, and links are connected. A link is a generic communication channel developed to allow communication between different processes. All data that is to be retained after the function is complete must be stored inside the function’s global data structure. Local variables are allowed inside the function; however, the variables do not retain their information following the function’s exit. Although ORTS allows static variables to be declared, the static nature of the variables prevents ORTS from using the function as a reentrant function. Use of these variables may cause unexpected results if the process is used more than once as each process with change or read the same variable.

The ORTS-PC/NT scripting environment allows setting up processes using the kernel’s list of processes and linking communication between the processes. The environment allows for easy modification of the program structure using a simple, parsed script file while allowing for customization of the kernel to provide efficient communication and code.

For further information on how to use ORTS to implement a control system see [122], [123], and the [124].

B.03 ORTS Functions – DSP Side

The following is a comprehensive list of all the DSP functions that were programmed for the control system. Several of the functions were also used for testing purposes. If more information is required, consult the source code contained on the CD-ROM under the directory \ORTS_3.0_c32. The main ORTS application was also modified to provide faster data transfers between the DSP and the ORTS application. Before modification, the data was transferred from the DSP to the application. The inefficiencies occurred when the application was saving the file. The application used fprintf statements to convert the binary data into text. However, fprintf is a significantly slow function and therefore the data is now written in raw binary format. Using a binary format caused a major speed improvement. Previously, data could only be
logged at 1 kHz for simple ORTS scripts. Following the improvement, data could be logged at up to 3 kHz or more depending on the script complexity.

The following has been surpassed and is now obsolete: The file format is very simple. The first integer in the file is the number of columns of data per row and following the first element is the data. An external program was created to convert the binary file into a traditional .log file that can be easily loaded into MatLab. The external program is called bin2txt.exe and takes the following command-line parameters:

```
bin2txt inputfile.ext outputfile.ext
```

The program performs no error checking of the input file to make sure that it is in the prescribed format. Caution must be taken that a non-conformant file is not loaded.

(a) **Delay**

```
Delay(DelayLength, NumofTimes) or Delay(DelayLength)
```

This function will generate a delay specified by DelayLength, which is in seconds, and will generate this delay NumofTimes when this group loops. For example,

```
Group EMGCtrl, priority=0, freq=1000:
{
  Delay(10,3)
  Other functions
...}
```

will generate a delay of 10 seconds, execute the other functions, delay for 10 seconds, execute other functions, delay for 10 seconds, execute other functions, and finally it will execute the other functions continuously without delay. If only one parameter is specified, only one delay of specified length will occur.

(b) **Determine Maximum and Minimum Voluntary Contraction and Force**

```
DetermineMVC(starttime, stoptime, N), input=(rmsemg,force)
```

This function determines both the minimum and maximum voluntary contraction and force from the output of the Windowed Root Mean Squared function and the Voltage to Force function. The starttime specifies when the function will begin determining the MVCs and Forces. The stoptime specifies when the function should terminate and allows the maximum and minimum VC determination over a specified period. The function maintains two pairs of arrays of length N containing all the maximum and minimum values read from the rmsemg and force input links over the time period. When the time period is complete, the function takes the average of these elements and prints them to the output window.

(c) **Digital Input/Output to Hexadecimal**

```
DI02Hex();
```

This function reads all of the MFI03B digital input/output ports and displays the results in hexadecimal to ORTS’ output window. It was created to test if ORTS was reading the data placed on the port correctly.

(d) **Emergency Shutdown**

```
EmergencyShutdown(port_number, bitmask), input=(inputlinks), output=(outputlinks)
```

This function reads a digital port specified by port_number and applies the bitmask to the data read. The bitmask is active low (i.e. the corresponding bit must be pulled to ground to trigger an emergency condition. If the resultant bitmask is not the same as the bitmask applied, then an emergency shutdown
has occurred. On an emergency shutdown condition, the outputlinks are set to a zero state otherwise the
inputlinks' data is passed directly to the outputlinks.

(e) **Encoder to Length**

Encoder2Length(initmuslen, lenperpulse), input=(encoder), output=(muslength, deltamuslength)
This function keeps track of the current length of the air muscle. The initial length of the muscle and the
length per encoder pulse are parameters to the function. The input is the output from the
MFI03B_ReadEncoder function and the outputs are the calculated muscle length and the change in
muscle length from the initial length.

(f) **Filter**

Filter(type, function, fc, order, Q, gaindB), input=(insignal), output=(outsignal) or
Filter(type, function, fc, order), input=(insignal), output=(outsignal)
This function will create a filter of the specified type and will filter the input signal using this filter. The
cutoff frequency (fc) is given in Hz and the order can be from 1st to 12th order. The quality factor (Q)
and gain (gain_dB) are only used for the equalizer filter function. The type and function are given
according to Table B-1 where the value corresponds to the type of filter.

<table>
<thead>
<tr>
<th>Type</th>
<th>Function</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>No Filtering</td>
</tr>
<tr>
<td>1</td>
<td>Low Pass Filter</td>
</tr>
<tr>
<td>2</td>
<td>High Pass Filter</td>
</tr>
<tr>
<td>3</td>
<td>Equalizing Filter</td>
</tr>
<tr>
<td>N/A</td>
<td>Equalizer</td>
</tr>
</tbody>
</table>

(g) **Force and Length to Mass**

FL2Mass(b, n, Mw), input=(force, length, temperature), output=(mass)
This function converts force and length to a mass. It takes three parameters, three inputs, and one
output. The parameters b and n specify the values for the thread length (in m) and the number of turns
on a thread for the braided air muscle. The force is the desired force (in N) for the air muscle to
produce. The length is the current length (in m) of the air muscle. Finally, the temperature is the air
muscle temperature (in Kelvin). The output is the mass of air (in g) required to generate the desired
force.

(h) **Impulse**

Impulse(Height, Time), output=(output)
This function generates a pseudo-impulse function to excite all the frequency components of a system.
Height specifies the height of the impulse function and Time specifies when the impulse is to occur. Output is the output link to which the impulse function is written.

(i) **Kill**

Kill(Time, value), input=(inlink), output=(outlink)
The purpose of this function is to change a links output to a static value after a specified length of time. Time specifies the time in seconds that the output is changed to value. Inlink is the input link changed and outlink is the output link that simply passes the Inlink to the output until the specified Time then it passes the value.

(j) **Link Limit**

LinkLimit(lower, higher), input=(inlink), output=(outlink)
This function limits a link to between lower and higher values. Inlink is the link to be limited and outlink is the limited link.

(k) **Mass to Duty Cycle**

Mass2Duty(TankAD, InVFR, InPWM, OutVFR, OutPWM, Mw), input=(mass, pressure, temperature), output=(posduty, negduty)

This function converts the required mass into the duty cycle for the valves. TankAD is the source tank air density (in g/L). InVFR and OutVFR are the input and output volume flow rates (in L/s). InPWM and OutPWM are the input and output pulse periods (in ms). Mw is the molecular weight of the gas (in g/mol) used to pressurize the air muscle. The inputs mass, pressure, and temperature correspond to the mass required (in g), and the pressure (in Pa) and temperature (in Kelvin) in the air muscle. The outputs are the positive and negative duty cycles (in percent), which drive the inlet and outlet valves respectively.

(l) **Percent Maximum Voluntary Contraction to Force**

percentMVC2Force(MaxForce), input=(percentMVC, URDev, FlexExten), output=(Force)

This function converts the percent maximum voluntary contraction (%MVC) to a force using the equation:

\[
\ln(\text{Force}\%) = 1.13 + 0.714 \ln(\% \text{MVC}) + 3.19 \times 10^{-3} \text{FlexExten} + 0.83 \times 10^{-3} \text{URDev} - 26.9 \times 10^{-6} \text{FlexExten}^2 + 15.7 \times 10^{-6} \text{URDev} \text{FlexExten}^2
\]

where \(\ln(\text{Force}\%)\) is the natural logarithm of the relative force; \(\ln(\% \text{MVC})\) is the natural logarithm of relative EMG; \(\text{FlexExten}\) is the relative angle in extension-flexion (%); \(\text{URDev}\) is the relative angle in the ulnar-radial deviation (%). The output from this function is the \(\text{Force}\%\) multiplied by the maximum output force \(\text{(MaxForce)}\) to give the output force (in N).

(m) **Polynomial**

Poly(gain \(_x^n\), gain \(_x^{n-1}\), ..., gain \(_x,\) gain), input=(insig), output=(outsig)

This function takes the input signal and applies the specified polynomial to it. The \(\text{gain} \_x^n\) terms correspond to the gain for the specified order of polynomial. For example, if the polynomial to be applied is \(y = 5x^3 + 6x + 7\), then the form of the poly function will be: Poly(5,0,6,7), input=(x), output=(y). Any order can be specified as long as the DSP has memory; however, the computation will take longer for larger order polynomials.

(n) **Pressure and Length to Force**

PL2F(b,n), input=(pressure,length), output=(force)

This function converts the air muscle's pressure (in Pa) and length (in m) into a force estimate (in N). The parameters b and n specify the values for the length of the braid (in m) and the number of turns on a thread for that particular air muscle.

(o) **Pulse Width Modulation**

pwm(port, bitmask1, bitmask2..., bitmaskN, shutoffport, shutoffbit), input=(freqbitmask1, dutybitmask1, freqbitmask2, dutybitmask2, ..., freqbitmaskN, dutybitmaskN)

This function will output a PWM signal to a port. The bitmasks allow the specification of different frequencies and duty cycles for each bit on the port. The shutoffport and shutoffbit correspond to the port and bit that will be used to shut off the PWM function. The input links correspond to each bit mask in frequency and duty cycle pairs. The ports are labeled as follows A=0, B=1, C=2.
(p) **Pressure Regulation**

Regulate(ADCPort, posport, minmuspress, calibslope, caliboffset), output=(muspress)

This function reads the specified ADC port (ADCPort), converts the result to a muscle pressure using calibslope, and caliboffset. The sensor is assumed as a linear pressure sensor. If the pressure is less than the minimum muscle pressure (minmuspress), the function opens a valve on the port posport until the pressure exceeds the minimum muscle pressure. The function will output the final muscle pressure (muspress) once the pressure exceeds the minimum muscle pressure.

(q) **Root Mean Squared to Duty Cycle**

RMS2Duty(minduty, maxduty, maxrms), input=(rms), output=(duty)

The function converts an RMS signal into a duty cycle signal. The parameters specify the limits for the output duty cycle using minduty and maxduty. The maxrms is the maximum value that the rms signal will experience. It uses this value to normalize the RMS signal between 0 and 100%.

(r) **Root Mean Squared to Percent Maximal Voluntary Contraction (MVC)**

RMS2MVC(minVC, maxVC, window_length), input=(rms), output=(percentMVC)

The RMS to MVC function converts the RMS EMG signal into a percent of maximal voluntary contraction (MVC). The minimum and maximum voluntary contraction are passed as parameters to the function. The maximum is necessary to determine the %MVC and the minimum is used to eliminate low-level noise. The window length is the same window length that the Windowed RMS function uses. If a different value is used, then the function will not operate correctly.

(s) **Route Signal**

RouteSignal(switchon, scalerA, scalerB, limitAlow, limitAhigh, limitBlow, limitBhigh), input=(signal), output=(OutputA, OutputB)

This function will take an incoming signal and will route it to either OutputA or OutputB depending on the value of switchon. If the incoming signal is less than or equal to switchon, then the signal is routed to OutputA otherwise it is routed to OutputB. The variables scalerA and scalerB will multiply the particular output by this scaler. The variables limitAlow, limitBlow, limitAhigh, and limitBhigh will limit each respective output between limitXlow and limitXhigh. When the signal reaches the limit, the limit is output and when the signal again falls inside the limit, the output becomes the signal.

(t) **Sinusoid Generator**

Sinusoid(Amp, Offset, StartTime, Freq), input=(signal), output=(SineSignal) or Sinusoid(Amp, Offset, StartTime, Freq), output=(SineSignal)

The sinusoid function creates a sinusoidal function output. If no input is given, the function will be used as the primary source. If an input signal is given, the sinusoid will be added to the input signal to produce the output. The Amp and Offset are the amplitude of the sinusoid and the DC offset of the sinusoid. The frequency of the sinusoid is given by Freq and the accuracy of the sinusoid depends on the process frequency. The larger the process frequency, the less accurate the sinusoid will become due to process overhead. However, the sinusoid frequency cannot exceed half of the process frequency.

(u) **Spring Force Calculator**

Spring(Kspring, initdeltaL), input=(deltaL, Fapplied), output=(Fout)

The spring force calculator is used to estimate the force created by the spring. Kspring is the spring constant and initdeltaL is the initial change in spring length from rest. The output from this function is given by the equation:

\[ F_{out} = F_{applied} - K_{spring} (\Delta L + \Delta L_{initial}) \]  \hspace{1cm} (47)
(v) **Step Generator**

Step(StepSize, InitOffset, StartTime), input=(signal), output=(stepsignal) or
Step(StepSize, InitOffset, StartTime), output=(stepsignal)

Similar to the sinusoid function, the step generator can either be the starting source or a path in a chain of different sources. StepSize, InitOffset, and StartTime specify the step size, initial offset of the step from DC, and the time the step is supposed to be activated respectively.

(w) **Staircase Generator**

StepCase(StartLevel, EndLevel, NumLevels, Tstart, Tend), input=(signal), output=(Signal) or
StepCase(StartLevel, EndLevel, NumLevels, Tstart, Tend), output=(Signal)

The StepCase function can either be a source or in a chain of other sources. The function will generate a staircase output with evenly spaced steps from the StartLevel to the EndLevel between the time Tstart and Tend. Time is given in seconds.

(x) **Single-Input-Single-Output (SISO) Proportional-Integral-Derivative (PID) Control**

sisoPID(Kp, Ki, Kd), input=(Setpoint, Feedback), output=(Control)

The sisoPID function takes the proportional, integral, and derivative parameters and constructs a digital PID controller based on these three parameters. The inputs are the set point signal and the feedback signal to the controller. The output is the control signal sent to the plant.

The digital PID controller is based on the equation:

\[ u(k) = u(k-1) + k_p e(k) + k_i e(k-1) + k_d e(k-2) \] (48)

where \( u(k) \) is the controller output, \( u(k-1) \) is the previous controller output, \( e(k) \), \( e(k-1) \), and \( e(k-2) \) are the current error signal, the previous error signal, and the error signal from two time steps back respectively. The constants \( k_p \), \( k_i \), \( k_d \) are given by the following equations:

\[ k_1 = k_p + \frac{T_s k_i}{2} + \frac{k_d}{T_s} \] (49)

\[ k_2 = -k_p - \frac{2k_d}{T_s} + \frac{T_s k_i}{2} \] (50)

\[ k_3 = \frac{k_d}{T_s} \] (51)

where \( k_p \), \( k_i \), \( k_d \) are the proportional, integral, and derivative gains respectively. \( T_s \) is the sampling period of the controller.

(y) **Variable**

variable(n0, n1... nN), output=(L0, L1... LN)

This function allows the setting of input links to constant values for testing purposes. Any number and size of output links can be used; however, the number of parameters must equal the total size of all the output links. For example, if \( L0 \) is of size two and \( L1 \) is size three, then the total number of parameters must be five.
(z) **Voltage to Actual**

\[ V_{\text{Actual}}(\text{channel1}, \text{slope1}, \text{channel2}, \text{slope2}, \ldots, \text{channel}, \text{slopeN}), \text{output}=(\text{Actual1}, \text{Actual2}, \ldots, \text{ActualN}) \]

This function assumes a linear sensor attached to the MFIO3B board with a specified channel with a calibrated slope. It will automatically calibrate for the offset in the sensor during the first 2 seconds (for each sensor). It is important to make sure that there is no load on the sensor during this calibration or the offset will be calibrated inaccurately.

(aa) **Voltage to Force (obsolete, surpassed by Voltage to Actual)**

\[ V_{\text{Force}}(\text{slope}, \text{offset}), \text{input}=(\text{voltage}), \text{output}=(\text{force}) \]

The function assumes a linear force sensor with a calibrated slope and offset as parameters and the input is a voltage. The output is the force that the force sensor is measuring. This function is identical to the V2Pressure function.

(bb) **Voltage to Pressure (obsolete, surpassed by Voltage to Actual)**

\[ V_{\text{Pressure}}(\text{slope}, \text{offset}), \text{input}=(\text{voltage}), \text{output}=(\text{pressure}) \]

This function assumes a linear pressure sensor. The calibrated slope and offset are the parameters for the function. The input is the voltage that is read off the sensor and the output is the air pressure the sensor is measuring. This function is identical to the V2Force function.

(cc) **Pseudo-White Noise Generator**

\[ \text{WhiteNoise(MinValue, MaxValue), output=(output)} \]

This function will output pseudorandom white noise to the output link with values between MinValue and MaxValue. This function was adapted from David Langois.

(dd) **Wait for Signal**

\[ \text{WaitForSig(port, bitmask);} \]

This function monitors a bit on a specified port and when the bit is high, the process will pause there until the bit goes low. When the bit is low, the processing resumes.

(ee) **Windowed Root Mean Squared (RMS)**

\[ \text{Windowed\_RMS(window\_length, adcChannel), output=(rms, actualLvl)} \]

This function computes the RMS signal for a given window length on the given adcChannel. The function reads the adcChannel and outputs both the RMS and actual signal levels. When no RMS value is available, zero is output to the RMS output.

Data is logged to file on the PC side with the help of two C++ classes that provide all the management for saving the file structure. The first class is the Trial class which manages each trial within the file and the second class is the Experiment class which manages the whole file.

**B.04 Trial Class**

(a) **Constructors, Destructors, and Operators**

\[ \text{Trial();} \]

This constructor is the default constructor that initializes the Trial object.

\[ \text{~Trial();} \]

This destructor is the default destructor that destroys the Trial object.

\[ \text{Trial(const Trial &t);} \]

The copy constructor which will copy the passed Trial object and create a copy of it. It was implemented to allow the use of the Standard Template Library (STL)'s vector class.
Trial& operator =(const Trial &t);
The assignment operator was also implemented to allow the use of STL’s vector class. It will copy all
the contents from t into a new class.

(b) Accessor Methods

unsigned int GetTrialNum() const;
This method returns the trial number currently assigned to this trial.

float GetDataFreq() const;
This method returns the frequency at which the data was logged.

unsigned int GetDecimation() const;
This method returns the decimation that was applied to the data. The decimation is basically a reduction
in the frequency that the data was logged.

float GetMaxVC() const;
This method returns the Maximum Voluntary Contraction (MVC) for this trial.

float GetMinVC() const;
This method returns the Minimum Voluntary Contraction for this trial.

float GetMaxForce() const;
This method returns the Maximum Force that was measured during this trial.

unsigned int GetNumColsPerRow() const;
This method returns the number of columns the data has in a row.

unsigned long GetNumDataPoints() const;
This method returns the number of data points that were logged in this trial (total number of floats
logged, not the number of rows).

unsigned long GetNumNewDataPoints();
This method returns the number of new data points that were logged to this trial from the last time that
it was saved.

unsigned long GetDataStart() const;
This method returns the starting offset from the beginning of the file for the data.

std::string GetColTitle( unsigned int thisTitle ) const;
This method returns the column title for the column specified by thisTitle.

bool IsModified() const;
This method returns true if the trial has been modified.

unsigned int GetNumTrials() const;
This method returns the total number of trial objects that have been created.

unsigned int GetHeaderSize() const;
This method returns the trial’s header size, which includes everything up to where the data is to begin.

fstream *GetLogDataHandle();
This method returns the trial’s temporary storage file of where the data should be logged. This method
was implemented to increase the speed at which the data can be logged.

std::string *GetLogFileName() const;
This method returns the name of the trial’s temporary storage file.

fstream *GetLoadedFileHandle() const;
This method returns the handle to the file that was loaded into the trial. If no trial has been loaded, then
the method returns NULL.

std::string *GetDataBegTag() const;
This method returns the beginning of the data's tag. The tag must be 8 characters long and the default tag is “DATA_BEG”.

std::string GetDataEndTag() const;
This method returns the ending of the data's tag. The tag must be 8 characters long and the default tag is “DATA_END”.

float GetAverage(unsigned int columnnum);
This method returns the average of the specified column number.

float GetVariance(unsigned int columnnum);
This method returns the variance of the specified column number.

(c) Set Accessor Methods

void SetTrialNum(unsigned int trialnum);
This method sets the current trial number. If this method is not used, then the trial number is automatically determined by the number of trials currently open.

void SetDataFreq(float datafreq);
This method sets the frequency at which the measurements are taken and is logged to file.

void SetDecimation(unsigned int dec);
This method sets the decimation at which the measurements were decimated.

void SetValue(int position, float value);
This method sets an application dependent float value for the trial. Three values can be added per trial. For example, the values could be Maximum Force, Minimum VC, and Maximum VC for the trial.

void SetNumDataPoints(unsigned long numdatapnts);
This method sets the number of data points that were added to the trial.

void SetDataStart(unsigned long databeg);
This method sets the starting point for the data within the file.

(d) Other Utility Functions

void AddColTitle(char *title);
void AddColTitle(std::string title);
These methods add a column title to the trial. Care must be taken not to add too many or too few titles as this class does not check for these situations.

void PrintTrialHeader();
This method is used for debugging purposes. The output displayed is all the data contained in the header of the trial. This function uses cout and should not be used with non-console applications.

void SaveTo(fstream *file);
This method handles the entire file saving operations for the trial. It must be passed an existing open file for the file handle at the position that the trial is to be written.

void LoadFrom(fstream *file);
This method handles the entire file loading operations for the trial. It must be passed an existing open file for the file handle at the position from which the trial is to be loaded.

B.05 Experiment Class

(a) Constructors, Destructors, and Operators

Experiment();
This constructor is the default constructor that initializes the experiment object.

Experiment(char *experimenttype);
Experiment(std::string experimenttype);
    This constructor is used to explicitly specify the experiment type for the experiment. The identifier must
    be six characters long.

Experiment(char *experimenttype, unsigned int subnum);
Experiment(std::string experimenttype, unsigned int subnum);
    This constructor is used to explicitly specify both the experiment type and the subject number. If no
    subject number is specified, then the number will assume 1.

Experiment(const Experiment &exper);
    The copy constructor will duplicate the experiment passed to it. This constructor was implemented to
    allow use with the STL vector class.

Experiment& operator= (const Experiment &exper);
    The assignement operator will assign an experiment to another experiment. This operator was
    implemented to allow use with the STL vector class.

~Experiment();
    This destructor is the default destructor and will free all files and memory.

(b) **Get Accessor Functions**

unsigned int GetNumExperiments() const;
    This method returns the number of experiment objects that have been created.

std::string *GetExperimentType() const;
    This method returns the experiment type. It is six characters long and the default value is "DEFAULT".

unsigned int GetSubjectNum() const;
    This method returns the subject number.

std::string *GetDate() const;
    This method returns the date that the experiment took place. It is assumed that all the trials in this file
    will be from the same date.

std::string *GetFileName() const;
    This method returns the filename used to which the data is logged.

std::string *GetTrialTag() const;
    This method returns the tag identifier where the trial is supposed to begin. It is nine characters long and
    the default is "TRIAL_TAG".

unsigned int GetNumTrials() const;
    This method returns the number of trials that have taken place during this experiment.

Trial *GetTrial(unsigned int trialnum);
    This method returns the trial object specified by trialnum.

(c) **Set Accessor Methods**

void SetExperimentType(std::string extype);
    This method sets the experiment type tag. It must be six characters long. If it is longer, the identifier
    will be truncated.

void SetSubjectNum(unsigned int subnum);
    This method sets a different subject number. It is recommended that a different subject number is used
    for each experiment.

void SetDate(std::string datetosetto);
    This method explicitly sets the date that the experiment is taking place. If it is not specified, the
    experiment object will determine the correct date and time. The date must be 24 characters long. If it is
    longer, it will be truncated to 24 characters.
void SetTrialTag(std::string tag);

This method sets the trial tag identifier. It must be 9 characters long and the default is "TRIAL_TAG".

(d) Other Utility Methods

void LoadFrom(char *filename);
void LoadFrom(std::string filename);

These methods load experimental objects from the file specified by filename. The trial objects are loaded automatically; however, the trial data is not loaded to memory to save on memory requirements.

void SaveTo(char *filename);
void SaveTo(std::string filename);

These methods save experimental objects to the file specified by filename. The trial objects are saved automatically, including the trials’ data.

void PrintMVCHeader();

This method is used for debugging purposes to verify the experiment object’s header.

void AddTrial(float freq, unsigned int dec, vector <std::string> coltitles);

This method is used to add a trial to the experiment using the frequency freq, the decimation dec, and a vector of column titles specified by coltitles.

void AddTrial(Trial t);

This method adds a premade trial to the experiment. This method allows the creation of a trial separate from the experiment and later added to the experiment.

B.06 ORTS Functions – PC Side

(a) Advanced LED Toggle

AdvLEDToggle(onlength, offlength, starttime, stoptime)

This function is a modification of the LEDToggle function. It allows independent control of how long the LED is on and off, which allows the LED to be on a different amount of time than it is off. The parameters onlength and offlength indicate how long the LED should be on and off respectively. The starttime and stoptime parameters specify when the LED toggling should begin and stop respectively.

(b) LED Toggle

LEDToggle(period)

This function will use the PC Speaker to toggle a light emitting diode (LED). The toggling of the LED turns the LED on and off for equal amounts of time each.

(c) Load Configuration From Disk

LoadConfigFromDisk("Filename", output=(link1,...,linkN))

This function loads floats from the file "filename" and outputs the values to links 1 to N following loading of the floats. The function takes into account the number and size of each of the links; however, the function requires the appropriate number of floats inside the configuration file. For example, two links with sizes five and eight would require 13 floats in the configuration file. The first five floats in the file would be output to the first link and the last eight floats would be output to the second link.

(d) Log MVC to Disk

LogMVC2Disk("Filename", Subjnum, Freq, Decimation), input=(link1);

This function will transfer the data transferred from the DSP via link1 and save it to the file named FileName. The subject number, frequency, and decimation are used to generate the experiment file in the appropriate format.
(e) **Log Experiment to Disk**

LogExper2Disk("FileName", Subjnum, Freq, Decimation),input=(link1);

This function is identical to the LogMVC2Disk function; however, it uses a different file identifier.

(f) **Periodic Beep**

PeriodicBeep(period);

This function will output an alternating high/low periodic beep through the PC speaker. The *period* is in seconds. This function was replaced with the LED Toggling functions for the purposes of the experiment.
Appendix C. Detailed Experimental Procedure

C.01 Experimental Setup Checklist

<table>
<thead>
<tr>
<th>Item</th>
<th>Done?</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gas on</td>
<td></td>
</tr>
<tr>
<td>Oscilloscope on &amp; setup</td>
<td></td>
</tr>
<tr>
<td>Fuzzy running</td>
<td></td>
</tr>
<tr>
<td>Function generator on</td>
<td></td>
</tr>
<tr>
<td>Directory structure setup</td>
<td></td>
</tr>
<tr>
<td>Ethics form signed</td>
<td></td>
</tr>
<tr>
<td>Wrist brace on</td>
<td></td>
</tr>
<tr>
<td>Find flexor digitorum profundus muscle &amp; place sensor</td>
<td></td>
</tr>
<tr>
<td>Adjust emg gain</td>
<td></td>
</tr>
<tr>
<td>Position wrist brace markers on setup</td>
<td></td>
</tr>
<tr>
<td>Find appropriate sized thumb spacer</td>
<td></td>
</tr>
<tr>
<td>Run DetMVC script and wait for calibration complete</td>
<td></td>
</tr>
<tr>
<td>Ask subject to grasp cylinder as hard as possible for 10 sec (repeat 3 times, with 2 min between each)</td>
<td></td>
</tr>
</tbody>
</table>

C.02 Pre-Experimental Setup

Open gas source valve and regulator valve.
Start the oscilloscope.
Enable storage on the oscilloscope by pressing the storage button below the screen. Change the time scale to 0.5 s and the voltage scale to 2 Volts per division.
Start the “Fuzzy” computer and login. Double click the ORTS_Greg shortcut on the desktop to open the modified ORTS.
Start the BK Precision function generator and adjust to a 30 Hz square wave between 0 and 5 V. The generator is used for deflating the air muscle in case of an emergency. To deflate the muscle, close ORTS, and touch the probe on the function generator to the pin marked Duty on the short side of the EMG circuit board.
On the “Fuzzy” computer, create a folder called “Subject <subjno>” where <subjno> is the current subject number.
In ORTS, open the scripts DetMVC.spt, Exper01.spt, Exper02, and Exper03.spt script and change the path for the SaveToDisk functions to the folder “Subject <subjno>” given in step 6.

C.03 Common Experimental Setup

Ask the subject to read and sign the Ethics form and if he\(^7\) has any questions about the experiments.
Ask the subject to sit on the stool by the experimental setup.
Ask the subject to put the wrist brace on his arm and ask if he requires any assistance.
Ask the subject to position his hand as if he were going to pick up the cylinder using only his index, middle, and ring fingers (as well as thumb).
Turn the gain on the EMG sensor board to its maximum position.
Ask the subject to grasp the cylinder with a moderate grip force.
Locate the stiff muscle bulge on the underside of the subject’s arm and place the EMG sensor slightly above the center of the bulge. This position should correspond to the flexor digitorum profundus. If there is trouble identifying the correct muscle, consult the “Electrode Placement in EMG Biofeedback” book.

\(^7\) For this section, he, him, and his can also refer to she, her, and hers. The masculine form is used for ease of reading and is not intended to be sexist.
Monitor the EMG on the oscilloscope while asking him to grasp the cylinder with a moderate grip force when necessary.

a) If the EMG sensor is saturated, then decrease the gain on the EMG board appropriately.
b) If the EMG signal is too weak at maximum gain, apply a small amount of EMG electrode gel to the sensor.

Once a repeatable and stable location is identified, ask the subject to hold the sensor to his arm in that location.

Tape the sensor to the subject’s arm using medical tape.

Position the markings on the wrist brace to the appropriate position on the experimental setup.

Position the upper arm strap to slightly below his elbow.

Identify which combination of thumb spacers is appropriate to maintain a comfortable thumb-to-wrist angle and place that combination behind the subject’s thumb on the Velcro strip.

Strap the subject’s arm into the setup using the straps. Watch for pinching his with the buckles! Ensure the subject is unable to move his arm and ask him if he has any significant discomfort.

Ask the subject to grasp the cylinder with a grip force as large as possible.

a) If the EMG signal is saturating at the person’s maximum, decrease the gain until the sensor no longer saturates.
b) If the EMG signal does not reach +5 V, increase the gain.
c) Repeat step 16 until the signal no longer saturates and maximum signal span is used.

Ask the subject to place his fingers on a marking points on the cylinder but to not to exert any force on it at all. Start ORTS and run the script DetMVC.spt. Wait until calibration is complete and record calibration factors.

Instruct the subject to grasp the cylinder as hard as possible for 10 seconds.

Perform steps 17 through 19 three times with 2-minute breaks in between each time. Record the minimum and maximum voluntary contraction (VC) and the minimum and maximum force values for each time.

Take the 3 minimum VCs and average them together. Take the 3 maximum VCs and average them together. Take the 3 maximum forces and average them together. Take the 3 minimum forces and average them together.

Open the Exper01.spt and the Exper02.spt ORTS scripts. In both files, find the function RMS2MVC and change the values for the minimum and maximum VC to the averages taken in step 19. Find the function percentMVC2Force and change the value for maximum force to the value taken in step 19.

Measurements to be taken are: EMG and Grasp force logged to file, the max and min VCs and forces, calibration data.

**C.04 Experiment 1 Setup – Base lining Subject**

Again, instruct the subject to place his fingers on the marking points on the cylinder.

Inform the subject that the platform below the cylinder will be removed and that he is supposed to hold the cylinder with as little force as possible.

Open the script Exper01.spt and change the filename where the data will be saved to: Exper01S<subjectno> T<trialno>.raw where <subjectno> and <trialno> are the subject number and the trial number respectively.

Instruct the subject to hold the cylinder with no grip force during calibration.

Instruct the subject to lift the cylinder on a low to high pitch sound and put it down on a high to low pitch. While using the least amount of force possible without dropping the cylinder.

Start the script Exper01.spt and wait until calibration is complete.

Once calibration is complete, the system will begin to emit alternating tones every 10 seconds for ___ minutes.

Perform steps 6 and 7 three times taking a 1-minute rest period between trials.

Measurements to be taken are: EMG and Grasp force logged to file, calibration data.

**C.05 Experiment 2 Setup**

Adjust the bar’s height to make the strap line up with the subject’s fingers.

Ask the subject to place his fingers into the elastic holders on the strap.

Make sure the subject’s fingertips are not obstructed by the elastic straps.

Connect the air muscle to the bar.
Ask the subject to hold the cylinder in the same manner as in Experiment 1.
Make sure that there is no force exerted on the subject’s hand by the strap from the air muscle.
Measure the air muscle’s length and record it. Measure the spring’s initial displacement and record it.
Open the ORTS script Exper02.spt and change the filename where the data will be saved to:
Exper01S<subjectno>T<trialno>.raw where <subjectno> and <trialno> are the subject and trial numbers respectively.
Find the function Encoder2Length and change the length of the air muscle to the value recorded in step 7.
Find the function Spring and change the initial offset to the value recorded in step 7.
Instruct the subject to not exert any force on the cylinder.
Start the script Exper02.spt and wait until calibration is complete. Record the calibration data.
Wait until the air muscle system is complete its initialization and the system is operating as expected (subject squeezes, air muscle squeezes).
Again, instruct the subject to lift the cylinder so the platform can be removed.
Make the subject hold the cylinder for 25 seconds and then replace the platform.
Repeat steps 8 to 15 three times.
Measurements to be taken are: EMG, air muscle force, grasp force, air muscle length, estimated air muscle force, setpoint force, pressure are all logged to file, calibration data.
Appendix D. MVC Calibration Plots

Figure D-1: MVC Calibrations for Subject 1
Figure D-2: MVC Calibrations for Subject 2

MVC for Subject 2 Trial 1

MVC for Subject 2 Trial 2

MVC for Subject 3 Trial 1
Figure D-3: MVC Calibrations for Subject 3
Figure D-4: MVC Calibrations for Subject 4
Figure D-5: MVC Calibrations for Subject 5
Figure D-6: MVC Calibrations for Subject 6
Figure D-7: MVC Calibrations for Subject 7
Figure D-8: MVC Calibrations for Subject 8
Figure D-9: MVC Calibrations for Subject 9
Appendix E. Control Experiment Plots

Experiment 1 for Subject 1 Trial 1

Experiment 1 for Subject 1 Trial 2
Figure E-1: Control Experiments for Subject 1
Figure E-2: Control Experiments for Subject 2
Figure E-3: Control Experiments for Subject 3
Figure E-4: Control Experiments for Subject 4
Figure E-5: Control Experiments for Subject 5
Figure E-6: Control Experiments for Subject 6
Figure E-7: Control Experiments for Subject 7
Appendix E - Control Experiment Plots

Figure E-8: Control Experiments for Subject 8
Figure E-9: Control Experiments for Subject 9
Appendix F. Assisted Lift Experiment Plots

Experiment 2 for Subject 1 Trial 1

Experiment 2 for Subject 1 Trial 2
Figure F-1: Assisted Experiments for Subject 1
Appendix F - Assisted Lift Experiment Plots

Experiment 2 for Subject 2 Trial 3

- Percent MVC
- Shoulder Force (N)
- Can Force (N)
- Arm Muscle Force (N)
- Hand Force (N)

Experiment 2 for Subject 2 Trial 4

- Percent MVC
- Shoulder Force (N)
- Can Force (N)
- Arm Muscle Force (N)
- Hand Force (N)
Appendix F - Assisted Lift Experiment Plots

Figure F-2: Assisted Experiments for Subject 2
Experiment 2 for Subject 3 Trial 1

Experiment 2 for Subject 3 Trial 2
Figure F-3: Assisted Experiments for Subject 3
Figure F-4: Assisted Experiments for Subject 4
Figure F-5: Assisted Experiments for Subject 5
Appendix F - Assisted Lift Experiment Plots

Figure F-6: Assisted Experiments for Subject 6
Appendix F - Assisted Lift Experiment Plots

Experiment 2 for Subject 7 Trial 2

Experiment 2 for Subject 7 Trial 3
Figure F-7: Assisted Experiments for Subject 7

Experiment 2 for Subject 8 Trial 1

Figure F-7: Assisted Experiments for Subject 7

Experiment 2 for Subject 8 Trial 1
Figure F-8: Assisted Experiments for Subject 8
Figure F-9: Assisted Experiments for Subject 9
Appendix G. Ethics Form

THE UNIVERSITY OF BRITISH COLUMBIA
Department Of Mechanical Engineering
(will be printed on letterhead paper)

INFORMED CONSENT FORM
August 25, 2003

Title of Study:
"Electromyogram Control of an Air Muscle Actuated Orthosis to Prevent Repetitive Strain Injuries"

Researchers:
Student: Gregory Forrest
Supervising Professors: Dr. Antony Hodgson, Dr. Elizabeth Croft

Name of Participant: (Please print) ____________________________

I understand that this study, in which I have agreed to participate, will involve measurement of EMG signals while picking up a 1.5 kg object. Electromyogram signals are electrical impulses that are sent from your brain to your muscles and are detected through a sensor placed on the skin. The experiment will be performed with and without an assistive device to determine if the assistive device will reduce factors that lead to repetitive strain disorder. There are no potential harms for me in this experiment.

I understand that participants must be in the age range of 18-60, have normal hand function, and no history of hand disorders that affect muscles, tendons, or nerves.

I understand that my participation in this study is voluntary and that I may withdraw from the study at any time and for any reason without penalty.

I understand that there will be no payment for my participation.

I understand that there is no obligation to answer any question/participate in any aspect of this project that I consider invasive, offensive or inappropriate.

I understand that all personal data will be kept strictly confidential and that all information will be coded so that my name is not associated with my answers. I understand that only the researchers named above will have access to the data.

Participant’s Signature ____________________________
Date ____________________________

If you have any questions or concerns about your participation in the study, you may contact Greg Forrest at (604) 822-3147, Dr. Antony Hodgson or Dr. Elizabeth Croft, Professors at Mechanical Engineering Department, UBC.

Questions or concerns regarding involvement in a research study may be directed to Dr. R. D. Spratley, Director, Office of Research Services, at (604) 822-8595

Any written explanation will be provided for you upon request.
Thank you for your help! Please take one copy of this form with you for further reference.

***
I have fully explained the procedures of this study to the above volunteer.

Researcher Signature ____________________________
Date ____________________________
Witness Signature ____________________________
Date ____________________________
Appendix H. ORTS Scripts

H.01 DetMVC.spt

DSP C32:

Link out(dsp2pc, buffered, 100, 3);
Group DetMVC, priority=0, freq=1000:

{ */ EMG Biomech Links */
    Link raw_emg(1);
    Link filt_emg(1);
    Link windowed_emg(1);
    Link Force(1);
    Link Fvolts(1);
    WaitForSig(1,0x04);
    /* Initialize variables */
    V2Actual(2,29.344), output=(Force); // offset=1.6174
    MFIO_ReadADC(1), output=(raw_emg);
    /* Filter(type, function, fc, order) */
    Filter(1,1,7,4), input=(raw_emg), output=(filt_emg);
    /* Windowed_RMS(window_length_in_ms) */
    Windowed_RMS(100), input=(filt_emg), output=(windowed_emg);
    /* DetermineMVC( starttime, stoptime, numsamples ) */
    DetermineMVC(0,32,100), input=(windowed_emg, Force);
    log(1), input=(windowed_emg, Force), output=(out);
}

PC:

PriorityClass=High;
AdvLEDToggle(2,6,0,32);
/* LogMVC2Disk(filename,subjectnumber,frequency,decimation,numberofcolumns,column names ... ) */
LogMVC2Disk("Sub09MVCT03.raw",1,1000,1,3,"Index","WinEMG","GripForce"), input=(out);
H.02 Exper1.spt (Control Experiment)

DSP C32:
/*********************/
Link dataout(dsp2pc,buffered,100,5);
/*********************/
Group EMGCtrl, priority=0, freq=1000:
{
 /* Link Declaration */
/*****************************/
    /* EMG Biomech Links */
    Link raw_emg(1);
    Link filt_emg(1);
    Link windowed_emg(1);
    Link vc(1);
    Link URDev(1);
    Link FlexExten(1);

    /* Calculated Force Links */
    Link Fsp(1);

    /* Measured Force Links */
    Link Ftotal(1);
    Link Fair(1);
    Link Pmeasured(1);

    /* Variable Declaration & Initialization */
    /*************************/
        /* Initialize variables */
        Variable(0,0),output=(URDev,FlexExten);
        /* Generate a delay so I can get setup */
        /* WaitForSig(port,bit) */
        WaitForSig(1,0x04);

    /* Process EMG */
    /***********/
Appendix H - ORTS Scripts

MFIO_ReadADC(1),output=(raw_emg);
/* Filter(type, function, fc, order) */
/* Type = 0=FLAT_FILTER, 1=BUTTERWORTH, 2=BESSEL, 3=LINKWITZ_RILEY, 4=EQUALIZER */
/* Function = 0=NO_FILTERING, 1=LPF, 2=HPF, 3=EQ */
Filter(1,1,7,4),input=(raw_emg),output=(filt_emg);

/* Windowed_RMS(window_length_in_ms) */
Windowed_RMS(200),input=(filt_emg),output=(windowed_emg);

/* RMS2MVC(MinVC,MaxVC,window_length) */
RMS2MVC(0.85,4.97,200),input=(windowed_emg),output=(vc);

/* percentMVC2Force(MaxForce), input=(percentMVC,URDev,FlexExten),output=(Force); */
percentMVC2Force(93),input=(vc,URDev,FlexExten),output=(Fsp);

/* Instrument Measurement Section */
/*-----------------------------*/

// V2Actual(channel,slope,...,channel,slope)
V2Actual(2,29.344,3,120300,4,-19.611),output=(Ftotal,Pmeasured,Fair);

/* Logging Data Section */
/*----------------------*/

log(1),input=(raw_emg,windowed_emg,vc,Ftotal),output=(dataout);
}

PC:
/*---------------------*/
PriorityClass=High;
AdvLEDToggle(2,6,6,6);
/* LogExper2Disk(filename,subjectnumber,frequency,decimation,numberofcolumns,column names ... ) */
LogExper2Disk("Sub09Exper1T06.raw",1,1000,1,5,"Index","RawEMG","WinEMG","VC","Fcan"),
input=(dataout);
Appendix H - ORTS Scripts

H.03 EMGCtrl.spt (Assisted Experiment)

DSP C32:

/*****************************************************************************/
Link out(dsp2pc,buffered,100,9);
/*****************************************************************************/
Group EMGCtrl, priority=0, freq=1000:
{

/] Link Declaration */
/*****************************************************************************/

/] EMG Biomech Links */

Link raw_emg(1);
Link filt_emg(1);
Link windowed_emg(1);
Link vc(1);
Link URDev(1);
Link FlexExten(1);

/] Calculated Force Links */

Link Fsp(1);
Link Fsilence(1);
Link Fsafesilence(1);
Link Fspkill(1);
Link Fsafe(1);
Link estForce(1);
Link Fspr(1);
Link adjForce(1);
Link modForce(1);
Link modForce2(1);
Link Fcontrol(1);
Link Fcontrolsilence(1);

/* Measured Force Links */
Link Fcan(1);
Link Fmus(1);

/* Air Muscle and Valve Parameters */
Link PWMFreq(1);
Link Pmeasured(1);
Link encoder(1);
Link Lmeasured(1);
Link deltaL(1);
Link InletDury(1);
Link OutletDuty(1);
Link dutysetvolt(2);

/* Variable Declaration & Initialization */

/* Initialize variables */
Variable(0,0),output=(URDev,FlexExten);
Variable(30,3.4,3.4),output=(PWMFreq,dutysetvolt);
Variable(0.32),output=(Lmeasured);

/* Set the voltage that determines the dutycycle */
MFIO_WritDAC(1,2), input=(dutysetvolt);

/* WaitForSig(port, bit) */
// Delay(10);
// WaitForSig(1, 0x04);

/* Process EMG */
/* ---------------- */

MFIO_ReadADC(1), output=(raw_emg);

/* Filter(type, function, fc, order) */
/* Type = 0=FLAT_FILTER, 1=BUTTERWORTH, 2=BESSEL, 3=LINKWITZ_RILEY, 4=EQUALIZER */
/* Function = 0=NO_FILTERING, 1=LPF, 2=HPF, 3=EQ */
Filter(1, 1, 7, 4), input=(raw_emg), output=(filt_emg);

/* Windowed_RMS(window_length_in_ms) */
Windowed_RMS(200), input=(filt_emg), output=(windowed_emg);

/* RMS2MVC(MinVC, MaxVC, window_length) */
RMS2MVC(0.85, 4.9, 200), input=(windowed_emg), output=(vc);

/* percentMVC2Force(MaxForce), input=(percentMVC, URDev, FlexExten), output=(Force); */
percentMVC2Force(93), input=(vc, URDev, FlexExten), output=(Fsp);

/* Safety Section */
/* ---------------- */
/** Instrument Measurement Section **/ 

// V2Actual(channel,slope,...,channel,slope)  
V2Actual(2,29.344,3,1203000,4,-19.611),output=(Fcan,Pmeasured,Fmus);

// MFIO_ReadEncoder(l), output=(encoder);  

// Encoder2Length(initmuslen,lenpulse),input=(encoder),output=(muslength,deltalength)  
Encoder2Length(0.32,0.133e-3),input=(encoder),output=(Lmeasured,deltaL);

/* Setpoint Force Determining Section */

// PL2F(b,n),input=(Pressure,Length),output=(Force)  
PL2F(0.4231,7.167),input=(Pmeasured,Lmeasured),output=(adjForce); //estForce);

// Spring(Kspring, initdeltaL),input=(deltaL,Fapplied),output=(Fout)  
Fout = Fapplied - Kspring*(deltalength+initdeltaL)  
Spring(73.71,0.04),input=(deltaL,estForce),output=(Fspr);

// Step(finalvalue,initoffset,starttime)  
Step(20,0,25),input=(Fspr),output=(adjForce);
// Poly(x^n gain, x^(n-1) gain... x^1 gain, x^0 gain)
Poly(-1.3399e-4,2.4102e-2,-3.8705e-1,-1.2212), input=(adjForce), output=(modForce);  // Polynomial for Est force to Measured Force
// Poly(5.188135e-6,-1.85445e-3,1.83149,-1.2212), input=(adjForce), output=(modForce);
// Before S01T08, this was at 10, causing major delay
Poly(1,0), input=(modForce), output=(modForce2);

/* Control & PWM Section */
/*---------------------------------*/

SilenceLinkDuring(1,0x04), input=(Fsafe), output=(Fsafesilence);
SilenceLinkDuring(1,0x04), input=(modForce2), output=(Fsilence);

// sisoPID(0.6,0.833,0), input=(Fsafesilence,Fsilence), output=(Fcontrol);
// sisoPID(0.6,0.833,0.01), input=(Fsafesilence,Fsilence), output=(Fcontrol);

// S01T01  sisoPID(0.6,0.55,0.01), input=(Fsafesilence,Fsilence), output=(Fcontrol);
// S01T02-S01T04 sisoPID(0.6,0.55,0.01), input=(Fsafesilence,Fsilence), output=(Fcontrol);
// S01T05  sisoPID(0.6,0.75,0.01), input=(Fsafesilence,Fsilence), output=(Fcontrol);
// S01T06  sisoPID(0.75,0.75,0.01), input=(Fsafesilence,Fsilence), output=(Fcontrol);
// S01T07  sisoPID(0.75,0.75,0.02), input=(Fsafesilence,Fsilence), output=(Fcontrol);
// S01T08+
// sisoPID(0.833,0.75,0.01), input=(Fsafesilence,Fsilence), output=(Fcontrol);

SilenceLinkDuring(1,0x04), input=(Fcontrol), output=(Fcontrolsilence);

// Sustained oscillations at a (P,I,D) of (X,X,X)
// Tuned with Ziegler-Nichols PI oscillations of X s
/ RouteSignal(switchon, scalerA, scalerB, limitAlow, limitAhigh, limitBlow, limitBhigh), input=(signal), output=(lessthan, greaterthan)

RouteSignal(0,-1,1,0,90,90), input=(Fcontrolsilence), output=(Outletduty, Inletduty);

/* Logging Data Section */
/*--------------------------------*/

/* For debugging purposes*/
log(1), input=(vc,Fsp,Fcan,Fmus,Fcontrol,Pmeasured,raw_emg,windowed_emg), output=(out);
}

PC:
/*--------------------------------*/
PriorityClass=High;
// S01T01&S01T02 AdvLEDToggle(2,6,6,66);
// S01T03+
AdvLEDToggle(6,8,6,112);
LogExper2Disk("Sub09Exper2T05.raw",1,1000,1,9,"Time","VC","Fsp","Fcan","Fmus","Fcontrol","Pmeasured","RawEMG","WinEMG"), input=(out);