Dynamic Control for a Pneumatic Muscle Actuator to Achieve Isokinetic Muscle Strengthening

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DYNAMIC CONTROL FOR A PNEUMATIC MUSCLE ACTUATOR TO ACHIEVE ISOKINETIC MUSCLE STRENGTHENING

A dissertation submitted in partial fulfillment of the requirements for the degree of Doctor of Philosophy

By

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ABSTRACT

Hall, Kara Lynn, Ph.D., Department of Biomedical, Industrial, and Human Factors Engineering, Ph.D. in Engineering Program, Wright State University, 2011. Dynamic Control for a Pneumatic Muscle Actuator to Achieve Isokinetic Muscle Strengthening.

A pneumatic muscle actuator (PMA) is a device that mimics behavior of skeletal muscle by contracting and generating force in a nonlinear manner when activated. PMAs have a high power to weight ratio and possess unique characteristics which make them ideal for human interaction. Due to their nonlinear dynamics, PMAs are difficult to control, presenting challenges in system implementation. Despite these challenges, PMAs have great potential as a source of resistance for strength training and rehabilitation. The main goal of this work was to control a PMA for use in isokinetic exercise, potentially benefiting anyone in need of optimal strength training through a joint’s range of motion. This includes astronauts who need to counteract muscle atrophy and bone loss during prolonged spaceflight. The lightweight PMA driven by pressurized air does not need gravity to produce resistance, making it an attractive option for a microgravity exercise device.

The control system developed is based on an inverse three-element phenomenological model and adaptive nonlinear control. The system operates as a type of haptic controller, automatically adjusting resistance to assist a simulated neuromuscular actuator in maintaining the desired velocity. A human quadriceps dynamic simulator (HQDS) was developed so that control effectiveness and accommodation could be tested prior to human implementation. A motor, which produces torque analogous to quadriceps’ torque production about the knee, is used in conjunction with the HQDS to simulate neuromuscular actuation. Tracking error results for motor shaft position (simulated joint angle), velocity (simulated lower leg angular velocity), and PMA displacement indicate that the control system is effective at producing PMA displacement and
resistance necessary for a scaled, simulated neuromuscular actuator to maintain low-velocity
isokinetic movement during simulated concentric and eccentric knee extension. This work is an
important step towards human implementation of PMA produced resistance for isokinetic
strength training and rehabilitation.
# TABLE OF CONTENTS

1. INTRODUCTION AND BACKGROUND................................................................. 1
   1.1 Problem and Research Goals........................................................................... 1
   1.2 Background and Review of Literature ............................................................ 4
      1.2.1 Pneumatic Muscle Background ................................................................. 4
         1.2.1.1 Description of Pneumatic Muscle......................................................... 4
         1.2.1.2 Pneumatic Muscle Operation Methods................................................. 6
         1.2.1.3 History of Pneumatic Muscle .............................................................. 7
      1.2.2 Applications of Pneumatic Muscle .......................................................... 8
         1.2.2.1 Robotics ............................................................................................... 8
         1.2.2.2 Power and Mobility Assistance Devices .............................................. 11
         1.2.2.3 Rehabilitation Devices ....................................................................... 13
         1.2.2.4 Technological Devices with Human Interaction ................................. 15
         1.2.2.5 Prosthetics and Artificial Body Parts ................................................. 15
         1.2.2.6 Industrial Applications ....................................................................... 16
      1.2.3 Types of Pneumatic Muscle ...................................................................... 16
         1.2.3.1 Braided/McKibben Pneumatic Muscle ............................................... 16
         1.2.3.2 Straight Fiber Pneumatic Muscle ......................................................... 17
         1.2.3.3 Pleated Pneumatic Muscle .................................................................. 18
         1.2.3.4 Curved Pneumatic Muscle .................................................................. 19
         1.2.3.5 Pneumatic Muscle Using Alternative Materials ................................. 19
         1.2.3.6 Hybrid Devices ................................................................................... 20
# TABLE OF CONTENTS (continued)

1.2.3.7 Commercially Available Pneumatic Muscle ........................................... 21

1.2.4 Modeling of Pneumatic Muscle ........................................................................ 22
  1.2.4.1 Geometric Model of Pneumatic Muscle ................................................... 23
  1.2.4.2 Biomimetic Model of Pneumatic Muscle .................................................. 30
  1.2.4.3 Phenomenological Model of Pneumatic Muscle ....................................... 31
  1.2.4.4 Other Models of Pneumatic Muscle ......................................................... 32

1.2.5 Control of Pneumatic Muscle Systems ............................................................... 33
  1.2.5.1 Open Loop Control of Pneumatic Muscle Systems ................................. 34
  1.2.5.2 P, I, and D Control for Pneumatic Muscle Systems ............................. 34
  1.2.5.3 Fuzzy Control Applied to Pneumatic Muscle Systems ......................... 37
  1.2.5.4 Learning/Adaptive Control of Pneumatic Muscle Systems .................... 39
  1.2.5.5 Neural Networks Applied to Pneumatic Muscle Systems .................... 40
  1.2.5.6 Variable Structure Control of Pneumatic Muscle Systems .................... 42
  1.2.5.7 $H_\infty$ Control of Pneumatic Muscle Systems .................................... 44
  1.2.5.8 Gain Scheduling Applied to Pneumatic Muscle Systems ...................... 45
  1.2.5.9 Feedback Linearization Applied to Pneumatic Muscle Systems .......... 46
  1.2.5.10 Pole Placement Control of Pneumatic Muscle Systems ....................... 46
  1.2.5.11 Impedance Control of Pneumatic Muscle Systems ............................ 47
  1.2.5.12 Model Predictive Control of Pneumatic Muscle Systems .................... 47
  1.2.5.13 Proportional Myoelectric Control of Pneumatic Muscle Systems ........ 48
  1.2.5.14 Biomimetic- Skeletal Muscle Control of Pneumatic Muscle Systems .... 48
  1.2.5.15 Biomimetic- Cerebellar Control of Pneumatic Muscle Systems .......... 48

2. SIMULATED TASK: ISOKINETIC KNEE EXTENSION STRENGTH TRAINING .......... 50
TABLE OF CONTENTS (continued)

2.1 Isokinetic Knee Extension Background ........................................................................ 50
2.2 Need for Accommodation of Different Neuromuscular Behaviors ............................ 52
2.3 Exercise in Microgravity .............................................................................................. 53
2.4 Task Details ................................................................................................................. 54

3. THREE-ELEMENT PHENOMENOLOGICAL PNEUMATIC MUSCLE ACTUATOR
   MODEL .......................................................................................................................... 55

4. EXPERIMENTAL SETUP (DYNAMIC TEST STATION) ............................................. 59
   4.1 Parts of the Dynamic Test Station ............................................................................ 60
   4.2 Interaction Between Motor and PMA ....................................................................... 65
   4.3 Motor Control ........................................................................................................... 67

5. TASK-SPECIFIC RESISTANCE PRODUCED BY PNEUMATIC MUSCLE ACTUATOR
   FOR EXERCISE IN MICROGRAVITY ......................................................................... 68
   5.1 Objectives ................................................................................................................ 68
   5.2 Methods .................................................................................................................... 69
      5.2.1 Experimental Methods ...................................................................................... 69
         5.2.1.1 Inverse Model Open Loop Control ............................................................... 69
         5.2.1.2 Determining Desired Profiles ....................................................................... 71
      5.2.2 Experimental Procedure .................................................................................... 77
   5.3 Results ...................................................................................................................... 78
   5.4 Discussion ................................................................................................................. 84

6. SIMULATING NEUROMUSCULAR ACTUATION ......................................................... 87
   6.1 Human Quadriceps Dynamic Simulator (HQDS) .................................................... 87
   6.2 Scaling Level of Performance .................................................................................. 93
   6.3 Simulating Different Neuromuscular Behaviors ...................................................... 95
7. HAPTIC CONTROL OF PMA ISOKINETIC STRENGTH TRAINING SYSTEM .......................... 100

7.1 Control Structure ........................................................................................................ 100

7.2 Task Motion Controller (TMC) ..................................................................................... 101

7.3 PMA Motion Controller (PMC) .................................................................................... 106

7.4 Simulation Studies ........................................................................................................ 109

7.5 Experimental Procedure ............................................................................................. 110

7.6 Statistical Analysis ....................................................................................................... 113

7.7 Results .......................................................................................................................... 115

7.7.1 Baseline Motor Performance ..................................................................................... 115

7.7.2 Haptic Control Performance ...................................................................................... 118

7.7.2.1 Performance Evaluation using RMSE .................................................................. 118

7.7.2.2 Time Dependent Performance Results ............................................................... 125

7.7.2.3 Accommodation of Different Neuromuscular Behavior Types ......................... 144

7.7.2.4 Adaptation Results ............................................................................................... 149

7.7.2.5 Effect of Velocity Disturbance ............................................................................. 151

7.8 Discussion ...................................................................................................................... 152

7.8.1 Controller Performance ............................................................................................. 152

7.8.2 Velocity Tracking Goals ........................................................................................... 158

7.8.3 Variability .................................................................................................................. 158

7.8.4 Limitations of the PMA Dynamic Test Station and Haptic Control System ......... 160

7.8.5 Limitations of the Human Quadriceps Dynamic Simulator ..................................... 161

8. IMPLICATIONS OF RESEARCH AND FUTURE WORK ............................................ 164

8.1 Suggestions for Human Implementation ....................................................................... 164

8.2 Potential Control Improvement ..................................................................................... 166
### TABLE OF CONTENTS (continued)

8.3 Expanding System Functionality .................................................................................. 167

9. CONCLUSIONS .................................................................................................................. 169

APPENDIX A. ADDITIONAL MOTOR CONTROL INFORMATION ........................................... 171

APPENDIX B. WIRING OF MAJOR COMPONENTS WITHIN DYNAMIC TEST STATION ... 177

APPENDIX C. DYNAMIC TEST STATION SETUP INSTRUCTIONS .................................... 179

APPENDIX D. CALIBRATION PROCEDURES ................................................................ 191

APPENDIX E. LABVIEW DTS CONTROL INFORMATION ............................................... 193

APPENDIX F. CLOSED LOOP CONTROL COMMAND SIGNALS ...................................... 206

APPENDIX G. DESIRED MOTION PROFILES .................................................................. 211

APPENDIX H. ADDITIONAL CLOSED LOOP CONTROL PERFORMANCE RESULTS ... 217

APPENDIX I. ANOVA RESULTS EVALUATING EFFECT OF SIMULATION SCENARIO AND PHASE ON TRACKING ACCURACY ......................................................... 221

APPENDIX J. ADDITIONAL CONTROL OPTIONS ............................................................. 234

APPENDIX K. DESIGN IDEA FOR PMA PRODUCED RESISTANCE STRENGTH TRAINING DEVICE ........................................................................................................ 237

REFERENCES ....................................................................................................................... 238
LIST OF FIGURES

Figure 1. Illustration of pneumatic muscle operation .......................................................... 5

Figure 2. Illustration of agonist-antagonist pneumatic muscle pair connected around pulley ........ 7

Figure 3. Three-element phenomenological model ................................................................ 55

Figure 4. Diagram of Dynamic Test Station .......................................................................... 59

Figure 5. Photograph of Festo fluidic muscle, slide and LVDT .................................................. 62

Figure 6. Photograph of motor and cable attachment ............................................................. 62

Figure 7. Photograph of additional motor attachments ............................................................ 63

Figure 8. Photograph (top-view) of PPR and pressure inlet of Festo fluidic muscle ............... 64

Figure 9. (a) Illustration of PMA, cable and spool before task begins  
(b) Illustration of PMA, cable and spool at different task angles ........................................ 66

Figure 10. Flow diagram of task-specific PMA resistance control ............................................. 71

Figure 11. Proximal, distal, and center of mass locations for scaled lower leg and foot through the range of motion of the simulated task ......................................................... 74

Figure 12. Resistance force (derived from torque requirements) as a function of motor position 77

Figure 13. Position and PMA displacement results for resistance profile equivalent to 4 lb ankle weight .............................................................................................................. 80

Figure 14. Force results for resistance profile equivalent to 4 lb ankle weight ....................... 80

Figure 15. Position and PMA displacement results for resistance profile equivalent to 8 lb ankle weight ........................................................................................................... 81

Figure 16. Force results for resistance profile equivalent to 8 lb ankle weight ................. 81
LIST OF FIGURES (continued)

Figure 17. Position and PMA displacement results for resistance profile equivalent to 16 lb ankle weight………………………………………………………………………………………………………...82

Figure 18. Force results for resistance profile equivalent to 16 lb ankle weight …………………..82

Figure 19. Position and PMA displacement results for accommodating resistance profile……..83
Figure 20. Force results for accommodating resistance profile …………………………………..83

Figure 21. Relationship between simulated peak torque and joint angle/motor position with $a_{k_T} = a_{k_T} = 1$……………………………………………………………………………………………………92

Figure 22. Relationship between simulated peak velocity and joint angle/motor position with $a_{k_v} = 1$………………………………………………………………………………………………………93

Figure 23. Torque-velocity relationships for artificially stimulated muscle (top curve) and scaled voluntary contractions (bottom curves) …………………………………………………………94

Figure 24. Haptic PMA control system diagram ………………………………………………………101

Figure 25. Diagram illustrating system forces………………………………………………………….101

Figure 26. Step response of motor ………………………………………………………………………116

Figure 27. Baseline motor performance (automatic command of desired motion profile with no haptic control)………………………………………………………………………………117

Figure 28. Position RMSE for each simulation scenario and each task phase…………………..118

Figure 29. Velocity RMSE for each simulation scenario and each task phase…………………..119

Figure 30. Torque RMSE for each simulation scenario and each task phase…………………..119

Figure 31. PMA Displacement RMSE for each simulation scenario and each task phase ……120

Figure 32. LS Means according to simulation scenario and task phase…………………………124

Figure 33. Position results for 0.175 rad/s simulation scenario …………………………………126

Figure 34. Velocity results for 0.175 rad/s simulation scenario …………………………………127

Figure 35. Torque results for 0.175 rad/s simulation scenario……………………………………127
LIST OF FIGURES (continued)

Figure 36. PMA displacement results for 0.175 rad/s simulation scenario ......................... 128

Figure 37. Position results for 0.314 rad/s simulation scenario ........................................ 128

Figure 38. Velocity results for 0.314 rad/s simulation scenario ........................................ 129

Figure 39. Torque results for 0.314 rad/s simulation scenario ........................................... 129

Figure 40. PMA displacement results for 0.314 rad/s simulation scenario ......................... 130

Figure 41. Position results for 0.524 rad/s simulation scenario ........................................ 130

Figure 42. Velocity results for 0.524 rad/s simulation scenario ........................................ 131

Figure 43. Torque results for 0.524 rad/s simulation scenario ........................................... 131

Figure 44. PMA displacement results for 0.524 rad/s simulation scenario ......................... 132

Figure 45. Position results for 0.314 rad/s, 20-repetitions fatigue simulation scenario .......... 132

Figure 46. Velocity results for 0.314 rad/s, 20-repetitions fatigue simulation scenario .......... 133

Figure 47. Torque results for 0.314 rad/s, 20-repetitions fatigue simulation scenario .......... 133

Figure 48. PMA displacement results for 0.314 rad/s, 20-repetitions fatigue simulation scenario ................................................................. 134

Figure 49. Position results for 0.314 rad/s, 40-repetitions fatigue simulation scenario .......... 134

Figure 50. Velocity results for 0.314 rad/s, 40-repetitions fatigue simulation scenario .......... 135

Figure 51. Torque results for 0.314 rad/s, 40-repetitions fatigue simulation scenario .......... 135

Figure 52. PMA displacement results for 0.314 rad/s, 40-repetitions fatigue simulation scenario ................................................................. 136

Figure 53. Position results for 0.314 rad/s, rapid fatigue simulation scenario ...................... 136

Figure 54. Velocity results for 0.314 rad/s, rapid fatigue simulation scenario ...................... 137

Figure 55. Torque results for 0.314 rad/s, rapid fatigue simulation scenario ...................... 137
LIST OF FIGURES (continued)

Figure 56. PMA displacement results for 0.314 rad/s, rapid fatigue simulation scenario .......... 138
Figure 57. Position results for 0.314 rad/s, erratic-sine wave simulation scenario .................. 138
Figure 58. Velocity results for 0.314 rad/s, erratic-sine wave simulation scenario .................. 139
Figure 59. Torque results for 0.314 rad/s, erratic-sine wave simulation scenario .................. 139
Figure 60. PMA displacement results for 0.314 rad/s, erratic-sine wave simulation scenario .... 140
Figure 61. Position results for 0.314 rad/s, erratic-pulse simulation scenario ...................... 140
Figure 62. Velocity results for 0.314 rad/s, erratic-pulse simulation scenario ...................... 141
Figure 63. Torque results for 0.314 rad/s, erratic-pulse simulation scenario ...................... 141
Figure 64. PMA displacement results for 0.314 rad/s, erratic-pulse simulation scenario ......... 142
Figure 65. Target torques for different task speeds .............................................................. 146
Figure 66. Output torques for different task speeds .............................................................. 147
Figure 67. Target torques for fatigue simulation scenarios compared to 0.314 rad/s baseline .... 147
Figure 68. Output torques for fatigue simulation scenarios compared to 0.314 rad/s baseline ... 148
Figure 69. Target torques for erratic simulation scenarios compared to 0.314 rad/s baseline .... 148
Figure 70. Output torques for erratic simulation scenarios compared to 0.314 rad/s baseline .... 149
Figure 71. Time course of adapted predicted human torque coefficient (\(\hat{\tau}\)) for different task
velocities .................................................................................................................................. 150
Figure 72. Time course of adapted predicted human torque coefficient (\(\hat{\tau}\)) for fatigue simulation
scenarios ................................................................................................................................... 151
Figure 73. Time course of adapted predicted human torque coefficient (\(\hat{\tau}\)) for erratic simulation
scenarios ................................................................................................................................... 151
LIST OF TABLES

Table 1. Pneumatic Muscle Actuator Characterizations Using Three-Element Phenomenological Model.................................................................57

Table 2. Set 1 of Calibration Equations for DTS Devices .................................................................65

Table 3. Set 2 of Calibration Equations for DTS Devices .................................................................65

Table 4. PMA three-element phenomenological model characterization for Festo MAS-20 fluidic muscle .................................................................................................70

Table 5. Mass and Lengths for Model and Scaled Human Body Segments ........................................73

Table 6. Relationship Between Resistance Replacement and Scaled Human Quadriceps ............76

Table 7. Total force resistance equations as a function of motor position (simulated joint angle in degrees of flexion) .................................................................77

Table 8. RMSE values between desired outputs and actual outputs for different PMA resistance profiles and different task phases .................................................................79

Table 9. Baseline motor performance results for independent motor control; no haptic control applied ........................................................................................................117

Table 10. Average RMSE and corresponding standard deviation for each simulation scenario and task phase ........................................................................................................121
### SYMBOLS, ABBREVIATIONS, AND ACRONYMS

<table>
<thead>
<tr>
<th>Symbol</th>
<th>Description</th>
<th>Unit</th>
</tr>
</thead>
<tbody>
<tr>
<td>PMA</td>
<td>pneumatic muscle actuator</td>
<td></td>
</tr>
<tr>
<td>$F$</td>
<td>tensile/contraction force</td>
<td></td>
</tr>
<tr>
<td>$P$</td>
<td>inflation gage pressure</td>
<td></td>
</tr>
<tr>
<td>$D$</td>
<td>tubing diameter when $\alpha = 90^\circ$</td>
<td></td>
</tr>
<tr>
<td>$\gamma$</td>
<td>fiber angle</td>
<td></td>
</tr>
<tr>
<td>$k_e$</td>
<td>elastic constant of inner tubing</td>
<td></td>
</tr>
<tr>
<td>$L$</td>
<td>pneumatic muscle/tube length</td>
<td></td>
</tr>
<tr>
<td>$D_0$</td>
<td>diameter of tubing when not inflated</td>
<td></td>
</tr>
<tr>
<td>$u_s$</td>
<td>coefficient of friction between strands of the helical sleeve</td>
<td></td>
</tr>
<tr>
<td>$u_{st}$</td>
<td>coefficient of friction between strands of the sleeve and inner tube</td>
<td></td>
</tr>
<tr>
<td>$P_c$</td>
<td>difference between $P$ and $P_i$</td>
<td></td>
</tr>
<tr>
<td>$P_i$</td>
<td>pressure required to inflate unconstrained inner tube to device diameter</td>
<td></td>
</tr>
<tr>
<td>$\varepsilon$</td>
<td>contraction rate</td>
<td></td>
</tr>
<tr>
<td>$a, b$</td>
<td>Rubbertuator constants</td>
<td></td>
</tr>
<tr>
<td>$P_0$</td>
<td>initial inflation pressure</td>
<td></td>
</tr>
<tr>
<td>$\Delta P$</td>
<td>pressure difference between two pneumatic muscles</td>
<td></td>
</tr>
<tr>
<td>$I$</td>
<td>mass moment of inertia</td>
<td></td>
</tr>
<tr>
<td>$C$</td>
<td>viscosity coefficient</td>
<td></td>
</tr>
<tr>
<td>$K_1$</td>
<td>proportional constant to $P_0$</td>
<td></td>
</tr>
<tr>
<td>$K_2$</td>
<td>proportional constant to $\Delta P$</td>
<td></td>
</tr>
<tr>
<td>$\theta$</td>
<td>angular rotation/displacement or simulated knee flexion angle</td>
<td></td>
</tr>
<tr>
<td>$t_k$</td>
<td>bladder thickness</td>
<td></td>
</tr>
<tr>
<td>$D_{cap}$</td>
<td>end cap diameter</td>
<td></td>
</tr>
<tr>
<td>$h$</td>
<td>helical fiber length</td>
<td></td>
</tr>
<tr>
<td>$N_T$</td>
<td>number of trapezoids in fiber mesh</td>
<td></td>
</tr>
<tr>
<td>$\gamma_o$</td>
<td>initial fiber angle</td>
<td></td>
</tr>
<tr>
<td>$k$</td>
<td>constant for which accounts for end deformation</td>
<td></td>
</tr>
<tr>
<td>$K_B$</td>
<td>compensation term based on the braid’s elasticity</td>
<td></td>
</tr>
</tbody>
</table>
SYMBOLS, ABBREVIATIONS, AND ACRONYMS (continued)

\( F_f \) compensation term related to Coulomb friction
\( K_e \) constant based on the rubber tube’s material properties
\( N \) number of rolling fiber surrounding the tube
\( L_o \) original pneumatic muscle/tube length
\( x \) linear displacement of PMA
\( A(x) \) virtual piston area
\( F(x) \) tension force of the spring within Hildebrandt model
NARX nonlinear autoregressive with exogenous input
ARNN adaptive recurrent neural network
FEM finite element method
\( P' \) internal gas pressure relative to water pressure
\( P_r \) pressure to overcome the elasticity of the bladder for expanding
\( N_{TX} \) number of trapezoids in fiber mesh along X direction
\( N_{TY} \) number of trapezoids in fiber mesh along Y direction
\( K_L \) contraction resistance constant
\( K_E \) contributed erosion factor of the latex bladder
PID proportional, integral, derivative
ILC iterative learning controller
DTS Dynamic Test Station
\( m \) mass of moving components connected to pneumatic muscle actuator
\( B_{PMA} \) damping coefficient of three-element phenomenological model
\( K_{PMA} \) spring coefficient of three-element phenomenological model
\( F_{ce} \) contractile force coefficient of three-element phenomenological model
\( F_L \) external load of three-element phenomenological model
\( F_R \) total load working against PMA, including effects of mass
\( \dot{x} \) PMA velocity
\( \alpha \) angular acceleration
\( T_{motor} \) torque provided by the motor
\( T_{lowerleg} \) torque required to move the lower leg
\( T_{foot} \) torque required to move the foot
\( T_{rr} \) replacement resistance torque
\( T_{PMA} \) total desired torque to be produced by the PMA
SYMBOLS, ABBREVIATIONS, AND ACRONYMS (continued)

\( g \)                   \text{acceleration due to gravity}
HQDS                \text{Human Quadriceps Dynamic Simulator}
\( T_{o,\text{max}} \) \text{peak isometric torque}
\( V_{o,\text{max}} \) \text{peak angular velocity}
\( a_{s,\text{Tre}} \) \text{concentric simulated torque scaling parameter within HQDS}
\( a_{s,\text{Tte}} \) \text{eccentric simulated torque scaling parameter within HQDS}
\( a_{s,V} \) \text{simulated velocity scaling parameter}
\( T_{s,c} \) \text{HQDS concentric torque output}
\( T_{s,e} \) \text{HQDS eccentric torque output}
\( V_{s,c} \) \text{HQDS velocity during concentric contraction}
\( V_{s,e} \) \text{HQDS velocity during eccentric contraction}
\( \omega_d \) \text{desired task velocity}
\( T_{s,\text{ooc}} \) \text{position-dependent peak value of simulated concentric torque}
\( T_{s,\text{oe}} \) \text{position-dependent peak value of simulated eccentric torque}
\( V_{s,o} \) \text{position-dependent peak value of simulated velocity}
\( T_{\text{out}} \) \text{torque output}
\( L_{\text{norm}} \) \text{normalized muscle length}
\( a_{n,\text{fatigued}} \) \text{fatigue simulation scenario scaling parameter}
\( a_{n,\text{initial}} \) \text{scaling parameter before fatigue begins}
\( \Delta t_f \) \text{time to fatigue}
\( a_{n,\text{erratic-sinewave}} \) \text{erratic-sine wave simulation scenario scaling parameter}
TMC                \text{Task Motion Controller}
\( T_{h,n} \) \text{predicted estimate of human neuromuscular actuator produced torque}
\( \hat{c} \) \text{estimate of predicted human torque coefficient}
\( \hat{b} \) \text{estimate of predicted leg torque coefficient}
\( \hat{I} \) \text{estimate of moment of inertia}
\( \tilde{c} \) \text{error of predicted human torque coefficient}
\( \tilde{b} \) \text{error of predicted leg torque coefficient}
\( \tilde{I} \) \text{error of moment of inertia}
\( a_{Tc} \) \text{concentric torque scaling parameter based on predicted system outputs}
\( a_{Te} \) \text{eccentric torque scaling parameter based on predicted system outputs}
SYMBOLS, ABBREVIATIONS, AND ACRONYMS (continued)

\( a_V \)  velocity scaling parameter based on predicted system outputs
\( \varphi \)  angle used to define location of COM of combined lower leg and foot
COM  center of mass
\( T_R \)  torque resistance command calculated within TMC
\( s \)  combined error
\( e \)  tracking error
\( k \)  nonlinear adaptive controller gain
\( \lambda_\omega \)  control bandwidth setting within nonlinear adaptive control law
\( V \)  Lyapunov function candidate
PMC  PMA Motion Controller
\( V_{PPR} \)  voltage command to PPR (proportional pressure regulator)
\( V_{motor} \)  voltage command to motor
RMSE  root mean square error
CV  coefficient of variation
1. INTRODUCTION AND BACKGROUND

1.1. Problem and Research Goals

A pneumatic muscle actuator (PMA) is a device that mimics behavior of skeletal muscle by contracting and generating force in a nonlinear manner when activated (pressurized). Due to their nonlinear dynamics, PMAs are difficult to control, presenting challenges in system implementation. However, PMAs have a high power to weight ratio and possess unique characteristics which make them ideal for human interaction. These traits have led several researchers to use pneumatic muscles as a source of actuation within devices designed for rehabilitation and mobility assistance. The development of rehabilitation and mobility assistance devices have focused on the upper body/arm (1) (2) (3) (4), hand (5) (6), lower body/leg (7) (8) (9) (10), combined upper and lower body (11), as well as gait training (12) (13) (14) (15).

Researchers at Wright State University (Dayton, Ohio) have developed proof of concept designs for pneumatic muscle actuator produced assistance during a sit-to-stand movement (16) and PMA produced resistance for an isokinetic knee extension exercise (17). This work builds on the work of (17), but now PMA displacement and force output are being controlled using a pneumatic muscle actuator (PMA) model developed by Reynolds and Repperger (18) and characterized by Serres (19).

Isokinetic concentric and eccentric knee extension will benefit those who need to build quadriceps strength through the joint’s range of motion. Isokinetic knee extension has the potential to improve performance in activities such as walking, bicycling, swimming, and rising from a seated position, something particularly important for the elderly in maintaining their mobility. PMAs possess unique characteristics which make them ideal for human interaction.
They also have a high force output to weight ratio, are clean, and can operate in harsh environments. These traits, along with the fact that gravity is not necessary for the pressure-activated PMA to produce resistance, make them an attractive option for a microgravity exercise device. A device such as this would enable astronauts to counteract muscle atrophy during prolonged spaceflight. Bone loss and muscle atrophy from a lack of gravity-related loading, along with aerobic deconditioning are major concerns during extensive flights in microgravity environments (20). PMAs may also have potential for use within a portable exercise device because of their high force to weight characteristics.

Control of a PMA system capable of producing controlled resistance is challenging due to the nonlinear dynamics of PMA, the need to produce a controlled resistance against an opposing dynamic actuator, and the need to control not only force output of the PMA but also PMA displacement within the dynamic system.

The first part of this work focuses on controlling PMA displacement and force according to task-specific time-varying force resistance profiles. Predefined force resistance profiles, as well as predefined PMA displacement profiles are fed into an open loop controller in an attempt to produce desired PMA behavior. A motor, controlled to work against PMA produced resistance, acts as a source of simulated human movement in order to demonstrate simulated concentric and eccentric isokinetic knee extension.

The second part of this work focuses on developing a human quadriceps dynamic simulator (HQDS). A motor, which produces torque analogous to the torque produced by the quadriceps about the knee during knee extension, is used in conjunction with the HQDS to simulate neuromuscular actuation. The purpose of implementing the simulated neuromuscular actuator within the experimental system setup is to test the effectiveness and robustness of a closed loop control system prior to human implementation. The simulated neuromuscular actuator developed in this study responds to the PMA force resistance changes by changing its velocity output, analogous to a human operator decreasing or increasing the speed of their movement in
response to resistance changes they feel. The simulated neuromuscular actuator also demonstrates various neuromuscular behaviors in order to test the controller’s ability to accommodate different types of simulated behavior.

The third part of this work focuses on developing closed loop control capable of producing controlled PMA resistance for a concentric and eccentric isokinetic knee extension exercise. A nonlinear adaptive control technique is used in combination with an inverse three-element phenomenological model to create haptic control designed to improve the simulated neuromuscular actuator’s velocity performance using appropriate force feedback. The force resistance provided by the PMA control system will automatically adjust in order to keep a simulated joint moving at a constant angular velocity, demonstrating isokinetic dynamic behavior of the simulated neuromuscular actuator.

The controller is not intended to be used for a device whose purpose is to measure performance (similar to an isokinetic dynamometer). Instead, the controller attempts to accommodate simulated human movement within a potential force output range. The controller adjusts the force resistance in order to assist in the production of isokinetic movement at the desired task speed. Accommodation of simulated physiological fatigue and erratic operational behavior is demonstrated. This work represents an important step towards implementation with a human operator.

This research is unique in its purpose, system setup, and control design. The purpose of this work is unique in that PMA research focused on rehabilitation of the leg, to the best of my knowledge, does not attempt to accommodate physiological fatigue or erratic operational behavior. Past PMA research has not focused on providing resistance for both concentric and eccentric contractions, and no PMA research has focused on exercise in microgravity environments (outside of the work done at Wright State University). A great deal of PMA applications (human inspired robots) try to replicate how human or animal joints operate. A goal of the control system developed here is to react to simulated human movement, not imitate human
movement. The control system is unique in that it uses a three-element phenomenological model developed by Reynolds and Repperger (18) (21). Very few groups have used the three-element phenomenological model in the development of PMA control design.

1.2. Background and Review of Literature

Approximately seventy-seven groups have published work on pneumatic muscle technology. Their work was reviewed in order to assess the current state of technology in the field of pneumatic muscle research. Pneumatic muscle construction, modeling, and control techniques have advanced in the last twenty years. Applications are as widespread as walking robots to upper arm rehabilitation devices to industrial equipment.

1.2.1. Pneumatic Muscle Background

1.2.1.1. Description of Pneumatic Muscle

A pneumatic muscle actuator (PMA) is a mechanical device that mimics the behavior of skeletal muscle in that it contracts and generates force in a nonlinear manner when activated (22). PMAs are constructed of a tubular shaped rubber bladder and an inextensible fiber mesh that either surrounds or is embedded in the rubber matrix. The fiber mesh provides support and enhances actuation. The rubber bladder is completely sealed except for an air valve that allows air to enter and exit. Once pressurized, the PMA expands in a radial direction resulting in contraction and force production in the longitudinal direction (analogous to skeletal muscle). Figure 1 illustrates the operation of a PMA. The level of contraction and force production is dependent on whatever is attached to or working against the PMA.
This type of actuator has several unique characteristics, some of which make it an ideal actuator for applications involving human interaction. Pneumatic muscles are capable of generating a high force output. They have higher power to weight and power to volume ratios than electric motors or hydraulic actuators (21). They have a higher force output than a pneumatic cylinder of equal volume (23). Pneumatic muscles are cost effective, clean, compact, and can be used in harsh environments because they do not have moving parts such as pistons or guiding rods (24). Pneumatic muscles can be used in microgravity environments because gravity is not necessary to produce contraction or force generation. They are also a safe alternative to other actuators. They provide “soft actuation” meaning safety is enhanced through a low mass/inertia structure that combines high strength with actuator and/or structural compliance (25). Their “soft actuation” allows them to be used around humans without posing safety risks associated with other actuators that are heavy and noncompliant. There is a potential safety mechanism built into the manner in which the PMA operates and transfers force in this research. Force is only produced if there is something working against the PMA. If, theoretically speaking, a human working against the force produced by the PMA for strength training purposes does not want to continue the exercise, they can stop safely and without consequence.
The main disadvantage in using PMAs is the difficulty involved with controlling them due to their nonlinear dynamics and time varying behavior. The main source of nonlinearity can be explained by the basic operation of a pneumatic muscle. The general force expression for the pneumatic muscle is pressure times cross sectional area of the rubber tubing (21). As pressure increases linearly, the tubing expands radially resulting in a nonlinear increase in the tube’s area. Thus the force (pressure times area) will increase nonlinearly. In this work, the PMA is operated in a configuration that has the potential to improve system response time and human safety. This configuration, however, increases control difficulty because of the need to control both PMA displacement and PMA force using only one parameter (PMA pressure).

1.2.1.2. Pneumatic Muscle Operation Methods

Many pneumatic muscle applications are driven by an agonist-antagonist pair of pneumatic muscles aligned in parallel, connected by a cable that moves around a pulley (see Figure 2). A pressure differential between the two pneumatic muscles produces movement and torque in an arm (or other torque transfer assembly) rigidly connected to the pulley. Other applications use two in-line opposing pneumatic muscles to move an assembly between them (26), while others configure mechanisms such as springs (13) (27) or magneto-rheological brakes (28) in opposition to a PMA. An independently operating PMA, with a single line of force transfer was used in this research. The PMA is connected to the external load by a cable that is rigidly fixed to a spool. The cable is able to wrap/unwrap around the spool as the PMA displacement changes.
1.2.1.3. History of Pneumatic Muscle

The earliest example of a braided pneumatic actuator, the “Expansible Cover”, was patented by Robert Pierce in 1936 (29). The device was a rubber bladder covered by a weaving of braided wires fixed by rings at the ends. The suggested use for the device was replacing dynamite in the coal mining industry. The idea was that when the rubber bladder is filled with air, the device expands radially and applies a force against the coal, knocking it down. This invention did not make use of the device’s longitudinal contraction. In 1949, a patent was granted to Hugh DeHaven for a “Tensioning Device for Providing a Linear Pull” (30). This device consisted of an expansible inner tube and an outer tube of diagonally woven strands which formed a helical outer structure. The source of gas pressure was built into the inner tube. DeHaven claimed that a nine ounce, three foot unit, pressurized to around 400 psi could contract about 30% and provide a tensile force of 1500 lb. In 1958, a patent was granted to Richard Gaylord for a “Fluid Actuated Motor System and Stroking Device” (31). This device was similar in operation to DeHaven’s but it was described as an actuator and had an external pressure source. Gaylord was the first to develop an equation to calculate force of the actuator.

The pneumatic muscle with helically wound fibers was popularized by Joseph McKibben in the 1950s when he used it to activate orthotic devices for upper extremities (32). He did not patent his device (33), but his work was published by Baldwin in 1963 (34). In the early 1960s, Schulte published details of the braided pneumatic muscle which he called McKibben artificial
muscle along with mathematical analysis included in Gaylord’s patent (35). Due to challenges controlling the pneumatic muscle and the large gas tank required for its activation, use of pneumatics muscles remained limited (32) (33). In the 1980s, pneumatic muscles resurfaced in the robotics industry. Bridgestone developed a commercial version of the pneumatic muscle called a Rubbertuator and used it in industrial robotic arms (32).

Work on the Rubbertuator ended by the 1990s, but others have continued working on pneumatic muscle due to advantageous features including high force generation, high power to weight and power to volume ratios, and soft-actuation (allowing safe pneumatic muscle-human interaction). In 2002, Festo Corporation patented a more robust pneumatic muscle design, called the fluidic muscle (32). At the present time, companies such as Festo and Shadow Robot are producing commercially available pneumatic muscles in many sizes and configurations (33).

1.2.2. Applications for Pneumatic Muscle

The use of pneumatic muscle technology has spread into many different fields including robotics, human power and mobility assistance, therapy and rehabilitation, and prosthetics design. Pneumatic muscles have also been used for several industrial applications. Examples of pneumatic muscle use in each of these fields are given.

1.2.2.1. Robotics

Several groups of researchers are using pneumatic muscles as actuators in robotic devices. Whole body humanoids, robotic arms, bipedal robots, and other robots modeled after the lower limb have been developed or are currently under development. A climbing robot has been developed as well as robots inspired by animals and insects.

Robotic arms have been created by groups from around the world. Here a robotic arm is considered to be something more complicated than a one link or one segment manipulator. Hesselroth et al. developed a five joint robot arm and gripper activated using pneumatic muscles
and controlled using neural networks and feedback from two video cameras (36). The researchers claim that the solution of the control problem has implications for biological visuo-motor control. Hannaford et al. created a replica human arm actuated with pneumatic muscles in an attempt to improve the understanding of the reflexive control of human movement and posture (37). Muscles of the biorobotic arm were replicated using fifteen pneumatic muscles, bones using fiberglass composite, joints using surgical joint replacements, and ligaments using knit-fabrics. In 1994, Tondu et al. developed a two degree of freedom robot arm actuated by two agonist-antagonist pneumatic muscle pairs (38). The natural compliance of the robot arm was studied. Natural compliance (also called passive compliance) is compliance without any computing assistance. It is beneficial in robotics because it may result in reduced control complexity and reduced power usage. It is widely used in robotics, particularly walking robots as will be discussed later. In 1997, Tondu and Lopez studied closed loop nonlinear control methodologies for their two degree of freedom robot arm including sliding mode control (39). In 2002, Eskiizmirliler, Forestier, Tondu, et al. applied control modeled after cerebellar pathways (40). This model used a linear approximation of biological muscle’s force-length relationship and equated it to variable stiffness of pneumatic muscle. The slope characterizing the force decrease relative to muscle contraction is proportional to neural activation. Neural activation was correlated to pressure in the pneumatic muscle. In 2005, Tondu et al. developed a joy-stick controlled seven degree of freedom robot arm actuated by agonist-antagonist pneumatic muscle pairs (41). Testing was done to study the robot’s compliance in interactions with the environment. Tuijthof and Herder created a four degree of freedom robot arm whose movement was controlled using an energy balance principle (42). The arm was designed to be in perfect equilibrium until a pneumatic muscle pressure change caused a stiffness change altering the potential energy field. This resulted in arm movement to a new equilibrium position.

Many groups have applied pneumatic muscles in the development of robotic legs and bipedal walking robots. Most groups focus on the adjustment of compliance or stiffness of
agonist-antagonist muscle pairs. Colbrunn et al. controlled position and stiffness (natural/passive compliance) at each joint by increasing pneumatic muscle pressure analogous to increasing a biological muscle’s activation level (43). Energy efficient control of a biped waking robot was the focus of Mao et al. (44). Pressures of muscles were derived experimentally based on desired joint motion and trajectory as well as required joint stiffness. In the design of Ichiro, a biped developed at Oita University’s Artificial Life and Robotics Laboratory in Japan, a program was developed to determine the basic shapes of lower limbs at different pneumatic muscle pressures (45). This was then used as a tool to create complex movements.

The bipedal robot Lucy developed in the Department of Mechanical Engineering at Vrije Universiteit Brussel in Belgium has six joints (ankle, knee and hip of both legs), each actuated by an agonist-antagonist pneumatic muscle pair (46). Lucy operates using joint trajectory control where trajectories are based on the objective locomotion parameters which can be changed from step to step, and it also uses compliance or stiffness control to reduce control efforts and energy consumption (47) (48) (49) (50). A dynamic simulation model (incorporating dynamic behavior, thermodynamic effects, and model errors) was used to evaluate the control system (51). Simulations show only small deviations between desired and attained locomotion parameters (including forward speed, step length, step height, foot lift) (52).

Another bipedal robot named Mowgli is able to jump and land using the compliance and natural dynamics of the legs actuated by six pneumatic muscles (27). Passive springs are used as the antagonists to the pneumatic muscles.

Takuma et al. controlled pneumatic muscle compliance using valve operations, relating valve opening times with the walking cycle of a bipedal robot (53) (54). Hosada, Takuma, et al. developed a biped that could walk and run by changing joint compliance according to desired dynamic locomotion, whether it be rigid for walking or compliant for jumping and running (55). Hosada et al. then implemented more sophisticated controls to achieve balance and stability with tuning parameters for sagittal movement, frontal balance and horizontal rotation (56) (57) (58).
Narioka and Hosaoda also developed a whole body humanoid which utilized the previously developed biped (59). Caldwell et al. have developed a whole body humanoid robot composed of a bipedal walking torso, two seven degree of freedom arms, and two five fingered hands (60). The robot, actuated by pneumatic muscles, was designed to model a human’s scale and range of motion (61).

A structure-climbing robot intended to carry cameras and sensors onto bridges, buildings, aircraft, and ships for inspection purposes is also controlled, like many of the bipedal robots, using a variable stiffness approach. The relationship between joint position and equilibrium pressure were obtained, and then used to control joint position (62).

In addition to the many human-inspired robots, several animal and insect inspired pneumatic muscle driven robots have been developed. A five degree of freedom cockroach inspired robot leg was developed (63). Recorded cockroach electromyograms were used to activate the pneumatic muscles and produce robot joint motion (64). A six legged robot named AirBug was designed to take advantage of passive compliance in rough terrain (65). A three degree of freedom flexible miniature robot based on inchworm type motion was developed for an endoscope inspection tool (66). An octopus inspired robotic mechanism was developed and controlled by finding the relationship between robot shape to pneumatic muscle length (67). A quadruped robot was developed using a two dimensional kinematic model using joint range of motion data and skeletal dimensions from large breed canines (68).

1.2.2.2. Power and Mobility Assistance Devices

Several devices have been designed to provide power and mobility assistance for humans. Many researchers have incorporated pneumatic muscles into these devices because of the inherent safety and high power to weight ratio provided by the pneumatic muscle.

Several upper body assistance devices have been developed, both exoskeleton devices worn by the operator and manipulators that only interact with the human operator. In 1999,
Tsagarakis et al. described their work on a seven degree of freedom powered arm exoskeleton (69). It was designed to replicate human arm kinematics and to form intimate contact with the arm of the user. Caldwell and Tsagarakis later reported work on a powered external arm support to aid those with partial paralysis or muscle atrophy (70). Closed loop position and torque control had a decent ability to cope with load variations. Kobayashi and Hiramatus developed a “muscle suit” to provide force assistance for those with paralysis or for those who need assistance in handling heavy loads (71). Here the pneumatic muscles were sewn into a garment worn by the operator. Martinez et al. develop an upper limb exoskeleton designed to provide force assistance during routine work-place related activities (72). All of the actuators (both pneumatic muscles and electric motors) were placed in a backpack while Bowden cable transmitted the force to the user.

Noritsugu and Tsuji developed a pneumatic muscle driven manipulator arm (not an exoskeleton or orthotic) designed to assist humans with force and mobility tasks (73). Van Damme et al. also developed a pneumatic muscle driven manipulator arm designed to provide humans force assistance (74). Here the human and robot both hold the load and work collaboratively to move it.

Lower body assistance devices include orthotics for assisting lower leg movement about the knee and ankle. Kim et al. developed a lower limb orthosis actuated by a pneumatic muscle to assist with flexion and extension of the knee (8). This research was performed to find the most suitable pressure to improve the muscular power of the elderly. A powered ankle-foot orthosis actuated by pneumatic muscles was developed at the University of Michigan, Departments of Biomedical Engineering and Movement Science, to provide plantar flexion and dorsiflexion torque during walking (75) (76) (9). Pneumatic muscle pressure was adjusted according to electromyography amplitude. Push button control was also used to test the effects of the orthotic device on subjects with incomplete spinal cord injury (77). A study using EMG control on neurologically intact subjects was used as a baseline to guide future studies on those with
neurological impairments (78). Researchers from the University of Michigan have also
developed a powered knee-ankle-foot orthosis to provide not only plantar flexion and dorsiflexion
torque, but also knee torque while walking (10). The design provided less torque assistance at the
knee than at the ankle, and thus must be improved.

1.2.2.3. Rehabilitation Devices

Many devices developed for therapeutic and rehabilitation purposes have incorporated
pneumatic muscles. Devices have been developed to improve gait training and the study of gait.
Devices have also been designed for upper extremity rehabilitation and hand rehabilitation.

Researchers at Wright State University (Dayton, Ohio) have developed proof of concept
designs for training the quadriceps muscle during a knee extension task (79) (17). Li et al. have
developed a basic legs rehabilitation exercise system with a parallel pair of agonist-antagonist
pneumatic muscle actuators connected by a chain around a chain wheel (7). The researchers
equate movement of an arm attached to the chain wheel to movement of knee joint. Experiments
show that movement of the pulley arm in continuous passive motion is smooth and that the
resistance of joints in active resistance motion is adaptive (adaptive meaning it changes with leg
position). However, there is no simulated operator in these experiments. The knee joint is simply
moving at the desired continuous speed and producing an amount of torque that is dependent on
joint angle.

Several gait training devices have been developed. Knight et al. developed a simple, low
cost gait trainer which used pneumatic muscles to assist with hip flexion and knee extension (12).
A push button trigger was used to activate the device. Bharadwaj et al. developed a gait trainer
that focuses on ankle rehabilitation for stroke patients (13) (80). Spring over muscle actuators, an
actuator combining pneumatic muscles with antagonist springs, assists ankle in
dorsiflexion/plantar flexion and inversion/eversion. Costa and Caldwell developed a gait trainer
consisting of twin five-link wearable legs powered by pneumatic muscles (14). Improvements
have to be made to control system in order to achieve acceptable performance. A gait training
device, developed by do Nascimento et al., was activated based on the angle of the hip joint rather
than an electromyography (EMG) signal to accommodate those with no EMG signal or a very
low quality EMG signal (15).

Researchers have used pneumatic muscle driven devices to not only improve gait, but
also to study gait. Studies have shown that robotic exoskeletons controlled by muscle activity
(EMG signal) could be useful in testing neural mechanisms of human locomotor adaptation (81).
Studies have also examined the metabolic cost of plantar flexion muscle-tendon work during
uphill walking using robotic ankle exoskeletons (82) (83).

A few pneumatic muscle driven rehabilitation devices have focused on recovery of the
arm. A two degree of freedom functional recovery training device for the arm was developed by
Noritsugu (1). Another upper body recovery training device was developed by Xiong et al. It had
two modes; one which applied pre-specified resistant torques to joints, and one which helped the
user improve functional movements such as eating or grasping (2). RUPERT (robotic upper
extremity repetitive therapy) was developed to restore stroke patients’ ability to perform activities
of daily living (84). Early work focused on developing a kinetic model of the integrated human
arm and robot used to calculate required forces for reaching and functional tasks under different
residual or spastic muscle forces. MATLAB’s SimMechanics helped to determine joint velocities,
accelerations, inertia and required torque (3). Different control methods have been implanted for
later versions of RUPERT including PID, iterative learning, fuzzy (4), and adaptive iterative
learning (85).

A device developed by Caldwell et al. combined upper and lower body rehabilitation
(11). The full body exoskeleton had seven degrees of freedom for the upper extremities and five
degrees of freedom for each lower extremity. The rehabilitation goal for the upper extremities
was functional therapy for stroke patients. The rehabilitation goal for the legs was gait training
utilizing an integrated treadmill and active body support system.
Hand therapy devices have been developed by Koeneman et al. (5) as well as Xing et al. (6). The Mentor, developed by Koeneman et al. works to extend the wrist and fingers. The device developed by Xing et al. focuses on the finger and thumb joints and takes the wearer’s intended and voluntary efforts into account rather than imposing predefined movements.

1.2.2.4. Technological Devices with Human Interaction

A few non-therapeutic devices specifically designed to interact with humans have utilized pneumatic muscles. A force feedback data glove has been developed as an interface for the human hand in virtual environments (86) (87). When the data glove interacts with virtual objects, a contact force is calculated according to the dimensions of the grasped object’s deformations. The data glove is designed to transfer the contact force sensation to the human wearing the glove using pneumatic muscles. A robotic “soft hand” activated by pneumatic muscles has also been developed to interact with a human (88). An example of the interaction is being able shake a human hand.

1.2.2.5. Prosthetics and Artificial Body Parts

A great deal of work has been done to design prosthetics and artificial body parts. There are a few examples where pneumatic muscles have been used in such a device. Versluys et al. developed a below the knee prosthesis driven by pneumatic muscles (89). Results showed potential for restoring ankle power to the user during level walking. Balan et al. developed a leg prosthesis that can be personalized according to patient characteristics (90). Movement of each segment was controlled using sensors and microcontrollers that command predefined movements. Lee developed an artificial hand modeled after the human hand in structure and motion (91). Scarfe and Lindsay also developed an artificial hand capable of grasping and relocating objects (92). This hand had ten controllable degrees of freedom actuated by twenty pneumatic muscles.
1.2.2.6. **Industrial Applications**

Pneumatic muscles have been used for several industrial applications including nuclear waste retrieval (93), construction machinery (94) (95) (96), a video-probe (97), and a nozzle positioning device (98). Underwater applications have also been studied (99). In some underwater applications, water can be used to activate the pneumatic muscle instead of air.

1.2.3. **Types of Pneumatic Muscle**

Several variations on the typical rubber, fiber-mesh, lab-constructed pneumatic muscle have been developed. These include pneumatic muscles with braided fibers, straight fibers, and pneumatic muscles constructed of alternative elastomeric and fiber materials. Pleated pneumatic muscles have been developed where the pleating either runs along the length of the muscle or perpendicular to the length of the muscle. Curved muscles have been developed where the rubber tube is no longer configured to be straight. Hybrid devices have incorporated springs with pneumatic muscles to act as a built-in antagonist, dampers and brakes designed to alter performance, and rings designed to increase contraction. Several pneumatic muscles are now commercially available.

1.2.3.1. **Braided/McKibben Pneumatic Muscle**

Dozens of groups have constructed pneumatic muscles with an inner rubber tube of some kind surrounded by a braided fiber mesh. The braided fibers are wound in a helical pattern and attached to the ends of the pneumatic muscle. Many groups in the past twenty years have called this type of braided pneumatic muscle a McKibben muscle after Dr. Joseph McKibben who helped to popularize pneumatic muscle use in the 1950s. Various tube lengths, diameters, fittings and materials have been used to construct these. The following paragraph provides information regarding braided pneumatic muscle construction from several research groups.
Caldwell et al. describe their device as having an inner layer made from thin-walled rubber tubing and an outer layer made from flexible sheathing formed from high-strength interwoven (but not bonded) nylon fibers (25). Perspex plastic plugs are used as end caps, sealing the ends of the tube. The nylon shell and rubber tubing are bonded to the end caps using a flexible adhesive and clamped securely in place using a rubber sealing ring. Pressure sensors used to monitor internal muscle pressure have been incorporated into at least some of their pneumatic muscles (11). Researchers associated with Wright State and Wright Patterson Air Force Base have constructed pneumatic muscles from bicycle tire tubing surrounded by either coax cable jacketing (100) or a nylon sheath used for supporting electrical cables (18). Ferris et al. from the University of Michigan constructed pneumatic muscles using latex tubing as the inner bladder, braided polyester sleeving as the muscle shell, plastic pneumatic fittings for the end caps and inlet/outlet valves. Double ear hose clamps were used to seal the ends (75). Chou, Hannaford, et al. have not given details of the construction of their pneumatic muscles, but describe them as having an airtight inner tube surrounded by a braided mesh shell with flexible yet inextensible threads (or threads having very high longitudinal stiffness) attached at either end to fittings (101) (37). Tondu and Lopez have also not given a lot of detail regarding their design, but it is known that their pneumatic muscles consist of an inner rubber tube surrounded by a double helix textile weave (39).

1.2.3.2. Straight Fiber Pneumatic Muscle

Straight fiber pneumatic muscle construction has also been used in the past. Baldwin noted that fibers aligned axially provide a potential for high mechanical efficiency (34). Nakamura, Saga, et al. have built pneumatic muscles reinforced by straight fibers (102) (103). These devices have a central ring which will be discussed in the Hybrid Pneumatic Muscle section. Nakamura reported that this type of muscle has a greater contraction ratio and a longer lifetime than conventional braided pneumatic muscles (102). Bettetto and Ruggiu studied straight
fiber pneumatic muscle using finite element method and found that straight fiber muscles can provide five times more tensile force than braided fiber muscles, but have a smaller contraction ratio (104).

1.2.3.3. Pleated Pneumatic Muscle

A pleated pneumatic muscle has a unique structure in that there is no outer sheath. The tube itself (referred to as a membrane) has pleats or folds. A pleated pneumatic muscle developed at the Vrije Universiteit Brussel, Department of mechanical Engineering, has folds along the longitudinal axis of the muscle (74). The folds hold the extra membrane material needed to expand. The membrane material has a high tensile stiffness in order to eliminate rubber-like strain (105). This type of pneumatic muscle has higher contraction forces and displacements (50% vs. only 20-30%) than a braided pneumatic muscle actuator, has low friction and no friction-related hysteresis, and can operate at low pressures. The fold faces are laid out radially so no friction is involved in the folding-unfolding process, and no loss of force output occurs during the unfolding process because no appreciable amount of energy is needed to unfold. The pleated pneumatic muscle was used in the walking bipedal robot Lucy (106).

Another pleated pneumatic muscle was developed by Zhang and Zhang, but here the pleats run perpendicular to the longitudinal axis of the muscle (107). This type of pleated pneumatic muscle actuator is referred to as a rubber bellows artificial muscle. The driving force of the actuator is determined by the size of the bellows and the internal pressure. The design of this actuator allows it to easily curve along its longitudinal axis, which is beneficial for joint actuation design. The radial elastic expansion of the actuator is limited, but the longitudinal elastic expansion range is large due to fiber weaving inside the rubber bellows.
1.2.3.4. Curved Pneumatic Muscle

Curved pneumatic muscle is discussed as a separate type of pneumatic muscle because of its unique configuration requiring unique model characterization.

Zhang et al. have developed a curved pneumatic muscle actuator constructed from Festo brand fluidic muscle for use in a wearable elbow exoskeleton (108). This type of pneumatic muscle configuration is based on a rotary actuator. It weakens the coupling relationship between the output torque/force and displacement so that torque/force feedback is more easily achieved. New modeling techniques including beam modeling and membrane modeling had to be implemented to obtain a mathematical model of a pneumatic muscle in the curved configuration.

Trivedi et al. developed a soft robotic manipulator named OctArm V that can bend into a wide variety of complex shapes when reacting to control inputs and/or gravitational loading (109). Nine pneumatic muscles are used in a curved configuration. This group of researchers developed a new approach for modeling the dynamics of the curved pneumatic muscle robotic manipulator that incorporates the effect of material nonlinearities and distributed weight.

1.2.3.5. Pneumatic Muscles Using Alternative Materials

Both alternative fiber materials and alternative elastomers have been used to construct pneumatic muscles. Alternative fiber materials include carbon (110) (103), glass fiber (102), Kevlar (111), and shape memory alloy wires (112). Wang et al. developed what was referred to as an intelligent pneumatic muscle using shape memory alloy wires in the braided shell. The pneumatic muscle operates as a typical pneumatic muscle would until the shape memory alloy wires are activated (via a temperature change). When activated, shape memory alloy wires contract, affecting movement of the braided shell and thus motion of the pneumatic muscle. Motion, contraction ratio, and force of the pneumatic muscle can therefore be adjusted by activating the shape memory alloy wires of the braided shell. Simulation results verify the
intelligent pneumatic muscle has a higher contractile force per cross-sectional area and more flexible stiffness than a traditional pneumatic muscle.

A pneumatic muscle developed by Goulbourne utilized a dielectric elastomer for the inner bladder of a pneumatic muscle (113). When activated by voltage, dielectric elastomers have a large strain actuation response (>100%), but a low force output. The low force output has limited its use for many applications, but by enclosing the dielectric elastomer in a helical network of inextensible fibers, the load-bearing capability of the dielectric elastomer improves. This so called electro-pneumatic actuator expands radially and contracts axially when activated by pressure and voltage.

1.2.3.6. Hybrid Devices
Several pneumatic muscle hybrid devices have been developed. In these devices, the pneumatic muscle does not act alone in its actuation.

The first hybrid device, spring over muscle, combined a spring and pneumatic muscle in a parallel configuration to create a bidirectional actuator (114). Contraction was produced using the pneumatic muscle while extension was produced with the spring (84). In early designs, the spring was designed to go over a commercially available pneumatic muscle. In later designs, the spring was inside the internal bladder of a custom made pneumatic muscle.

A second hybrid device consisted of a straight fiber pneumatic muscle with a ring mounted around the outside. The ring was located halfway along the length of the muscle. The purpose of the ring was to increase the contraction ratio (110) (102) (103) (111). Contraction was shown to be four times larger than a typical pneumatic muscle.

A third hybrid device combined a pneumatic muscle with a hydraulic damper in an attempt to improve the actuator’s velocity-dependent properties (115). Klute et al. designed the device with the hydraulic damper and pneumatic muscle in parallel. This preserved the muscle-
like force-length characteristics of pneumatic muscle, but altered the force-velocity properties to better replicate force-velocity properties of biological muscle.

Another hybrid device attached a magneto-rheological brake to the joint of pneumatic muscle driven manipulator (28). The magneto-rheological brake acts as a variable damper. It is turned on (to improve position control) or off (to maintain response speed) depending on external inertia loads.

1.2.3.7. Commercially Available Pneumatic Muscles

Commercially available pneumatic muscles include the fluidic muscle from Festo Corporation, the Air Muscle from Shadow Robot Company, the McKibben muscle from Hitachi Medical Corporation, and the Rubbertuator from Bridgestone.

Festo fluidic muscle is constructed of a three-dimensional rhomboidal woven fiber mesh embedded in a rubber bladder (24). This construction improves hysteresis, nonlinearity and durability (94). Fluidic muscles are available in three basic diameters (internal resting diameters of 10 mm, 20 mm and 40 mm) and a multitude of nominal lengths (24). The maximum contraction is approximately 25% of the nominal length. The largest pneumatic muscle is capable of lifting 6000 N (1349 lb) at its maximum pressure of 600 kPa (87.2 psi). Many research groups have used Festo fluidic muscles. Some of these groups include the following: Thanh and Ahn of the University of Ulsan, Korea (28), several researchers from the State Key Laboratory of Fluid Power Transmission and Control, Zhejiang University, and Jiangsu University of Science and Technology in China whose work was supported by Festo corporation (116), Choi et al. from the Korea Advanced Institute of Science and Technology (117), Chettouh et al. from the Lab de Robotique et d'Intelligence Artificielle (118), Chang et al. from St. John’s University and the National Taipei University of Technology (119), and Xiong et al. from the State Key Laboratory of Digital Manufacturing Equipment and Technology in China (2).
The Air Muscle from Shadow Robot Company has an inner rubber tube encased in a plastic woven shell (120). Various sizes are produced. The largest model (resting internal tube diameter of 30 mm) can generate forces up to 687 N (154 lb) at its maximum rated pressure of 400 kPa.

The pneumatic muscle produced by Hitachi Medical Corporation is modeled after the McKibben muscle and is referred to as a McKibben muscle actuator. It has been used in passive walker designs (53), bipedal robots (55) (57), and an upper limb muscle suit (71).

The Bridgestone Rubbertuator was first marketed in the 1980s (121). Its structure was modeled after the McKibben muscle, but materials were chosen for improved robustness and performance. Bridgestone marketed two multi-joint robots that used Rubbertuators; a horizontal robot known as RASC and a suspended robot known as SoftArm (32). The practical use of these robots was limited, and both arms were discontinued by the 1990s. Bridgestone Rubbertuator’s are still being used today in physical rehabilitation robots (88).

1.2.4. Modeling of Pneumatic Muscle

Much of the past work on pneumatic muscles has been focused on modeling the force output and behavior accurately. A highly accurate model that can be applied to different muscles has not yet been developed. Many have worked on creating geometric models where pneumatic muscle properties are in some way based on the construction geometry. Biomimetic models describing pneumatic muscle behavior have been developed using models of skeletal muscle. Phenomenological models have used viscoelastic parameters to describe the behavior of pneumatic muscle. Other pneumatic muscle models have used fluid flow theory, neural networks and finite element theory. Special models have been developed for underwater applications.
1.2.4.1. Geometric Models

Geometric models developed in past research have been used to describe the relationship between force and pressure using properties of the pneumatic muscle construction and geometry. Gaylord in his 1958 patent, “Fluid Actuated Motor System and Stroking Device”, performed the first force analysis of the double helix braided pneumatic muscle (31). Schulte analyzed the relationship between pressure and force, and arrived at the same general expression as Gaylord (33). His static model related tensile force, $F$, inflation pressure, $P$, tubing diameter, $D$, and fiber angle, $\gamma$, according to Eq. (1) shown below (35). More specifically, the angle $\gamma$ is between the helical fibers and the longitudinal axis of the tubing, and $D$ is the tubing diameter when $\gamma$ is 90˚.

$$F = \frac{\pi D^2 P}{4} [3 \cos^2 \gamma - 1]$$

(1)

According to Davis et al., Schulte’s model was typically accurate to within 10-20% for forces over the full operating range and has since been refined by several groups. (122)

Schulte also developed an expression that included effects of tubing elasticity and internal friction as shown in Eq. (2) (35).

$$F = \frac{\pi D^2 P}{4} [3 \cos^2 \gamma - 1] + \pi D k_e \left[ L \sin \gamma - \frac{\pi \cos^2 \gamma}{\sin \gamma} (D \sin \gamma - D_o) \right]$$

$$- \pi L D P_c (u_s + u_{st}) \sin \gamma$$

(2)

The term $k_e$ is the elastic constant of inner tubing found through experiments, $L$ is the length of the device at any value of $\gamma$, $D_o$ is the diameter of tubing when not inflated, $u_s$ is the coefficient of friction between strands of the helical sleeve, and $u_{st}$ is the coefficient of friction between strands of the sleeve and inner tube. $P_c$ is the difference between $P$ and $P_i$ where $P_i$ is the pressure required to inflate the unconstrained inner tube to a diameter equivalent to device diameter at any value of $\gamma$. Due to its dependence on a changing $\gamma$, the nonlinearities of $k_e$, $u_s$, and $u_{st}$, and the experimental determination of the model parameters, this expression was recommended for use as guidance in the mechanical design of the actuator (35).
Inoue worked with the Bridgestone made Rubbertuator (121). In his work he uses an equation for contraction force of a Rubbertuator (Eq. (3)) to develop equations for the contraction forces for a pair of agonist-antagonist Rubbertuators and an equation of motion for the pair of Rubbertuators.

\[ F = \frac{\pi D_o^2 P}{4} \left[ a(1 - \varepsilon)^2 - b \right] \]  

(3)

Here \( F \) is contraction force, \( D_o \) is the effective diameter before displacement, \( P \) is the internal pressure, and \( \varepsilon \) is contraction rate (found through experimentation). The terms \( a \) and \( b \) are original constants of the actuator also found through experimentation. The agonist-antagonist pair of Rubbertuators are positioned in parallel and connected by a wire wrapped around a pulley as previously illustrated in Figure 2. In the contraction force equation for a pair of Rubbertuators, the \( P \) is replaced by a \( P_0 + \Delta P \) term. The \( \Delta P \) term refers to the difference in pressure between the actuators and \( P_0 \) is the initial pressure. The contraction rate is also slightly different and is based on not only the contraction rate of the individual muscle, but also the original length of the Rubbertuator, the radius of the pulley and the rotation around the pulley. The contraction force equations along with an expression for torque generated by the system were used in the development of Eq. (4), a second order equation of motion for the system.

\[ I \ddot{\theta} + C \dot{\theta} + K_1 \theta = K_2 \Delta P \]  

(4)

Here \( I \) is the moment of inertia, \( C \) is the viscosity coefficient, \( K_1 \) is the proportional constant to \( P_0 \), \( K_2 \) is the proportional constant to \( \Delta P \), and \( \theta \) is the rotation around the pulley. The third term in this equation, \( K_2 \Delta P \), allowed for imprecise open loop position control based on change in pressure (121).

Two approaches have been used to model force output of pneumatic muscle (123). The first approach, virtual work, is based on conservation of energy in the system and has been used by Chou and Hannaford (101), Caldwell et al. (25), Tondu and Lopez (124), and Kimura et al.
The second approach entails using detailed force analysis and has been implemented by Doumit et al. (33). Both of these approaches, with certain assumptions, result in the same base force model originally developed by Gaylord and Schulte (123). The base model produces good accuracy over a broad range of forces and motions, but errors of 20% are possible in force-displacement results. The base model assumes that the system is lossless, no energy is stored in its elastic wall, it has zero wall thickness, and it retains a cylindrical shape (33). Many groups have changed these assumptions and built upon the base model in order to increase accuracy.

Chou and Hannaford used the concept of virtual work to arrive at the base model, but they also built on the model to consider the thickness of the rubber bladder and the effects of friction within the actuator (126). The force equation considering the effects of bladder thickness is shown as Eq. (5).

\[
F = \frac{\pi D^2 P}{4} \left( 3 \cos^2 \gamma - 1 \right) + \pi P \left[ D t_k \left( 2 \sin \gamma - \frac{1}{\sin \gamma} \right) - t_k^2 \right]
\]

(5)

The force, \( F \), is now a function of not only \( D \) (diameter when \( \alpha \) is 90°), \( P \) (pressure), and \( \gamma \) (angle between a braided thread and the longitudinal axis of the bladder), but also \( t_k \) (bladder thickness) as well.

The resulting model was more accurate but also more complex so Chou and Hannaford considered a simplified model (126). The simplified model considered the pneumatic muscle to be a variable-stiffness elastic element, or “gas spring”, where the tension is a function of the pressure and length. The model took the energy stored in the bladder and mesh into consideration, and it included a nonlinear term to adjust for the non-cylindrical shape of the pneumatic muscle at extreme contraction. Parameters of the model were found through experimentation.

Chou and Hannaford studied viscous friction and Coulomb friction and concluded that the dominant source of friction in the pneumatic muscle is Coulomb friction caused by the contact between the bladder and the shell, between the braided threads and each other, and the shape changing of the bladder (126). There is force/displacement hysteresis in the actuator which
makes this friction difficult to predict, so when calculating forces they suggested using an experimentally determined force offset, added to calculate forces during muscle contraction and subtracted during extension. It has been reported that there is approximately 15% error between Chou and Hannaford’s modeled response and measured data (33).

Chou and Hannaford (101) and Sugisaka and Zhao (127) independently developed gas pressure transmission models based on gas dynamics and thermodynamics. The pneumatic muscle was modeled along with all the parts of the system that provide air pressure to the pneumatic muscle.

Klute and Hannaford attempted to improve the base model by using a nonlinear-material model developed by Mooney and Rivlin in the 1940s to characterize the bladder (128). This model included material properties of the inner bladder constrained by braid kinematics. Model validation testing showed a significant discrepancy between the predicted model results and actual results. Klute and Hannaford suggested that elastic energy storage of the system should also be considered.

Caldwell et al. showed that the force balance equation of a pneumatic muscle could also be produced using the concept of surface displacement (129). As a braided pneumatic muscle contracts or expands, the interweave angle changes and the surface area of the pneumatic muscle varies. Force is found using the surface area of the muscle, surface area of the endplates, and pressure by means of virtual work concepts (the actuator wants to maintain its minimum energy/force state) (25). The improvement to the base model comes from the incorporation of end plate effects as seen in Eq. (6) (130). When the actuator approaches its minimum length, there is a reduction in contraction force. Incorporating the end plate effects into the model captures this behavior. Eq. (6b) is the base model force expression. The term \( h \) is helical fiber length, \( N_T \) is the number of trapezoids in the fiber mesh, and \( D_{cap} \) is the end cap diameter.

\[
F = \frac{Ph^2 \cos^2 \gamma}{2N_T^2 \pi} - \frac{Ph^2 \sin^2 \gamma}{4N_T^2 \pi}
\]  
(6a)
\[ F = \pi D^2 P \left( 3 \cos^2 \gamma - 1 \right) \text{if } \gamma > \sin^{-1} \frac{D_{\text{cap}}}{D} \] (6b)

\[ F = \pi D^2 P \left( 2D^2 \cos^2 \gamma - D_{\text{cap}}^2 \right) \text{if } \gamma < \sin^{-1} \frac{D_{\text{cap}}}{D} \] (6c)

Tsagarakis and Caldwell further improved the model by incorporating the following effects. They considered the distortion effects at the ends of the pneumatic muscle (123). The area around the end-cap is not a cylinder so an integration technique was used to determine the end muscle surface area. They considered muscle wall thickness, the effects of rubber elasticity on radial pressure loss and contractile force, and the effects of radial rubber expansion before contact is made between the liner and the side walls. They also experimentally determined a constant friction force, based on Chou and Hannaford’s recommendation (126) (123). After comparing predicted model results with experimental results, it was determined that the static model achieved 30% to 50% better accuracy than that of the Chou and Hannaford model.

Davis et al. improved the base model by considering the stretch of braid fibers (32) (122). At high pressures (450 kPa), fiber strands stretch as a function of muscle pressure and muscle geometry. Incorporation of strand extension in the model reduced the error approximately from 15% to 5%. Discrepancies between the model’s predicted results and the experimental results are accounted for by including parameters specific to particular pneumatic muscles and thus this model may only be valid for their specific pneumatic muscle (33).

Tondu and Lopez modeled pneumatic muscle using the virtual work approach, geometrical relations of the braid, and classical rubber theory (39). The virtual work approach was applied to determine the transformation of circumferential forces into an axial contracting force, leading to an equation giving the cylindrical artificial muscle contracting force as a function of contraction rate. Although they used a different approach, the developed force model was equivalent to the Gaylord/Schulte base model. Tondu and Lopez improved the base model by
adding a term which accounts for the conic shape that occurs at both ends of the muscle when it contracts. The resulting force model is shown as Eq. (7).

\[
F(\varepsilon, P) = \pi \left( \frac{D_o}{2} \right)^2 P\left[ a(1-k\varepsilon)^2 - b \right] \text{for } 0 \leq \varepsilon \leq \varepsilon_{\text{max}}
\]

(7a)

\[
a = \frac{3}{\tan^2 \gamma_o}, \quad b = \frac{1}{\sin^2 \gamma_o}, \quad \text{and} \quad \varepsilon_{\text{max}} = \frac{1}{k} \left( 1 - \sqrt{\frac{b}{a}} \right)
\]

(7b,c,d)

In these equations, \(D_o\) is the initial diameter, \(\gamma_o\) is the initial weave angle, \(k\) is a constant accounting for end deformation, and \(P\) is internal muscle pressure. Hysteresis effects due to friction within the braided mesh and friction between the bladder and braided mesh were not considered in this model, which was the explanation behind poor accuracy results at low pressures.

In an attempt to include friction effects not previously included in their model, Tondu and Lopez modeled friction as a function of fiber contact surface and muscle pressure instead of simply adding an offset force like Chou and Hannaford (124). The model was more accurate than that of Chou and Hannaford, but it still relied on experimental data (33).

Kimura et al. developed a model of pneumatic muscle by using the virtual work developed base model equation and incorporating effects of elasticity, viscosity and frictional force of rubber (125). An energy loss term was added to account for energy losses due to elasticity of rubber and friction forces between fibers and between the rubber and fiber layer. Elasticity of the rubber was modeled as a spring-damper system.

Colbrunn et al. built on work published by Chou and Hannaford as well as Tondu and Lopez. They incorporated end effects, or changes in force output at the length limits of a pneumatic muscle (131). At the maximum length, an increase in length would cause the braids to stretch. The stiffness of the braids is high in comparison to the rubber, so this stretching was modeled as a spring of very high stiffness. This model does not consider bladder thickness,
friction within the actuator, or non-linear elastic energy storage of the bladder as other models have.

Zhou et al. attempted to improve the base model using rules of elastic theory to analyze the force properties of the rubber bladder (132). The model is shown below as Eq. (8). The coefficient $K_B$ is a compensation term based on the braid’s elasticity. $F_f$ is a compensation term related to Coulomb friction. Viscous damping was shown to be negligible. $K_e$ is a constant based on the rubber tube’s material properties. $N$ is number of rolling fiber surrounding the tube. The $L$ and $L_o$ terms are length of rubber tube and original length of rubber tube, respectively while $h$ is length of fiber surrounding rubber tube.

$$F = K_B \left[ \frac{P - K_e (L_o - L)}{4\pi N^2} \left(3L^2 - h^2\right) \right] - F_f$$  \hspace{1cm} (8)

Kothera et al. looked at the base Gaylord model and evaluated many different additional considerations including elastic energy effects, braid effects, effects of non-cylindrical ends, effects of non-constant tube thickness, and pre-stretch effects (133). The correction of the elastic energy term was shown to improve the model significantly. The inclusion of braid effects showed a slight improvement for high pressures and large pneumatic muscles. The effect of non-constant thickness was shown to considerably improve model prediction.

Doumit et al. used force and stress analyses to derive an improved static model that does not depend on experimentally determined parameters valid only for a specific pneumatic muscle (33). According to Doumit, the base model (Gaylord/Schulte force expression) is only valid for specific muscle conditions that are rarely true. End cap diameter effects, mechanical properties of the bladder, frictional losses due to fiber-to-fiber contact within in the braided mesh, and netting analysis of the braid were considered in the model. Mechanical property testing results showed that hysteresis, fiber elongation, and material properties along the longitudinal axis of the bladder did not need to be included in the model.
1.2.4.2. Biomimetic Models

Comparisons have been made between biological skeletal muscle and pneumatic muscle; both contract and expand radially after activation resulting in a tensile force. Both have nonlinear dynamic behavior of force and position (22). The pneumatic muscle models based on models describing biological skeletal muscle are classified as biomimetic models, biomimetic referring to the application of biological principles to design.

Amy Neidhard-Doll’s work focused on the development, implementation, and validation of a biomimetic phenomenological model for pneumatic muscle (100). The biomimetic approach models the pneumatic muscle by revising the Hill muscle model describing an isometric force twitch contraction of skeletal muscle to include energetic and viscoelastic parameters. The energetic parameter refers to the chemo-mechanical energy conversion and the viscoelastic parameter refers to the internal-element stiffness variation (134). This model was an improvement over existing biomimetic models in that it accounted for spatiotemporal recruitment and neuromuscular control strategies employed by the peripheral nervous system (100). The model was used to create a software controller for open- and closed-loop isometric force control of pneumatic muscle.

Work by Klute et al. focused on developing a model for an artificial muscle-tendon system driven by pneumatic muscle to predict expected force-length-velocity performance of the system (115). Static and dynamic properties of biological muscle and tendon (primarily Hill’s phenomenological model) were taken from literature to design the system and describe the force, length and velocity relationships. The force-length properties of pneumatic muscle are muscle-like, while the force-velocity properties are not. A hydraulic damper was added to the system to better replicate biological muscle force-velocity properties.
1.2.4.3. Phenomenological Models

Phenomenological models aim to better capture the dynamic behavior of pneumatic muscles as compared to the static behavior upon which geometric models focus.

Colbrunn et al. developed a pneumatic muscle model consisting of a spring, viscous damper and Coulomb friction element arranged in parallel (131). The spring represents the nonlinear force-length relationship while the viscous damper represents the fluid flow losses. The Coulomb friction element represents effects of friction between the internal bladder and outer braided shell. Tests showed that this model has potential for reasonably predicting the dynamic behavior of pneumatic muscle.

Repperger et al. developed a pneumatic muscle model consisting of a spring element, viscous damping element, and contractile force element arranged in parallel (21). The spring coefficient was taken to be a function of position and the damping coefficient a function of velocity. The contractile force element varied with internal pressure. In 2003, this model was improved and characterized by Reynolds et al. (18). In 2007, Serres used this model to characterize a commercially available pneumatic muscle (135).

Balasubramanian et al. believed that the model of (18) was incomplete because the model parameters were not functions of both input pressure and external load and because the model did not account for all the factors that can affect a pneumatic muscle’s dynamic behavior such as magnitude of contraction, damping ratio of rise response, natural frequency of rise response, damping ratio of decay response, and natural frequency of decay response (136). These parameters were modeled as quadratic functions of external load for a given input pressure. In Serres’ characterization work, she addressed the issue of the phenomenological model parameter dependence on pressure as well as external load (19). Ranges of external loading for which the spring coefficient could be considered a constant were found for each pressure level.
1.2.4.4. Other Models

A unique pneumatic muscle model was developed by Hildebrant et al. Here the pneumatic muscle is considered to be a one-way cylinder with a virtual flexible diameter as a function of the contraction, which moves against a spring with a contraction dependent on spring tension (137). The force model is shown as Eq. (9).

\[ F(p, x) = p \cdot A(x) - F(x) \]  \hspace{1cm} (9)

The variable \( x \) refers to pneumatic muscle contraction, \( A(x) \) is the virtual piston area and \( F(x) \) is the tension force of the spring. The term \( p \) is absolute pressure inside the muscle. \( A(x) \) is a second order polynomial. \( F(x) \) is estimated by a 3\textsuperscript{rd} order polynomial. Parameters of the model are identified using optimization algorithms to minimize the mean square error between analytical and experimental force results.

One dimensional flow theory was used by Prior and White to model the dynamic behavior of a pneumatic muscle attached to a robot arm (138). Ahn and Anh have used several types of neural network models to model the behavior of pneumatic muscle. The advantage to these neural networks is that they can take a model with a high level of uncertainty and/or error, and through adaptation, change it into a more accurate model which can be readily incorporated into a control system. The neural network models developed include a nonlinear autoregressive with exogenous input (NARX) model (139), an adaptive recurrent neural network model (140), and an inverse double nonlinear autoregressive with exogenous input fuzzy model (141).

Finite element methods (FEM) have been used by several groups to model pneumatic muscles, particularly for the purpose of designing an ideal pneumatic muscle. Bertetto and Ruggiu used FEM to predict the behavior of different types of pneumatic muscles (helical braided and straight fiber) (104). Zhou et al. used virtual work principles and total Lagrange formulation to derive the finite element internal force vector and tangent stiffness, which were then combined with other effects such as inertia, damping, internal strain, and external forces in a general finite
element formulation (142). No physical model was built to validate the results. Zhang et al. used geometrically nonlinear anisotropic membrane elements to simulate the outer braided layer of a pneumatic muscle (143). Fiber friction and fiber slippage were not considered. A quasi-static model validation was completed by comparing finite element force results with theoretical solutions for different internal pressures. Kim et al. used finite element modeling and analysis to design a pneumatic muscle with the proper contraction ratio and force (144).

A special model has been developed to describe pneumatic muscle behavior under water. The model, like many of the geometric models previously described, was developed using the principle of virtual work (conservation of energy) and geometric properties of the pneumatic muscle (145). In the underwater force model, Eq. (10), $P'$ is internal gas pressure relative to water pressure and $P_r$ is the pressure to overcome the elasticity of the bladder for expanding. $L_o$ and $L$ are the lengths of the original and inflated muscle, respectively. $N_{TX}$ and $N_{TY}$ are the numbers of trapezoids on the muscle in the X and Y directions, respectively. $K_L$ is the contraction resistance constant and $K_E$ is the contributed erosion factor of the latex bladder.

\[
F = \frac{(P' + P_r)L_o^2 N_{TX}^2}{\pi} \left[ 3 \left( \frac{L}{2LN_{TY}} \right)^2 - 1 \right] \tag{10a}
\]

\[
L = \sqrt{\left(\frac{F \pi}{(P' + P_r)L_o^2 N_{TX}^2} + 1\right)\frac{4L_o^2 N_{TY}^2}{3} + K_L + K_E}. \tag{10b}
\]

1.2.5. Control of Pneumatic Muscle Systems

The control of pneumatic muscles is complicated due to the inherent nonlinearity of pneumatic muscle dynamics. Control methodologies applied to pneumatic muscle and systems activated by pneumatic muscles have been reviewed. The control methods used by researchers have been broken up into fifteen categories. Examples of each type of control applied to a
pneumatic muscle system are listed. There is some overlap between the uses of these methods. Several researchers have applied multiple control concepts to their systems.

1.2.5.1. Open Loop Control of Pneumatic Muscle Systems

The first control category to be discussed is open loop control. In this type of control, no signals are fed back into the control system. It relies on the developed model to produce adequate results. For simple, accurate models or systems that do not require precise control, open loop control can successfully be used. Serres used open loop control to demonstrate a resistance exercise task (135). The resistance was provided by a pneumatic muscle while a servo motor represented the human operator.

1.2.5.2. P, I, and D Control of Pneumatic Muscle Systems

The second control category to be discussed is PID (proportional integral derivative) or any combination of P, I, or D control. A proportional controller makes a change to the output that is proportional to the current error value (error is multiplied by a proportional gain) (146). This helps to decrease rise time and steady state error. An integral controller sums the instantaneous error over time, multiplies it by an integral gain, and adds it to the controller output. This accelerates the movement of the system towards its set point, decreasing rise time and steady state error. A derivative controller multiplies the rate of change of the error by a derivative gain, slowing the rate of change of the controller output and decreasing overshoot.

Colbrunn et al. developed a four degree of freedom walking robot where position and passive stiffness were controllable at each joint (43). A high level controller generated joint trajectories, performed inverse kinematics on their model and modulated control gains as a function of the sensory feedback signals. A low level controller made decisions based on desired values given by the high-level controller and performed proportional control on the pressure output signals communicated to valves.
Van Damme et al. developed an assistive pneumatic manipulator to help with handling heavy loads (74). The manipulator was an inverse elbow configuration (two joints and two links each with an agonist-antagonist parallel pair of pneumatic muscles). The difference in pressure/force output of the two pneumatic muscles in each pair produced angle change and torque for the joint. The angle of each joint was controlled using a $\Delta P$ approach to reduce the number of calculated outputs. Instead of two pressure inputs, a pressure difference, $\Delta P$, is added to one muscle and subtracted from the other to control position. A PID controller is used in this system to correct disturbances and model inaccuracies. Stability was demonstrated but control performance with regard to settling time and overshoot was poor.

Tsagarakis and Caldwell developed a seven degree of freedom upper arm training/rehabilitation system (147). Each joint is activated using an agonist-antagonist pair of pneumatic muscles connected by a cable around a pulley. Joint torque control was implemented at each joint. The torque control loop used the torque error (based on torque feedback from a torque sensor on the joint) to calculate the required amount of pressure change in each muscle pair using a PID control law. The coefficients of the PID law were experimentally estimated. This control system improved torque response.

Xiong et al. developed an exoskeleton robotic arm for stroke rehabilitation (2). The robotic arm had two modes; one for providing pre-specified resistances on the patient’s arm and one for allowing the patients to generate the trajectory of the device for functional training. The control system was able to switch between position, force, force/position and impedance based on the desired behavior of the system. A feedback and feedforward PID controller was used to calculate the desired outputs of the robotic arm. A proportional-derivative (PD) controller was used in position control to calculate the outputs of the robotic arm. The position control method was used to drive the arm of the patient back and forth repeatedly until a stop command is received. The force control method was used to apply pre-specified resistant torques to the joints. Force/position control was used to deal with an abnormal event such as a spasm. The system was
designed to continue training if a mild, instantaneous spasm occurred. The provided force was
designed to drop to zero if a sustaining spasm occurred. Impedance control was used for
functional training, when the human drove the trajectory.

Noritsugu and Tanaka developed a two-joint, three-link rehabilitation robot (148). The
joints were driven by agonist-antagonist pairs on two of the links. The angle position was
controlled with feedforward compensation and PI and PD controllers. The proportional gain was
determined by the relation between pressure and measured displacement of the pneumatic
muscle, taking gravity into consideration. The nonlinearity between the controller’s commanded
pressure and the resultant pressure was compensated for by using slopes evaluated at desired
pressure values on a static characterization approximation curve. Control gains were found
through trial and error. The control scheme was considered successful, but not highly accurate.

Noritsugu and Tsuji developed a two joint, three link task assist system (73). The task
assist system was in reality a manipulator with a fixed proximal link and two distal links, each
having an agonist-antagonist pneumatic muscle pair. The force and position of the manipulator tip
were controlled. The control system used a calculated external load force (based on the task), a
force generated by the human, and proportional and integral gains in the forward part of the force
control loop. A derivative feedback gain acting on the operational force was used in the feedback
loop. Force control deviations were controlled to become zero. Both desired position and position
at the end of the manipulator were used for the internal position control system where position
control deviations were driven to zero.

Costa and Caldwell developed a lower limb exoskeleton for force augmentation and
active assistive walking training powered by pneumatic muscles (14). A mathematical model
relating positions, velocities and accelerations of the links making up the exoskeleton structure
was developed. The angle of each joint of the structure was actuated by an agonist-antagonist
pair. Each muscle was controlled by three PID controllers (two low-level ones for pressure and
one higher-level one for position/torque). The control performance was less than what would be
acceptable for a knee prosthesis, but muscle efforts, ability to handle load disturbances, and human gait tracking/guiding capabilities were demonstrated with a healthy individual.

1.2.5.3. Fuzzy Control Applied to Pneumatic Muscle Systems

The third control category to be discussed is fuzzy control. Fuzzy control can be used to overcome poor models or models with a lot of uncertainty because it provides a way to arrive at a definite conclusion using vague, ambiguous, imprecise, noisy, or missing input information (149). The approach is not mathematical and mimics how a person would make decisions.

Chan et al. designed a position controller for a simple pneumatic muscle system consisting of a mass hanging from a vertically oriented pneumatic muscle (150). The controller was designed to track length of the pneumatic muscle and position of the mass using a PID controller with fuzzy proportional and derivative parts and a non-fuzzy integral. The fuzzy inverse model incorporated internal pressure to help closed-loop performance and had a learning ability that improved closed loop performance over time by using measured data from the operating system to update the controller. Tracking was found to be accurate after a few seconds.

Chang and Lilly designed a controller for the same simple pneumatic muscle, hanging mass system described above in order to force the pneumatic muscle length to follow a reference signal (track position) (151). A fuzzy P+ID controller was constructed (P+ representing an incremental fuzzy logic controller in place of the proportional term). Control parameters were tuned using an evolutionary algorithm, which optimized membership functions in the fuzzy system according to some performance criterion. Excellent tracking performance was accomplished with an actual pneumatic muscle after the evolutionary algorithm tuned the controller parameters.

Bugarski et al. designed a position control system for a nozzle with three degrees of freedom using MATLAB’s Fuzzy Inference System (98). The fuzzy control system (with no
feedback) was implemented using a Programmable Logic Controller unit and proportional pressure regulators.

Balasubramanian and Rattan developed two control systems using fuzzy logic to track position for a simple pneumatic muscle simulated system (vertical pneumatic muscle with hanging weight) (152). The three-element phenomenological model developed by Repperger (21) and improved by Reynolds (18) was used for this research (152). A fuzzy inverse dynamics controller whose inputs were position and velocity was developed to determine pressure based on the given force command. The fuzzy rules for this controller were obtained from training data using a least square errors method, which produced better results compared to a weighted average method. The developed fuzzy inverse dynamics model was used as part of a feedforward controller accompanied by a feedback controller to eliminate tracking error and improve stability. The developed fuzzy inverse dynamics model was also used as part of linearizing control scheme (153). The linearizing control scheme consisted of a model-based portion, which acted to cancel out the system dynamics by feeding back the dynamic terms of the system, and a servo portion, which tracked a trajectory by providing the necessary acceleration. The model-based portion was implemented using fuzzy logic. A force input, normalized to the desired range, was fuzzified and sent to an inference engine which derived the output result. The result was then defuzzified, scaled back to normal range, and sent through another inverse fuzzy model to give a pressure command for the muscle based on the input force command. Tracking errors were reduced using proportional gains on the position and velocity feedback included in the servo-portion of the control system.

Fixed structure fuzzy models such as these can be imprecise due to nonlinearities and disturbances. A change in system dynamics can leave them ineffective. An adaptive fuzzy algorithm was used by Balasubramanian to tune the fuzzy models over time in case a change in system parameters occurred (154). The adaptive part of the control system continuously
monitored the system and updated the fuzzy model weights, eliminating steady state position error and velocity error during the latter phase of the transient response.

Chang et al. designed a self-organizing fuzzy control system for an agonist-antagonist pair of Festo fluidic muscles connected by a cable around a pulley (119). As previously discussed, fixed fuzzy rules produce steady state errors. Self-organizing fuzzy control has a learning mechanism to generate and modify fuzzy rules. Experimental results verified that the learning mechanism can improve steady state tracking error and transient response.

1.2.5.4. Learning/Adaptive Control of Pneumatic Muscle Systems

The fourth control category to be discussed is learning or adaptive control. This type of method modifies the controller over time to better handle uncertain or time-varying system parameters or disturbances.

Zhu et al. have developed posture trajectory tracking control strategies for a parallel manipulator driven by three vertically oriented pneumatic muscles (155). In this manipulator, one end of the pneumatic muscles was fixed to a base while the other end was attached to a moving platform. In 2007, this group developed a nonlinear pressure observer based adaptive robust controller. The pressure observer was used to estimate unknown pressures while the adaptive robust controller effectively compensated and attenuated uncertainties (parametric and nonlinear uncertainties coming from the model, dynamics uncertainties coming from pressure estimation errors). Experiments showed that the system had good control accuracy, smooth movement, and robustness to disturbances. In 2008, this group developed an adaptive robust controller that used a parameter estimation algorithm based on composite error minimizing criterion (156). The parameter estimation algorithm was a complex method by which unknown parameters of the model were estimated. Experiments showed that the control system achieved precise posture trajectory tracking.
Tonietti and Bicchi applied independent joint position and stiffness adaptive control to a simple manipulator consisting of an agonist-antagonist pair of pneumatic muscles connected by a cable around a pulley (157). Simulations showed that adaptive control, which can better cope with model uncertainties, had better trajectory performance than PID control, particularly when trajectory deviations were high.

Ruihua et al. developed an adaptive control strategy combining PID-based feedback with an Iterative Learning Controller (ILC) to control the five degree of freedom robot named RUPERT (Robotic Upper Extremity Repetitive Therapy) (85). The control system consisted of two loops; an outer loop and an inner loop. The outer loop generated the trajectory command for the inner-loop. The inner loop consisted of five independent joint controllers, one for each degree of freedom. Three of the joint controllers (shoulder flexion/extension, elbow flexion/extension, and humeral rotation) had an ILC in parallel with the PID feedback controller. The ILC improved performance by using error information over several iterations to update the feedforward control commands. This method also saved time because PID parameters did not have to be tuned for every degree of freedom and for different subjects reaching different target locations or moving through different specified trajectories. A fuzzy rule-base was used to estimate the ILC learning rate. Rules for the fuzzy rule-base were selected to ensure that the ILC learned only underlying nonlinearities of the plant and not unwanted disturbances. The controller was found to have consistent performance for subjects performing different reaching tasks, but there was jerkiness at the shoulder.

1.2.5.5. Neural Networks Applied to Pneumatic Muscle Systems

The fifth control category to be discussed is neural networks. Neural networks are modeled after how information is processed in the brain (158). They can be used to model high or low level reasoning and are built from computational elements called nodes. In general, each node receives inputs from other nodes or external sources and has an associated weight relative to
a bias node, which is modified as part of neural network training. The output of each node can serve as inputs to other nodes. The neural network is formed by connecting multiple nodes and typically has a layered structure; a layer of input nodes connected to a layer of output nodes by a layer of weights. There are also hidden layers depending on the problem to be solved. Different methods are used to train neural networks. One of the most common methods, error backpropagation, provides a way to train networks with any number of hidden units arranged in any number of layers.

Fife et al. developed a control system using neural networks and genetic algorithms to track position of a five degree of freedom robot leg activated by pneumatic muscles (63). The inputs of the model were joint angles and valve states including intake, exhaust and closed. Controlling the valve states was the method by which pressure inside the pneumatic muscles was controlled. The output of the model was force (based on length and pressure) generated by the actuator. The developed continuous time recurrent neural network controller was numerically integrated alongside the model so that they could communicate with one another. The control system was capable of dealing with inaccurate signals, and it had robust behavior and resistivity to noise. The neural network was trained using a genetic algorithm to arrive at a better controller. The genetic algorithm worked by producing a new generation of possible controllers whose performance was evaluated and assigned fitness based on how well the behavior matched the commanded behavior. The controllers which produced good results were chosen to make more offspring in the next generation, and the process continued from there.

Thanh and Ahn used nonlinear PID control with neural networks to control joint position of a simulated three-link manipulator driven by pneumatic muscles (159). Two of the manipulator links were equipped with an agonist-antagonist pneumatic muscle pair. The controller parameters were adapted iteratively within the neural network using the input and output of the plant and the conventional backpropagation algorithm. The PID gains were nonlinear functions of error and
were tuned using the steepest descent method. The conclusion was that PID with neural networks performed better than conventional PID control.

Tian et al. applied neural networks for joint angle tracking control of an agonist-antagonist pneumatic muscle pair connected by a cable around a pulley (160). A recursive prediction error algorithm was applied to train the neural networks instead of the conventional backpropagation algorithm, producing faster convergence. Experimental results showed faster response and higher control accuracy compared to a traditional linear control scheme. However, it was necessary to train the neural networks thousands of times.

Ahn and Anh have developed a joint angle position controller for a 2 joint manipulator driven by pneumatic muscles (140). The manipulator was constructed of three links, two of which had an agonist-antagonist pair. An adaptive recurrent neural network controller updated the weights of the manipulator’s inverse characteristics and applied them to the inverse dynamic model to generate an appropriate voltage control signal. The adaptive recurrent neural network controller was unique in that it had a dynamic self-organizing structure, fast learning speed, and good generalization and flexibility in learning. Simulations indicated that this controller was superior to a conventional PID controller.

1.2.5.6. Variable Structure Control of Pneumatic Muscle Systems

The sixth control category to be discussed is variable structure control. In this form of control, the state feedback control law is not a continuous function of time (161). The structure of the control law varies based on the position of the state trajectory. As the state crosses each discontinuity, the structure of the feedback system is altered or switched. A variable structure control design has switching surfaces, usually fixed hyperplanes passing through the state space origin. The state trajectories may switch between switching surfaces at high-speed based upon some switching criteria (162). The controller’s switching surfaces are designed so the nonlinear state trajectory always moves toward a switching condition. In other words, the controller is
designed to drive the trajectory onto the switching surface. The intersection of or boundary between the switching surfaces forms the sliding surface (161). Once the state starts sliding, the controller must only maintain the state on or near the sliding surfaces. Sliding mode control, a form of variable structure control, helps to keep the trajectories sliding along the boundaries of the switching surfaces. A side effect of sliding mode control is chattering about the switching surface at high frequency. Chattering can excite high frequency, non-modeled dynamics in the system and lead to degradation in performance.

Repperger et al. designed position tracking control of an agonist-antagonist pair of pneumatic muscles using a variable structure controller (21). The previously described three-element phenomenological model was used. The variable structure controller was designed to drive the state space trajectories to the sliding surface. Switching conditions were based on two Lyapunov function surfaces; one surface defined by an inflation Lyapunov function and the other defined by a deflation Lyapunov function.

Lilly and Quesada used a two-input sliding mode controller for tracking of a two link, one joint, planar arm under load actuated by an agonist-antagonist pneumatic muscle pair (163). The control law is designed to determine the pressures necessary to produce certain arm torques which force the end effector to follow a desired path.

Chettouh et al. applied a generalized variable structure algorithm to reduce chatter of a position controller for a three degree of freedom robot arm actuated by pneumatic muscles (118). The generalized variable structure algorithm yielded a control law which linearized the dynamic state feedback. The control system eliminated chattering in the steady state, and it improved transient response.

Chettouh et al. also developed a high-order sliding mode controller for their three degree of freedom robot arm (164). This controller acted on the higher time derivatives of the sliding constraint instead of acting on its first time derivate. Chatter being driven back onto the higher time derivatives of the constraint led to attenuation of the chattering in steady state.
Chettouh et al. developed a third variation of variable structure control, a hybrid classical variable structure-PID controller (165). This approach effectively reduced chatter, but its parameters needed to be finely tuned to gain better results, resulting in the loss of robustness.

Chang and Yuan designed an adaptive self-organizing fuzzy sliding mode controller for a three degree of freedom rehabilitation robot actuated by pneumatic muscles (166). The goal of the controller was to improve spatial tracking performance. The self-organizing learning mechanism was developed from the fuzzy sliding surface. The fuzzy sliding surface reduced the required number of fuzzy sets and fuzzy rules. The adaptive self-organizing fuzzy sliding mode controller was found to have less tracking error than a fuzzy sliding mode controller. Experiments showed angle tracking errors within 0.7°.

Ahn and Tu designed a switching algorithm using a learning vector quantization neural network to improve position control performance of an agonist-antagonist pneumatic muscle pair acting against different inertial loads (167). When the inertial load of the system changes quickly, deterioration of the system transient response can occur. The neural network is applied as a supervisor of the switching controller by recognizing the condition of the inertial load. Results showed that the steady state error was reduced to within 0.1°.

### 1.2.5.7. \( H_\infty \) Control of Pneumatic Muscle Systems

The seventh control category to be discussed is \( H_\infty \) control. In this type of control, the control problem is set up as a mathematical optimization problem, and the controller is the solution to the optimization (168). The advantage of this type of control is that it is able to deal with errors in the plant model and unknown disturbances. The disadvantages are that a high level of mathematical understanding is needed in order to apply it and a reasonably good model of the system is needed.
Djouadi et al. developed a robust $H_\infty$ position tracking controller for a vertically oriented pneumatic muscle with attached mass (169). The feedback $H_\infty$ controller scheduled on pressure optimally rejected external disturbances and coped with dynamic uncertainty. The overall tracking performance was excellent, but tracking inaccuracy did occur. They attributed this to the fact that only one pneumatic muscle was lifting the weight vertically, with no opposing force from another pneumatic muscle. Another group, Osuka et al., effectively applied $H_\infty$ control to a Rubbertuator pair (170).

1.2.5.8. Gain Scheduling Applied to Pneumatic Muscle Systems

The eighth control category to be discussed is gain scheduling. This approach breaks up a nonlinear control problem into several linear problems, allowing well established linear control methods to be applied to a nonlinear problem (171). Each linear controller is designed to provide satisfactory control for a different operating point of the nonlinear system.

Repperger et al. used a gain scheduling approach for force regulation of a pneumatic muscle (172). The previously described three-element phenomenological model was used here. A set of operating points was developed based on the dynamic operating range of the pneumatic muscle. The spring and damping coefficients of the model at each operating point were found through experimentation. Linear approximations passing through the origin were developed for each model coefficient and a table of scheduled gains was determined from this data. The tracking paradigm used here consisted of a triangular wave desired force input signal. It was determined that time delay and nonlinear effects were missed in the linearization, but reasonable force tracking was accomplished.

Martinez et al. used PID control and gain scheduling in the design of a five degree of freedom force amplification upper limb skeleton intended to work in collaboration with the human arm (72). Using a simplified two degree of freedom setup, a position PID algorithm was
tuned in different angle ranges and a gain scheduling strategy was implemented. Very little detail was given regarding their control feedback design or gain scheduling approach.

1.2.5.9. Feedback Linearization Applied to Pneumatic Muscle Systems

The ninth control category to be discussed is feedback linearization and its use in nonlinear control systems. In this approach, nonlinear system dynamics are algebraically transformed into linear ones, so that linear control techniques can be applied (173). Linear approximations of the dynamics are not made. The approach helps to reduce uncertainty and stabilize unstable systems (174).

Kimura et al. applied a feedback linearization technique to an agonist-antagonist pneumatic muscle pair with attached mass in order to control force (175). Their feedback linearization technique included step-type disturbance rejection. The step-type disturbance was considered to be hysteresis caused by static friction between the fiber mesh and rubber tube. This work verified that linearizing control is effective for pneumatic systems and that the proposed feedback linearization with disturbance rejection worked effectively.

1.2.5.10. Pole Placement Applied to Pneumatic Muscle Systems

The tenth control category to be discussed is the pole placement technique, where the control system’s pole locations are chosen to obtain desired performance. Pole placement control uses state feedback to generate the proper command input. Caldwell, Medrano-Cerda, et al. used this technique in the mid-1990s to control an elbow joint actuated by a pair of pneumatic muscles. In 1994, adaptive pole-placement controllers were used to control position (129). Accuracies of 1° were reported for constant set-points. However, the system response was very slow. In 1996, adaptive pole-placement techniques were again used to control joint position, and force was controlled using a PID controller (176). Position accuracy of ±0.5 degrees was reported at pressures up to 800kPa.
1.2.5.11. Impedance Control Applied to Pneumatic Muscle Systems

The eleventh control category to be discussed is impedance control, an approach common in robotics which is used to control the dynamic interaction between a robotic manipulator and its environment (177). Impedance control does not attempt to track position and force trajectories. It attempts to regulate mechanical impedance according to a specified target. Mechanical impedance is the ratio of applied force to resulting velocity at the point of applied force. Therefore, the appropriate impedance depends on the application. Impedance control is appropriate to use for robots that must interface with humans because there is a focus on controlling the interaction between the robotic manipulator and the human.

Noritsugu and Tanaka used an impedance control strategy in a pneumatic muscle manipulator rehabilitation robot (148). Four different motion modes were made possible with the impedance control system. Also, mechanical impedance of the human arm was used as a tool to assess the physical condition of the patient so that impedance parameters of the controller could be modified appropriately.

Caldwell et al. developed powered exoskeletons driven by pneumatic muscle for upper and lower body rehabilitation (11). An impedance control scheme allowed for the execution of complex assistive/resistive exercises.

Xiong et al. used impedance control in their exoskeleton robotic arm for stroke rehabilitation (2). Impedance control was used when the robot was interacting with the human, particularly during functional training when the human was trying to eat or grasp objects.

1.2.5.12. Model Predictive Control of Pneumatic Muscle Systems

The fourteenth control category, Model Predictive Control (MPC), is a method often used in process control. It uses a model of the process to predict future outputs and attempts to bring the predicted output as close as possible to the reference signal by minimizing an error function.
between the reference and predicted output a model. Schindele and Aschemann used model predictive control to control two sets of pneumatic muscles driving a high speed linear axis (178).

1.2.5.13. Proportional Myoelectric Control of Pneumatic Muscle Systems

The thirteenth control category to be discussed is proportional myoelectric control. Here the input command signal is based on processed muscle electromyography (EMG) signal activation patterns. Ferris et al. used proportional myoelectric control to regulate the pressure of pneumatic muscles actuating a powered ankle-foot orthosis (76). Gordon and Ferris used this type of control to regulate air pressure of pneumatic muscles actuating a lower limb robotic exoskeleton used to study human locomotor adaptation to disrupted muscular coordination (81). Sawicki and Ferris used proportional myoelectric control to dictate the timing and magnitude of pneumatic muscle forces produced in a powered knee-ankle-foot orthosis (10).

1.2.5.14. Biomimetic- Skeletal Muscle Control of Pneumatic Muscle Systems

The fourteenth control category to be discussed is biomimetic control based on the behavior and properties of skeletal muscle. Neidhard-Doll et al. developed a biomimetic controller based upon the physiological spatiotemporal strength-recruitment strategies characteristic of motor units within the biological peripheral nervous system in order to track isometric force (100).

1.2.5.15. Biomimetic- Cerebellar Control of Pneumatic Muscle

The final control category to be discussed is biomimetic control based on the behavior of the cerebellum. Lenz et al. implemented cerebellar function into the control system of a robot eye actuated by pneumatic muscles (179). Eskiizmiriler et al. developed a motion control system for a simple robot arm based on the connectivity of the cerebellar cortex (40). An artificial neural network controller with an architecture based on the connectivity of the cerebellar cortex was
trained using various biologically plausible learning schemes. After learning, the control system was able to accurately control velocity and position of the robot arm.
2. SIMULATED TASK: ISOKINETIC KNEE EXTENSION STRENGTH TRAINING

2.1. Isokinetic Knee Extension Background

The knee joint is formed by the articulation of the distal femur and the proximal tibia. It is capable of extension and flexion in the sagittal plane (180). Knee extension (straightening of the leg) occurs through contraction of the quadriceps femoris muscles (referred throughout this document as quadriceps). Knee flexion (bending of the knee) occurs through contraction of the hamstring muscle group with assistance from the gastrocnemius muscle. The plantaris muscle, which passes the posterior aspect of the knee, is an insignificant contributor to production of knee flexion due to its small cross-sectional area.

In a knee extension exercise, the upper leg remains stationary while the lower leg is mobile and moves through its range of motion. The goal is to strengthen the quadriceps muscles, and there is potential for improvements in activities such as walking, jumping, sprinting, bicycling, kicking, swimming, and rising from a seated position.

There are typically two training modes for strengthening exercises; concentric and eccentric. During concentric muscle action, muscle tension increases while the origin and insertion points of the muscle approach each other. In a knee extension exercise, the concentric phase occurs while the lower leg is extending and the quadriceps muscles are shortening. During eccentric muscle action, muscle tension increases while the origin and insertion points of the muscle move away from each other. In a knee extension exercise, eccentric muscle action can occur (if properly loaded) while the lower leg is flexing and the quadriceps muscles are lengthening. Both concentric and eccentric strength training is important for improvements in
strength and activities of daily living and thus will be incorporated into a device with PMA produced resistance.

There are many types of strength training including isometric (constant muscle length), isotonic (constant force), variable resistance (variable force), and isokinetic (constant velocity). In isokinetic strength training, a limb moves about its joint at a constant angular velocity despite changing force/load requirements (181). Correspondingly, the muscle producing the torque about the joint that results in the limb rotation is changing length at a constant rate. After studying both dynamic constant resistance training and isokinetic training, it was found that isokinetic training of the knee extensors and flexors produces higher isokinetic and isometric strength increases (including isometric knee extension and flexion force and isokinetic peak torque) (182). This is most likely due to the fact that the muscle undergoes a maximal load during the entire range of motion (even at extremes of joint motion when muscle is at physiological and mechanical disadvantage). Loading during isokinetic training will change as a function of limb position. Higher loads will occur at a limb position that corresponds approximately to optimal muscle length of the major muscle group producing the action. The muscle’s force output potential peaks at optimal muscle length when maximally activated. Lower isokinetic training loads will occur away from the optimal muscle length, where the potential muscle force is reduced. In reality, the relationship between limb position, muscle length and torque is more complicated than this generalization because of the biomechanics of the structures creating movement of the limb. However, this concept explains how optimal strength increases can potentially occur during isokinetic exercise, even in other activities involving isometric, isotonic and variable resistance (183). The added benefit of isokinetic exercise is that more work can be performed in the same amount of time because the muscle is working optimally through its entire range of motion. Increases in lean body mass and decreases in percent fat resulting from isokinetic type exercise are of the same magnitude as other types of training (183).
2.2. Need for Accommodation of Different Neuromuscular Behaviors

Humans display many behaviors such as physiological fatigue and erratic volitional control when performing strength training for either exercise or rehabilitation. Fatigue and erratic volitional control will be simulated using the HQDS in order to test PMA control operation and accommodation.

Physiological fatigue is defined as a decrease in the maximum force or torque generating capability of the muscle (184). Studies have shown that mechanical output during concentric contraction appears as two phases (185) (186). The first phase, fatigue phase, is characterized by a steep decrease in force/torque output. The second phase, endurance phase, is characterized by no additional decrease in the force/torque output. For a knee extension, the peak torque decreases over the first forty to fifty contractions, and then the endurance phase is reached. Studies have shown that peak torque reduces between 44 and 54% over the course of fifty isokinetic knee extensions (185) (187). The so-called fatigue and endurance phases are not seen during eccentric contraction (186). Mechanical output typically decreases linearly with each repetition and at a slower rate than concentric exercise. The level of torque reduction and the rate at which torque output decreases varies from person to person. There is a potential benefit of creating a resistive training device that accommodates different individuals undergoing physiological fatigue during concentric and eccentric exercise.

Humans undergoing rehabilitation may have trouble moving at a constant speed. They may be weak, feel pain, or become distracted resulting in a velocity change and/or drastic activation level change during the course of the task. They may display erratic volitional control of their lower leg. Different humans may also use different neuromuscular strategies in completing the same task (different acceleration, deceleration, and velocity patterns) (188). This may also result in a velocity change and/or drastic activation level change during the course of the task. There is a potential benefit of creating a resistive training device that accommodates individuals displaying erratic operational behavior.
By changing force resistance in response to changes in performance outputs, the control system assists the simulated neuromuscular actuator in moving isokinetically through the full range of motion despite simulated fatigue and erratic operational behavior.

2.3. Exercise in Microgravity

Losses of muscle strength and bone mineral density along with aerobic deconditioning are major health concerns for astronauts during prolonged space flights (20) (189). The lack of gravity-related loading in a microgravity environment leads to a loss of bone mineral density at a rate of 1 to 2% per month and a reduction of muscle strength in the lower limbs at a rate of 2 to 8% per week (190). In bed-rest studies simulating weightlessness, knee extension strength losses of 12 to 15% have been seen over 14 to 16 days (189). In-space studies aboard Skylab 2 have shown knee extension strength losses of 20% over 28 days. Strength losses can compromise not only astronaut in-flight and post-flight health and safety, but mission success as well. Therefore it is necessary to prevent muscle strength losses during prolonged spaceflight. It is also necessary to exercise efficiently in order to reduce the demand on life support system consumables and save astronaut time (190).

In-flight exercise programs have included aerobic exercise and resistive training (191) (190). One form of aerobic exercise uses special treadmills to increase loading on the lower body, although not to an equivalent level of what would be felt on Earth (192). Aerobic exercise has not been successful at preventing loss of muscle strength (189), so it seems that upper and lower-body resistance training is a very important component of in-flight exercise (190). There is limited data available to determine the optimal resistance training program, thus bed-rest studies have been completed to simulate the effects of weightlessness (189). Studies by Caruso and Bamman have indicated that a device that can provide both high tension concentric and eccentric resistance in a microgravity environment would be ideal. This finding is in agreement with other
studies showing that both concentric and eccentric training are important for activities of daily living, for increasing strength, and for stimulation of muscle hypertrophy (193).

As previously noted, the PMA driven by pressurized air does not need gravity to produce resistance. This characteristic, along with the fact that pneumatic muscle itself is very lightweight, clean, and compact, make it an attractive option for a microgravity exercise device. Other equipment necessary for the operation of the device (air tank and/or compressor, fixtures, and electronics) can be designed to minimize size, weight, and resource consumption. Many resistive training devices only allow for concentric exercise (194), and in-flight exercise devices seem to be no exception (190). A PMA system configured to produce both concentric and eccentric resistance is reported here.

2.4. Task Details

The simulated isokinetic knee extension task which is used throughout this work consists of three phases. During the first phase, the concentric phase, the lower leg is extending at a constant angular velocity about the knee joint from 1.57 rad (90°) of knee flexion to approximately 0.087 rad (5°) of knee flexion (extended leg). During the second phase, the isometric phase, the lower leg is held at approximately 0.087 rad (5°) of flexion for closed loop studies and 0 rad for open loop studies for a duration of 5 seconds. During the third phase, the eccentric phase, the lower leg is flexing at a constant angular velocity about the knee joint from the isometric hold location to 1.57 rad (90°) of knee flexion. In reality, the limb must accelerate at the beginning of the exercise to get to the proper velocity and decelerate once the task is complete, but for the majority of the task the limb is moving at a constant angular velocity (183). The acceleration and deceleration are considered in this research.
3. THREE-ELEMENT PHENOMENOLOGICAL PNEUMATIC MUSCLE ACUATOR MODEL

The control of the PMA is based on a three-element phenomenological model consisting of a spring element, damping element and contractile element arranged in parallel. An illustration of the model, including relative directions of force, is shown as Figure 3. Governing equations for the model are shown as Eq. (11) and (12).

![Three-element phenomenological model](image)

Figure 3. Three-element phenomenological model

\[ m\ddot{x} + B_{PMA} \dot{x} + K_{PMA} x = F_{ce} - F_L \]  
(11)

\[ B_{PMA} \dot{x} + K_{PMA} x = F_{ce} - F_R \]  
(12)
In Eq. (11) and (12), $x$ is displacement of the pneumatic muscle actuator, $B_{PMA}$ is the damping coefficient, $K_{PMA}$ is the spring coefficient, $F_{ce}$ is the contractile force coefficient, and $F_L$ is the external load. The contractile force coefficient $F_{ce}$ is the contraction force of the pneumatic muscle actuator which acts throughout an entire contraction for a given pressure (135). This element is dependent on pressure. At each pressure, there is an external force which when applied against the pneumatic muscle actuator, results in no length contraction. This external force is the $F_{ce}$ value. In Eq. (11), $m$ is the mass of objects being moved by the PMA. In Eq (12), $F_R$ is introduced as the total resistive force acting against the PMA.

Characterization of the Festo fluidic muscle was completed by Serres as reported in (19). The pressure dependent elements of the phenomenological model specific to a Festo fluidic muscle (MAS20-250N) were found through experimentation. Two linear pressure-dependent equations for $K_{PMA}$ were found (one for low pressures and one for high). Three expressions for $B_{PMA}$ were found. $B_{PMA}$ during PMA contraction was found to be a constant as was $B_{PMA}$ during PMA relaxation at low pressures. $B_{PMA}$ during PMA relaxation at high pressures was found to be a linear function of pressure. $F_{ce}$ was also found to be a linear function of pressure. The exact expressions are listed in Table 1 along with other expressions describing the three-element phenomenological model parameters from other studies using different pneumatic muscles.
Table 1. Pneumatic Muscle Actuator Characterizations Using Three-Element Phenomenological Model

<table>
<thead>
<tr>
<th></th>
<th>$K_{PMA}$</th>
<th>$B_{PMA}$</th>
<th>$F_{ce}$</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Function [N/mm]</td>
<td>$P$ Range [kPa]</td>
<td>Function [Ns/mm]</td>
</tr>
<tr>
<td>Lab Constructed PMA (195)</td>
<td>5.71+0.0307$P$</td>
<td>200-621</td>
<td>1.01+0.00691$P$</td>
</tr>
<tr>
<td>Preliminary Festo PMA (195)</td>
<td>0.256$P$+101-0.0468$P$+47.8</td>
<td>150-253 253-550</td>
<td>0.009$P$+6.77</td>
</tr>
<tr>
<td>Final Festo PMA (135)</td>
<td>32.7-0.0321$P$ 17.0+0.0179$P$</td>
<td>150-314 314-550</td>
<td><strong>Contraction</strong>: 2.90</td>
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<tr>
<td></td>
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<td><strong>Relaxation</strong>: 1.57</td>
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<td></td>
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<td>0.311+0.00338$P$</td>
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<td>2.91$P$+44.6</td>
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</tbody>
</table>

It is important to use a model that sufficiently characterizes the nonlinear dynamics of the pneumatic muscle actuator. The three-element phenomenological model is simple and maintains an intuitive feel of what is going on in the system, yet is has been shown to decently predict force and length output from the pneumatic muscle actuator (18) (195) (135). This model is fairly straightforward to implement and the characterization of the PMA has already been performed. As Balasubramanian et al. pointed out (136), the model does not include dynamic force response parameters such as damping ratio of rise and decay response and natural frequency of rise and decay response, which are important for accurate force feedback response. Characterization work on the three-element phenomenological model has, however, addressed Balasubramanian’s concerns regarding model parameters as functions of pressure and external load (19). The model does include two force terms. $F_R$ will be treated as an input of the model, which will be explained further, while $F_{ce}$ is a function of pressure. $F_{ce}$ is the amount of force the contractile force element produces. This force contribution was found to increase linearly with increasing pressure. According to Reynolds (18), it was determined that $F_{ce}$ was approximately independent of the
applied load. Serres work (19) determined that the model parameters remained valid for a specified range of external load at different pressure levels.
4. EXPERIMENTAL SETUP (DYNAMIC TEST STATION)

The system which will be used to demonstrate control of the PMA system will be referred to as the PMA Dynamic Test Station or DTS. In order to achieve the preliminary goals laid out, the DTS located in the Bioengineering Lab at Wright State University was enhanced with additional capabilities. These enhanced capabilities included utilizing configurable motor analog outputs, operating the motor in different control modes, and using digital motor command signals for added functionality. The system setup is shown in Figure 4.

![Diagram of Dynamic Test Station](image)

**Figure 4. Diagram of Dynamic Test Station**

The arrows in the figure represent signal paths between elements of the DTS. As seen in the figure, the computer and data acquisition card communicate back and forth with each other using LabVIEW as the interface. The computer also communicates directly with the motor driver via motor configuration software. The motor driver sends command signals to the motor (encoded
torque, encoded velocity or digital position command signals depending on the motor mode) and outputs data (analog and digital outputs) to the data acquisition card. The data acquisition card sends analog command signals to the proportional pressure regulator which controls the pressure inside the pneumatic muscle. The data acquisition card sends both analog and digital signals to the motor driver. Signals from the sensors, including pressure transducer, load cell, LVDT, and rotational potentiometer, are sent to the data acquisition card. The DTS is comprised of the equipment described below.

4.1. Parts of the Dynamic Test Station

A Dell Dimension 9200 computer (Windows XP operating system) is equipped with LabVIEW 8.0 (copyright 2005 National Instruments) and Pacific Scientific 800 Tools Release 1.02 (copyright 2002 Pacific Scientific). LabVIEW is the interface for controlling all signals leaving and entering the computer. It also runs control calculations, sends commands to the motor and proportional pressure regulator (which controls pressure inside the PMA), and records command outputs and sensor signal inputs. Pacific Scientific 800 Tools is the interface for configuring the motor driver, which controls the motor. The data acquisition card, PCI-6025E, made by National Instruments (Austin, TX) allows signals to be sent between the computer and the DTS electronics.

The source of simulated human movement in this research is a motor, which provides the torque for a knee extension exercise, analogous to the quadriceps muscles providing torque about the knee in a human. The motor, PMA45N-00100-00, was manufactured by Pacific Scientific (now part of Danaher Motion). The motor driver, model PC 833-001-T, also manufactured by Pacific Scientific, which is used to control motor behavior and is configured to measure torque and velocity output of the motor.

Many variations of pneumatic muscles have been created in the past; various construction methods, different diameters, different lengths, different materials, and different fiber patterns. A
commercially available PMA was chosen for this research, offering repeatability and device consistency. The PMA (MAS-20-250N-AAMCK manufactured by Festo Corporation) has a resting inner diameter of 20 mm (0.79 in) and length of 250 mm (9.84 in). It is rated for a maximum pressure of 600 kPa (87.2 psi). The maximum force output is 1200 N, while the maximum contraction displacement is 25% of nominal (resting) length (62.5 mm or 2.46 in). An aluminum slide, which rides on low-friction lubricated Teflon bearings, guides horizontal movement of the PMA. According to Gerschutz, the load required to overcome static and dynamic friction in the LVDT is 1.96 N (0.44 lb) and 1.47 N (0.33 lb), respectively (195). These loads are significantly less than the loads applied to the system, and as a result the aluminum slide friction is considered negligible. An inextensible cable fixed to the mobile end of the PMA is the mechanism by which force is transferred from the PMA to the motor. The cable is able to wind or unwind around a spool, rigidly fixed to the motor shaft, as motor shaft rotation and PMA contraction/relaxation occurs.

A linear variable differential transducer (LVDT), model JEC-060-G317-03 made by Honeywell-Sensotec, measures length change of the PMA. The rod within the LVDT is attached to the slide that guides PMA movement. A rotational potentiometer, model 6637S-1-502 made by Bourns, Inc., measures the position of the motor shaft. The range is 0-340° corresponding to 0 to 10V. The Festo fluidic muscle, slide, LVDT, and motor can be seen in Figure 5.
Figure 5. Photograph of Festo fluidic muscle, slide and LVDT

The motor shaft spool, an aluminum part rigidly attached to the motor shaft about which the force transfer cable winds/unwinds, is shown in Figure 6.

Figure 6. Photograph of motor and cable attachment

The blocking plate is another aluminum part attached to the motor shaft spool which prevents the cable from jumping off the spool. This, along with the rotational potentiometer, can be seen in Figure 7.
A proportional pressure regulator (PPR), model MPPE-3-1/8-6-010-B made by Festo Corporation (Hauppauge, NY) and shown in Figure 8, is used to control the pressurization of the pneumatic muscle. It has a response time of 0.22 seconds and a maximum flow rate of 800 L/min. The PPR is controlled by analog voltage inputs originating from LabVIEW. It has a closed loop mechanism that shuts off the air flow when it reaches the specified set point. A pressure transducer, model SDE1-D10- G2-W18-L-PU-M8 made by Festo Corporation (Hauppauge, NY), monitors the pressure in the inner bladder of the pneumatic muscle actuator. It has a range of 0-10V corresponding to 0-10 bars. Nitrogen gas, chosen for its non-flammable properties and cleanliness, flows from a regulated gas tank.
Additional electronics within the DTS include several wire block terminals (for motor driver, PPR, and sensor connections), power supplies, amplifiers, an emergency power-off switch for the motor, and an enable/disable switch for the motor. Wiring connections between the motor driver and the data acquisition card, between the sensors and data acquisition card, and between the PPR and data acquisition card are shown in Appendix B.

DTS component calibration techniques described in (135) were used to maintain consistency between the calibrations performed before the characterization of the PMA and the calibrations performed before use of the characterized PMA model in this research. A new rotational potentiometer calibration technique, described in Appendix D, was used prior to closed loop testing due to the importance of the relative starting signals between the rotational potentiometer and motor. The calibration curves for DTS equipment are listed below in Table 2 and 3. \( V \) represents voltage in \([V]\), \( x \) is PMA displacement in \([\text{mm}]\), \( \theta \) is angle of knee flexion in \([\text{deg}]\) for set 1 and \([\text{rad}]\) for set 2, and \( P \) is pressure in \([\text{bar}]\). The rotational potentiometer equation varies based on starting position and thus must be recalibrated each time it is
reconnected to the motor shaft. More DTS component equations specific to closed loop testing can be found in section 7.5.

Table 2. Set 1 of Calibration Equations for DTS Devices

<table>
<thead>
<tr>
<th>Device</th>
<th>Calibration Equation</th>
</tr>
</thead>
<tbody>
<tr>
<td>LVDT (relative to neutral PMA length)</td>
<td>( x = 10.15^*V - 12.76 )</td>
</tr>
<tr>
<td>Rotational Potentiometer</td>
<td>( \theta = -69.77^*V + 178.23 )</td>
</tr>
<tr>
<td>Proportional Pressure Regulator</td>
<td>( V = 1.668^*P + 0.1589 )</td>
</tr>
</tbody>
</table>

Table 3. Set 2 of Calibration Equations for DTS Devices

<table>
<thead>
<tr>
<th>Device</th>
<th>Calibration Equation</th>
</tr>
</thead>
<tbody>
<tr>
<td>LVDT (relative to neutral PMA length)</td>
<td>( x = 10.206^*V - 13.058 )</td>
</tr>
<tr>
<td>Rotational Potentiometer</td>
<td>( \theta = -1.243^*V + 2.467 )</td>
</tr>
</tbody>
</table>

4.2. Interaction Between Motor and Pneumatic Muscle Actuator

For proper force transfer to occur, PMA displacement must be controlled. A cable is the force transfer mechanism. It is necessary for the PMA to contract or relax according to how the motor shaft is moving to ensure that there is always a proper length of cable between the PMA and motor. A large error in PMA displacement would cause either cable slack or PMA stretching, resulting in disruptions in proper force transfer.

At the starting position of 1.57 rad (90°) of flexion, there is enough slack in the cable so that it can wrap around the motor spool through the full simulated task range of motion (1.57 rad). Before initializing any motor movement, the slack must be taken up by pressurizing the PMA, causing it to contract (see Figure 9a). Once the slack is taken up, the control system must adjust the PMA pressure to achieve the necessary PMA resistance force and PMA length required for the task (see Figure 9b). The PMA will mainly be operating in relaxation mode during the simulated concentric phase as the PMA must get closer and closer to its nominal length. During
the eccentric phase, the PMA will mainly be operating in contraction mode so that the PMA length shortens and slack does not form in the cable.

This configuration may make control more difficult due to the need to control both force and displacement, but it has the potential for improved response time and safety. Only small incremental PMA displacements throughout the task are necessary, produced by small incremental changes in internal PMA pressure. Small changes in pressure occur more rapidly than large step-type changes in pressure required in other PMA configurations while also reducing dry to kinetic friction. This configuration takes advantage of not only the contraction cycle of the PMA (increasing internal PMA pressure to shorten), but also the relaxation cycle (decreasing PMA pressure to lengthen). This configuration also has the potential to enhance safety for the human operator because PMA produced force is only transferred if there is something opposing the PMA contraction (statically or dynamically) and if there is proper PMA displacement.

Figure 9. (a) Illustration of PMA, cable and spool before task begins (b) Illustration of PMA, cable and spool at different task angles
Theoretically, the human is in control of stopping the force transfer by dropping their lower leg to the starting position after the task has begun. As an added safety feature within the current control system, the PMA displacement is fed back to the system and compared to simulated joint angle so that a stop switch is triggered when a certain amount of slack occurs.

4.3. Motor Control

The Pacific Scientific motor and motor driver within the DTS is able to work in multiple control modes and uses both analog and digital commands. It is able to switch control modes during operation, and it is able to output information regarding its operation and performance. The motor control modes used in this research are position mode-predefined moves and velocity mode-analog command (with switches between the two). Information detailing motor control can be found in Appendix A.
5. TASK-SPECIFIC RESISTANCE PRODUCED BY PNEUMATIC MUSCLE ACTUATOR FOR EXERCISE IN MICROGRAVITY

5.1. Objectives

The objectives of this work are to control a PMA as a task-specific source of time-varying resistance and to demonstrate such PMA control using a motor to simulate human movement during an isokinetic strength training task. The internal PMA pressure is controlled to generate a predefined resistance as well as to control the PMA displacement ensuring proper force transfer between the PMA and the motor working against the PMA. The PMA produced force opposing the motor-simulated human movement can change in any desired fashion within the capability limits of the PMA (force level 0 N up to 1200 N depending on the corresponding PMA displacement necessary for proper force transfer).

The ideal amount of resistance that the PMA should produce depends on the specific exercise task being simulated. To simulate a knee extension exercise where the human is working against a simple ankle weight, the PMA produced resistance must equal the force required to move the ankle weight, making the ankle weight no longer necessary. To simulate a knee extension exercise in a microgravity environment, the PMA produced resistance should equal not only the force required to move the ankle weight, but also the lower leg and foot. While in a microgravity environment, the weight of the lower leg and foot is not felt by the operator in the same way as on Earth due to lack of gravity related loading. Therefore, it is beneficial to incorporate this load into the desired task resistance. In either scenario (normal environment or microgravity environment), the level of PMA resistance will change with leg position. In this study, the force resistance provided by the PMA is a combination of a baseline force (equivalent
to what is required to lift the lower leg and foot of a scaled human operator) and the force resistance felt while lifting an ankle weight. The PMA also provides an accommodating force resistance that changes as a function of joint angle and corresponding muscle length so that the muscles can work closer to optimal levels.

5.2. Methods

5.2.1. Experimental Methods

5.2.1.1. Inverse Model Open Loop Control

The control of the PMA is based on a three-element phenomenological model consisting of a spring element, damping element and contractile element arranged in parallel. An illustration of the model and governing equations can be found in section 3. The governing equations of motion are also shown below for reader convenience as Eq. (13) and (14).

\[ m\ddot{x} + B_{PMA} \dot{x} + K_{PMA} x = F_{ce} - F_L \] (13)

\[ B_{PMA} \ddot{x} + K_{PMA} x = F_{ce} - F_R \] (14)

In Eq. (13) and (14), \( x \) is displacement of the pneumatic muscle actuator, \( B_{PMA} \) is the damping coefficient, \( K_{PMA} \) is the spring coefficient, \( F_{ce} \) is the contractile force coefficient, and \( F_L \) is the external load. The contractile force coefficient, \( F_{ce} \), is the pressure-dependent contraction force of the PMA. In Eq. (13), \( m \) is the mass of objects being moved by the PMA. In Eq. (14), \( F_R \) is introduced as the total resistive force acting against the PMA.

As previously explained, the forces attributed to each element of the model (spring \( (K_{PMA}) \), damping \( (B_{PMA}) \), and contractile \( (F_{ce}) \)) have been found to be pressure dependent. In addition, the functions derived to define the relationship between force and pressure have been found to vary with pressure range. Table 4 contains characterization details for the Festo PMA used within the DTS.
Table 4. PMA three-element phenomenological model characterization for Festo MAS-20 fluidic muscle

<table>
<thead>
<tr>
<th>Model Element</th>
<th>Equation Representation</th>
<th>Characterization Equations</th>
</tr>
</thead>
<tbody>
<tr>
<td>$K_{PMA}$ [N/mm]</td>
<td>$a_K P + b_K$</td>
<td>$K_{PMA} = 0.0321P + 32.7$ for $150 &lt; P \leq 314$ kPa&lt;br&gt;$K_{PMA} = 0.0179P + 17.0$ for $314 &lt; P &lt; 550$ kPa</td>
</tr>
<tr>
<td>$B_{PMA}$ [Ns/mm]</td>
<td>$a_B P + b_B$</td>
<td>$B_{PMA \text{ contraction}} = 2.90$ for $150 &lt; P &lt; 550$ kPa&lt;br&gt;$B_{PMA \text{ relaxation}} = 1.57$ for $150 &lt; P \leq 372$ kPa&lt;br&gt;$B_{PMA \text{ relaxation}} = 0.00338P + 0.311$ for $372 &lt; P &lt; 550$ kPa</td>
</tr>
<tr>
<td>$F_{ce}$ [N]</td>
<td>$a_F P + b_F$</td>
<td>$F_{ce} = 2.91P + 44.6$ for $150 &lt; P &lt; 550$ kPa</td>
</tr>
</tbody>
</table>

By expanding these pressure dependent terms and rearranging the governing equation (Eq. (14)), an inverse of the phenomenological model was derived. The inverse model, shown as Eq. (15), is the basis of the open loop controller used to calculate the PMA pressure corresponding to a desired PMA resistance and displacement. The diagram shown in Figure 10 illustrates the flow of information and decisions that need to be made for the task demonstration and PMA control to occur. Pressure is the output of the inverse model, whereas desired force resistance (according to the strength training exercise being simulated), PMA displacement and PMA velocity are the inputs.

$$P = \frac{b_F - b_B \ddot{x} - b_K x - F_R}{a_K \ddot{x} + a_F x - a_x}$$

(15)
5.2.1.2. Determining Desired Profiles

There is a relationship between speed of the exercise and training outcome. Slow speeds (below 90°/s) produce strength and power gains related specifically to angular velocity of movement (183). Faster speeds (between 180°s/s and 240°/s) produce strength gains over a wide range of velocities (183) (196). If the goal is to increase strength at one particular velocity, the isokinetic exercise should be performed at that velocity. A slow speed, 18°/s, was chosen for experimental evaluation because the focus here is to maintain and/or increase strength for slower movements such as walking. The slow speed also allows us to monitor the system more closely.

Desired task speed (18°/s in this demonstration), gives us a desired position profile of the motor shaft over time. In our configuration, the PMA displacement must remain equal to motor displacement in order to maintain proper force transfer between the PMA and motor (source of simulated human quadriceps torque production). PMA displacement is therefore calculated based on desired motor shaft position using the corresponding motor spool arc length. Knowing PMA displacement over time allows us to calculate desired PMA velocity.

Figure 10. Flow diagram of task-specific PMA resistance control
The desired PMA produced resistance profile is based on one of two simulation scenarios. The first scenario is a knee extension exercise in microgravity with the PMA acting to replace resistance provided by ankle weights in addition to resistance attributable to the lower leg and foot felt in a 1g environment. The second scenario is also a knee extension in microgravity, but an accommodating resistance is defined instead of a calculated ankle weight equivalent. The accommodating resistance changed as a function of joint angle and corresponding muscle length. Based on the capabilities of the human quadriceps simulation method (motor limited to 22 Nm of available torque), the resistance needed to replace the lower leg and foot felt in 1g was calculated using a scaled simulated human operator (1:0.4 body mass ratio with anthropometrically scaled limbs based on the scaled mass).

An average sized human operator was chosen as the model for this research. A healthy 5'8" (1.73 meter) adult has a body mass index between 19 and 25 kg/m$^2$, which equates to between 125 and 164 lbs. A weight in the middle of this range was chosen; 144 lbs or 65 kg. The mass was then scaled down (1:0.4 body mass ratio) based on how much torque was necessary to move the scaled lower leg with ankle weights about the knee. The maximum torque had to be within the limitations of the motor so that the motor was capable of rotating the scaled lower leg, foot, and ankle weight through the 90° range of motion. The scaling factor of the leg allowed for a range of resistances to accommodate different simulated human operator capabilities. The scaling of the leg was done in such a way that the mass was reduced and then the height was selected based on the mass to maintain normal proportions of an adult human being. The correctly proportioned lower leg mass, lower leg length, foot mass, and foot length were calculated using anthropometric data from (188) and are shown in Table 5.
Table 5. Mass and Lengths for Model and Scaled Human Body Segments

<table>
<thead>
<tr>
<th>Body Segment</th>
<th>Scaling Equation</th>
<th>Model Human</th>
<th>Scaled Human</th>
</tr>
</thead>
<tbody>
<tr>
<td>Height [m]</td>
<td>0.617*Height</td>
<td>1.73</td>
<td>1.07</td>
</tr>
<tr>
<td>Mass [kg]</td>
<td>0.4*Mass</td>
<td>65</td>
<td>26</td>
</tr>
<tr>
<td>Lower Leg Length [m]</td>
<td>0.245*Height</td>
<td>0.42</td>
<td>0.26</td>
</tr>
<tr>
<td>Lower Leg Mass [kg]</td>
<td>0.046*Mass</td>
<td>2.99</td>
<td>1.20</td>
</tr>
<tr>
<td>Foot Length [m]</td>
<td>0.15*Height</td>
<td>0.26</td>
<td>0.16</td>
</tr>
<tr>
<td>Foot Mass [kg]</td>
<td>0.014*Mass</td>
<td>0.91</td>
<td>0.36</td>
</tr>
</tbody>
</table>

In order to calculate torque about the knee, proximal and distal leg coordinates were calculated along with segment centers of mass. The ankle weight is assumed to be acting at the distal end of the lower leg (the ankle weight’s center of mass is located at the ankle). See Figure 11 for a plot of a subset of the angle-dependent proximal and distal locations as well as the center of mass for each of the segments, calculated using anthropometric data from (188). The knee is located at (0,0). The estimated lower leg center of mass is located 0.435 times the lower leg length away from the proximal end of the segment (knee). The estimated foot center of mass is located half way along its length.
Motor torque necessary to move the lower leg, foot, and any additional weight is calculated about the center of rotation (simulated knee) using the estimated centers of mass. Important simplifications are made in order to do this. The leg and foot are assumed to be symmetrical about the sagittal plane allowing us to work in two dimensions instead of three. The parts of the leg are treated as rigid bodies (non-deformable, fixed center of mass, and material homogeneity). The foot stays at a right angle relative to the lower leg through the task. The quadriceps muscles have a line of action at a distance and angle away from the central axis of the lower leg depending on the insertion point of the quadriceps muscle (patellar ligament). The location of the insertion point varies from person to person and varies with movement of the knee. As the knee joint moves, the line of pull of the quadriceps muscles will change as will the axis of rotation because the knee joint rotates, slides, and rolls to maintain stability at different leg positions. Here, rotation (provided by the motor) is occurring about a fixed axis. This will produce general, but still useful approximations of a human leg. The focus here is on knee
extension, so effects of the hamstrings will not be considered although there is some co-
contraction.

The moments of the system include the torque provided by the motor, \( T_{motor} \), the torque
required to move the lower leg, \( T_{lowerleg} \), and foot, \( T_{foot} \), and finally the replacement resistance
torque, \( T_{rr} \). Replacement resistance torque for scenario 1 is equivalent to the torque required to
move an ankle weight through the range of motion. Replacement resistance torque for scenario 2
is the accommodating resistance that changes as a function of position, peaking near the optimal
muscle length. The goal is to move the leg at a constant velocity, so theoretically there is no
angular acceleration in the system. The system torques can therefore be summed and set to zero
yielding an expression for motor torque \( T_{motor} \), which can now be thought of as the total desired
torque to be produced by the PMA, \( T_{PMA} \) (see Eq. (16)).

\[
T_{motor} = T_{lowerleg} + T_{foot} + T_{rr} = T_{PMA} \quad \text{(16)}
\]

Recommendations regarding the proper resistance force (replacement resistance
component of this system) for a knee extension exercise vary depending on the goal of the
exercise and, of course, the capabilities of the person performing the exercise. Table 6 lists three
levels of ankle weights (4, 8, and 16 lb) and their corresponding percentage of body weight for
our scaled human operator (scaled body mass/weight is 26 kg/255 N). Please note this force is not
reflective of the quadriceps or PMA force required to lift the weight. These ankle weights cover a
range of 7 to 28% of body weight. The resistances can also be thought of in terms of percentage
of peak torque output. The average peak torque for a 30 to 40 year old female at different speeds
was found from published literature (197). A linear fit was applied to reported data so that
information about different speeds could be interpolated from the available data. At \( 18^\circ/s \) (desired
task speed), the peak torque approximation was found to be 162 Nm. That value was scaled down
to 40Nm using scaling parameters for both the force (2.5) and distance (1.62) components of the
torque. Table 6 lists three levels of ankle weights and their corresponding percentage of peak torque (ranging from 12 to 46%). The range of resistances allows for the evaluation of a range of simulated human movement within the system.

Table 6. Relationship between Resistance Replacement and Scaled Human Quadriceps

<table>
<thead>
<tr>
<th>Resistance Level</th>
<th>Ankle Weight to Body Weight Ratio</th>
<th>Profile Peak Torque [Nm]</th>
<th>Percentage of Peak Scaled Human Torque</th>
</tr>
</thead>
<tbody>
<tr>
<td>4 lb (1.81 kg) ankle weight</td>
<td>7%</td>
<td>4.65</td>
<td>11.6%</td>
</tr>
<tr>
<td>8 lb (3.63 kg) ankle weight</td>
<td>14%</td>
<td>9.30</td>
<td>23.3%</td>
</tr>
<tr>
<td>16 lb (7.26 kg) ankle weight</td>
<td>28%</td>
<td>18.60</td>
<td>46.5%</td>
</tr>
</tbody>
</table>

With the desired replacement resistance torques set, the total desired resistance, \( T_{motor} \) or \( T_{PMA} \) of Eq. (16), is calculated. The total desired resistance was evaluated in terms of force, found by dividing \( T_{PMA} \) by the distance from the center of rotation to the force’s line of action (27 mm). For each scenario, the resultant \( F_{PMA} \) was plotted relative to motor shaft position (simulated joint angle). Second order polynomials were fit to these values (see Figure 12). Final desired resistance equations and coefficients of determination (\( R^2 \) values calculated to verify goodness of fit) are shown in Table 7. The first three force profiles (profiles A, B, and C shown in Table 7) represent an equivalent resistance to a 4 lb (1.81 kg), 8 lb (3.63 kg), and 16 lb (7.26 kg) ankle weight in a microgravity environment. The fourth force profile (profile D shown in Table 7) represents an accommodating force resistance changing according to muscle length, with a 338 N peak near optimal muscle length and gradual force reduction away from optimal muscle length. These equations are used as the command resistance profile within the open loop control system.
Figure 12. Resistance force (derived from torque requirements) as a function of motor position

Table 7. Total force resistance equations as a function of motor position (simulated joint angle in degrees of flexion)

<table>
<thead>
<tr>
<th>Force Resistance Profile</th>
<th>$F_{PMA}$ Equation [N]</th>
<th>$R^2$</th>
</tr>
</thead>
<tbody>
<tr>
<td>A 4 lb ankle weight equivalent</td>
<td>$F = -0.0275 \theta^2 - 0.373 \theta + 260.75$</td>
<td>0.9992</td>
</tr>
<tr>
<td>B 8 lb ankle weight equivalent</td>
<td>$F = -0.0452 \theta^2 - 0.772 \theta + 436.23$</td>
<td>0.9992</td>
</tr>
<tr>
<td>C 16 lb ankle weight equivalent</td>
<td>$F = -0.0806 \theta^2 - 1.57 \theta + 787.19$</td>
<td>0.9992</td>
</tr>
<tr>
<td>D Accommodating Profile</td>
<td>$F = -0.1006 \theta^2 + 11.664 \theta$</td>
<td>n/a</td>
</tr>
<tr>
<td>E Scaled lower leg and foot only</td>
<td>$F = -0.0097 \theta^2 + 0.026 \theta + 85.274$</td>
<td>0.9992</td>
</tr>
</tbody>
</table>

5.2.2. Experimental Procedure

The different knee extension scenarios were demonstrated using the experimental method previously described and the DTS signal paths shown in Figure 4. Pressure commands were calculated using the controller previously discussed to produce the desired PMA resistance and displacement. Analog voltages corresponding to pressure commands were sent to the PPR, while analog and digital commands were sent to the motor to simulate an isokinetic concentric and eccentric knee extension task. Position, velocity and motor torque outputs were recorded and analyzed.

Statistical analysis was performed to evaluate the PMA force transfer (based on motor torque output), PMA displacement, and motor position. RMSE values were calculated to compare
the desired values to the actual outputs of the system for force profiles A, B, C, and D shown in Figure 12 (also listed in Table 7).

5.3. Results

Table 8 lists the RMSE values for each parameter (motor shaft position, PMA displacement, and resistance) broken down by phase (concentric, isometric, eccentric, and entire task) and resistance profile. Evaluation of the concentric and eccentric phases, the phases of most interest in this study, reveals that motor position RMSE values range from 0.2 to 4.4 mm (0.5 to 9.3°). PMA displacement RMSE values range from 0.6 to 4.5 mm. Force resistance RMSE values range from 13.5 to 99.2 N (equivalent torque RMSE ranges from 0.4 to 2.7 Nm).
### Table 8. RMSE values between desired outputs and actual outputs for different PMA resistance profiles and different task phases

<table>
<thead>
<tr>
<th>Resistance Profile</th>
<th>Phase</th>
<th>Position RMSE</th>
<th>PMA Disp. RMSE</th>
<th>Resistance RMSE</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>[mm]</td>
<td>[°]</td>
<td>[mm]</td>
</tr>
<tr>
<td>A</td>
<td>Concentric</td>
<td>1.10</td>
<td>2.33</td>
<td>1.59</td>
</tr>
<tr>
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<td>Entire task</td>
<td>2.98</td>
<td>6.33</td>
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For each resistance profile, desired and actual motor position, desired and actual PMA displacement, and actual motor position angle are plotted as a function of time. Figures 13, 15, 17, and 19 show position/displacement tracking results for the 4 lb equivalent resistance profile, 8 lb equivalent resistance profile, 16 lb equivalent resistance profile, and accommodating profile, respectively. Desired and actual force outputs (based on motor torque outputs) are also plotted as a function of time for each resistance profile. Figures 14, 16, 18, and 20 show force tracking results for the 4 lb equivalent resistance profile, 8 lb equivalent resistance profile, 16 lb equivalent resistance profile, and accommodating profile, respectively.

Between 5 and 10 seconds is the concentric phase of the task where the motor is simulating extension of the lower leg. Between 10 and 15 seconds is the hold phase, simulating an isometric exercise. Between 15 and 20 seconds is the eccentric phase where the motor is...
simulating lowering of the lower leg in a controlled fashion while the PMA is trying to pull the leg downward.

Figure 13. Position and PMA displacement results for resistance profile equivalent to 4 lb ankle weight

Figure 14. Force results for resistance profile equivalent to 4 lb ankle weight
Figure 15. Position and PMA displacement results for resistance profile equivalent to 8 lb ankle weight

Figure 16. Force results for resistance profile equivalent to 8 lb ankle weight
Figure 17. Position and PMA displacement results for resistance profile equivalent to 16 lb ankle weight

Figure 18. Force results for resistance profile equivalent to 16 lb ankle weight
Figure 19. Position and PMA displacement results for accommodating resistance profile

Figure 20. Force results for accommodating resistance profile
5.4. Discussion

Results for each of the ankle weight equivalent resistance profiles indicate that the open loop control method provides decent tracking for the application with motor-simulated human movement. PMA displacement RMSE values within 1.6 mm during the concentric and eccentric phases indicate that proper force transfer can occur between the PMA and source of opposing actuation (motor in this case). Position tracking results (motor position RMSE values within 1.3 mm or 2.8°) as well as constant slopes from 5 to 10 seconds and 15 to 20 seconds as shown in Figure 13, 15, and 17, indicate near constant velocity motor shaft rotation in the regions of interest, analogous to concentric and eccentric isokinetic movement. Force tracking results are relatively accurate, ranging from 13.5 to 59.5 N (0.36 to 1.61 Nm) during the simulated concentric and eccentric phases. However, there is variation in tracking accuracy between scenarios. Larger force errors are not necessarily a result of increased position/PMA displacement errors or a higher magnitude resistance profile. A potential source of variation may be differences in relative starting positions between the PMA and motor as this was observed to impact force outputs during preliminary studies. Another potential source of variation may be hysteresis-induced effects. As reported in Festo technical literature (24), hysteresis is specified to be ≤ 2.5% of nominal length or 6.25 mm. This amount of hysteresis has the potential to affect repeatability of different system parameter outputs.

Results for the accommodating resistance profile show increased tracking error as compared to the ankle weight equivalent resistance profiles. This is most likely due to the more complicated nature of the accommodating profile. Between 5 and 10 seconds (concentric phase) and 15 to 20 seconds (eccentric phase), the desired resistance rises from 0 N to 338 N and then falls back down to 0 N. The ankle weight replacement profiles rise to 273, 442, or 780 N (depending on the profile), hold, and fall over 15 seconds. Delays in system response originating from the time constant of the PMA itself, as well as a 0.22 s reaction time of the PPR which regulates internal PMA pressure, may be making it difficult for the PMA to react as quickly as
necessary to track the more complicated force profile (and thus more complicated pressure profile). A major contributing factor to the increased motor position tracking error is an initial shift in motor position corresponding to the onset of PMA produced resistance as shown in Figure 19. The system cannot recover from an initial offset like this with open loop control. With the incorporation of closed loop control, there will be an attempt to improve errors associated with delayed system response, variation between resistance profiles, and offset. Closed loop control has the potential to reduce tracking errors of the PMA produced resistance, PMA displacement and motor position (simulated joint angle) in either type of resistance scenario discussed here.

This research utilized a motor with autonomous control as the human quadriceps simulator. There are several limitations of using the motor in this way. First, the motor lacks the adaptability that occurs with neuromuscular activation. The motor controls speed as commanded and has limited reactions to increases or decreases of resistance acting against it. Second, the motor operates in a more repetitive manner. Typical human behaviors, such as physiological fatigue or erratic control, were not present. Future work addresses these issues with a more sophisticated human quadriceps simulator, which reacts to the force acting against it and simulates both fatigued and erratic control. This human quadriceps simulator will be used to test a closed loop controller based on the inverse model tested here. A third limitation of the motor as human quadriceps simulator is the limited torque output. The motor torque output of 22 Nm is about 10-20% of the maximum human quadriceps torque values (17). However, the work done here is at a scaled level so that the limited motor torque does not restrict progress on developing effective PMA control. The maximal force output of the PMA itself (1200 N or 32 Nm of torque with the current moment arm) is also a limitation. When incorporating these PMAs into an exercise device to be used by humans, additional PMAs can be used simultaneously to increase the force output. The moment arm can also be increased somewhat, depending on what is practical for the mechanical design.
This work presents a successful demonstration of the task-specific resistance that the PMA can provide. This task-specific force output, in addition to the PMA benefits previously discussed, make PMAs an attractive actuator option for strength training and rehabilitation devices, here on Earth and in microgravity environments. Despite the challenges and limitations, the results indicate that task-specific force output of the PMA is possible and that the inverse model is a good foundation upon which to build closed loop control.
6. SIMULATING NEUROMUSCULAR ACTUATION

6.1. Human Quadriceps Dynamic Simulator (HQDS)

A necessary component of this work has been the development of a human quadriceps dynamic simulator (HQDS), which together with a motor, acts as the simulated neuromuscular actuator that was used to test control operation of the experimental PMA system prior to human implementation. The HQDS was developed to produce realistic torque and velocity reactions to external stimulation, specifically the force resistance produced by the PMA. The task that the HQDS will be simulating is isokinetic knee extension strength training.

Simulated neuromuscular actuation of the system is produced using a motor, which acts as the actuation mechanism, and the human quadriceps dynamic simulator (HQDS), which determines motor commands based on system parameters. Simulated neuromuscular actuation is in terms of torque production rather than force production to avoid the confusion in comparing motor force production to human quadriceps force production. The goal is not to exactly replicate quadriceps force production; rather we are using a motor to simulate torque production about the knee joint, which is produced primarily through contraction of the quadriceps. Working with torque also eliminates the need to estimate the moment arm of the human quadriceps muscles.

The motor itself is configured to perform closed loop velocity control in an attempt to simulate neuromuscular behavior. The simulated neuromuscular actuator receives motor position, velocity, and torque feedback from system sensors. The HQDS determines how the motor should react to the system feedback, taking fatigued or erratic operational behavior into account (depending on the simulation settings). Velocity and torque output variation are introduced by the HQDS so that PMA control system effectiveness and stability can be tested. The motor carries
out the HQDS command by increasing or decreasing motor torque according to the simulated ability of the neuromuscular actuator.

The motor’s position will be controlled at the simulated extremes of range of motion. This ensures that the starting angle is correct and creates a simulated hard stop at the 0.087 rad (5°) of flexion position, preventing further extension past this point.

The heavily studied torque-velocity relationship is used to estimate the relationship between torque produced about the knee joint and angular velocity of the joint. The torque-velocity relationship provides a simple method in which to estimate the simulated limb velocity relative to the amount of torque being produced by the motor as it’s working against the PMA produced resistance. If the level of PMA resistance changes, resulting in a change of motor torque, the HQDS will alter motor commands in an attempt to simulate human neuromuscular reaction to resistance changes.

It is important to note that the torque-velocity relationships for concentric and eccentric contractions are very different. It is also important to note that there is a difference between the force-velocity-length relationship of the isolated quadriceps and the torque-velocity-position relationship of the knee joint during knee extension. The quadriceps muscles are the dominant muscle group and produce the majority of the torque for a knee extension, but only in cooperation with the other structures of the leg. The focus here is on the knee joint in general because this is what can be directly measured (for both simulated human torque production and actual human torque production) and thus more easily simulated. Setting up control using torque produced about the knee will be more useful for future implementation as well.

A.V. Hill’s well established force-velocity model (198) has a concentric region of decreasing force with increasing muscle shortening velocity. The eccentric region is different in that force increases with increasing muscle lengthening velocity. Hill’s work is based on single muscle specimens, and researchers have since tried to replicate the model’s predicted torque-velocity results in the case of voluntary muscle contractions. Many have used electrical
stimulation and electromyography to study the relationship between torque, velocity and activation level.

The concentric phase of maximal voluntary knee extension has been shown to follow the torque-velocity relationship from Hill’s model (199) (200). The eccentric phase, on the other hand, deviates from the model in that maximal torque remains around the isometric torque output and after a certain lengthening velocity, the torque output levels off instead of continuing to increase with increasing velocity (201) (202) (203) (186) (204).

The eccentric peak torque levels of single muscle specimens, which rise above isometric and eccentric voluntary levels, have been produced using artificial activation of the knee extensors (203). This has led researchers to hypothesize that there is an underlying neural control element limiting activation during voluntary high tension eccentric knee extension (201) (203) (186), perhaps to protect the knee (204).

Dudley et al. studied voluntary and artificially activated eccentric knee extension across a wide range of velocities and found that with artificial activation, torque increased to 1.4 times isometric between 0 and 1.57 rad/s (203). The torque level plateaued beyond 1.57 rad/s.

Rutherford et al. studied voluntary eccentric knee extension and found that at low speeds (0.524 rad/s) forces were approximately 20 to 30% lower than isometric and rose to isometric at higher speeds (199). Low-velocity eccentric findings from the Dudley and Rutherford studies in conjunction with a Hill-based force-velocity model developed by Phillips (188) were used to create a baseline torque-velocity curve representing maximum activation. Five parameters are used to scale the HQDS torque-velocity curve, allowing customization and flexibility in modeling the neuromuscular element of knee extension. These parameters include peak isometric torque, $T_{o,\text{max}}$, peak angular velocity, $V_{o,\text{max}}$, eccentric simulated torque scaling parameter, $a_{s,Te}$, concentric simulated torque scaling parameter, $a_{s,Tc}$, and simulated velocity scaling parameter, $a_{s,V}$. 

89
For this work, human quadriceps torque output is simulated using motor torque output and recorded as a function of motor current. When transferred to human use, the external torque production about the knee, a measure of voluntary contraction, can be easily measured. Finding the force produced by the quadriceps muscles is much more complicated, as there are many individual muscle and ligament forces acting around the knee joint during isokinetic exercise. Analyzing electromyography data measuring activation level of the quadriceps is one possible alternative, but would not be ideal, particularly for a dynamic task. Therefore, a voluntary contraction type torque-velocity curve will be modeled instead of a curve based on activation level.

Equations for simulated human quadriceps concentric torque output, \( T_{s,c} \), and simulated human quadriceps eccentric torque output, \( T_{s,e} \), are shown below as Eq. (17) and (18). Equations for velocity output during concentric contraction, \( V_{s,c} \), and eccentric contraction, \( V_{s,e} \), are shown below as Eq. (19) and (20). Torque units are \([\text{Nm}]\) and velocity units are \([\text{rad/s}]\).

\[
T_{s,c} = \frac{2T_{s,oc}}{\pi} \sin^{-1} \left( 1 - \frac{\omega_d}{V_{s,o}} \right)
\]

\[
T_{s,e} = \frac{2T_{s,oe}}{\pi} \cos^{-1} \left( 1 - \frac{\omega_d}{V_{s,o}} \right) + T_{s,oe}
\]

\[
V_{s,c} = V_{s,o} \left( 1 - \sin \left( \frac{\pi T_{out}}{2 T_{s,oc}} \right) \right)
\]

\[
V_{s,e} = V_{s,o} \left( 1 - \sin \left( \frac{\pi T_{out}}{2 T_{s,oe}} \right) \right)
\]

Desired velocity, \( \omega_d \), is the velocity value from a predefined desired task velocity profile. Experimental torque output, \( T_{out} \), is fed back into the system from the motor. \( T_{s,oc}, T_{s,oe}, \) and \( V_{s,o} \) are peak values of simulated concentric torque, eccentric torque and velocity, respectively, dependent upon simulated joint angle and thus corresponding muscle length.
There is a biophysical relationship between potential force and velocity outputs of a muscle and the muscle’s length. Each muscle has an optimal length at which it will generate its maximal isometric force when maximally stimulated (180) (188). Away from this optimal length, the muscle will have reduced potential force output when maximally stimulated. Joint angle was related to quadriceps muscle length by first determining at what angle of knee flexion maximum torque occurs. Peak torque output typically occurs between 1.05 and 1.31 rad (60 and 75°) of knee flexion, depending on velocity (205) (206) (200). The peak torque angle used for this research was 1.18 rad (67.5°) with a range of motion of 2.36 rad (135°). The peak torque angle will be used as an approximation of optimal muscle length location. Taking 1 to be normalized optimal length (corresponding to peak torque angle), 0.5 to be normalized minimum length (corresponding to fully extended leg), and 1.5 to be normalized maximum length (corresponding to full flexed leg), Eq. (21) was developed to relate flexion angle, \(\theta\) in [rad], to normalized muscle length, \(L_{\text{norm}}\).

\[
L_{\text{norm}} = \frac{1}{\sqrt{135}} \theta + \frac{1}{2}
\]  

(21)

Joint angle can be related to potential torque output using an equation derived from (188) which has a peak torque at optimal muscle length. The potential torque output was estimated to drop to 25% of peak at the range of motion extremes. A similar equation can be used to describe the relationship between joint angle/muscle length and potential velocity output. The velocity of concern here is angular velocity of the simulated limb, but the angular velocity of the lower leg about the knee produced by the quadriceps muscles relates directly back to rate of shortening or lengthening of the quadriceps muscles. The maximal velocity occurs at optimal length and drops to 0 at the range of motion extremes. Eq. (22) and (23) defines the relationships between maximum potential concentric/eccentric torque output and simulated joint angle. Eq. (24) defines the relationship between maximum potential velocity output and simulated joint angle. The
scaling parameters, $a_{s, Tc}$, $a_{s, Te}$, and $a_{s, V}$ are brought in at this stage to scale peak values according to simulated neuromuscular actuator performance limits.

$$T_{s, oc} = a_{s, Tc} T_{o, max} \left( -0.5404 \theta^2 + 1.273 \theta + 0.25 \right)$$

(22)

$$T_{s, oe} = a_{s, Te} T_{o, max} \left( -0.5404 \theta^2 + 1.273 \theta + 0.25 \right)$$

(23)

$$V_{s, o} = a_{s, V} V_{o, max} \sin \left( \frac{4}{3} \theta \right)$$

(24)

Figures 21 and 22 illustrate the torque-joint angle and velocity-joint angle relationships.

![Figure 21](image)

**Figure 21.** Relationship between simulated peak torque and joint angle/motor position with $a_{s, Tc} = a_{s, Te} = 1$
Figure 22. Relationship between simulated peak velocity and joint angle/motor position with \( a_{v} = 1 \)

6.2. Scaling Level of Performance

As previously discussed, there are five parameters used to adjust the torque-velocity curve based on simulated neuromuscular actuator performance limits and limits of the system, particularly limited force output potential from the PMA (maximum rated output is 1200 N at maximum pressure and nominal length) and limited available motor torque (22 Nm).

An initial scaling was performed based on simulated neuromuscular actuator performance limits of the DTS to arrive at a scaled maximum isometric torque. The mass and height of a typical human operator (65 kg, 1.73 m), with an average level of torque production capability, was scaled down based on the amount of available motor torque. A body mass scaling factor of 0.4 was used along with a 0.625 lower leg scaling factor derived by anthropometrically scaling the limbs in relation to the scaled body mass as described in section 5. The scaled maximum isometric torque \( T_{o,max} = 57 \text{ Nm} \) allows the HQDS to simulate human quadriceps torque output against a wide range of PMA produced resistances. The maximum velocity \( V_{o,max} \) was set to 5 rad/s because knee flexion rarely exceeds this during gait (207) and the velocity limits of isokinetic dynamometers are near this value (208). The concentric torque scaling parameter
(a_{s_T}=0.25) was set at a level that allowed the PMA to perform effective control throughout the simulated task. For PMA driven haptic control to occur, PMA torque production has to exceed that of the motor throughout the task. The relationship between the concentric torque scaling parameter and velocity scaling parameter (a_{s_V}=0.4) was derived from (209). Findings indicate that at 25% of a muscle’s maximum force output, approximately 40% of the maximum velocity is observed, giving us a relative starting relationship between T_{o,max} and V_{o,max} at a submaximal level. The selection of the eccentric torque scaling parameter is based on observed voluntary eccentric behavior reported in literature as previously discussed. The assigned scaling parameter (a_{s_Te}=0.15) creates an eccentric torque curve that begins 20% below isometric, rises with increasing eccentric contraction velocity, and plateaus near isometric at 1 rad/s.

Figure 23 displays the torque velocity curve shaped using the selected parameters discussed. The large curve represents a maximally artificially activated muscle with scaled T_{o,max} and V_{o,max} based on DTS capabilities. The smaller curves represent the voluntary torque-velocity relationship at a submaximal level constructed using the three scaling parameters (a_{s_Tc}, a_{s_Te}, and a_{s_V}).

Figure 23. Torque-velocity relationships for artificially stimulated muscle (top curve) and scaled voluntary contractions (bottom curves)
The scaling parameters create a torque-velocity curve representative of scaled neuromuscular behavior at submaximal activation. The scaling parameters can also be set to one, with a corresponding change in default simulated neuromuscular actuator settings, in order to represent behavior at maximal activation. They are not set to one in this study because the purpose is to demonstrate the use of the scaling parameters to produce different relative torque-velocity relationships within the maximal performance envelope of the scaled simulated neuromuscular actuator. The scaling parameters are not intended to indicate the level of activation; rather they act to scale the relationship between torque and velocity. The activation level changes throughout the task depending on velocity and type of contraction (210), and the ability to fully activate depends on the type of muscle action (211). Activation levels during eccentric knee extension, for instance, are significantly lower than during concentric contraction (201) (186). At submaximal levels of activation (studied using artificial activation), the characteristic relationship between torque and velocity remains intact (212) (213).

The parameters listed here can easily be adjusted within the programmed control design. The values chosen for this study represent one approximation of human neuromuscular actuation behavior for our simulation based on the capabilities of the DTS.

### 6.3. Simulating Different Neuromuscular Behaviors

Different neuromuscular actuator behaviors were demonstrated using the HQDS. An ideal neuromuscular behavior type setting was used as the default. In this case, the scaling parameters \( a_{s,Tc}, a_{s,Te}, \) and \( a_{s,V} \) remained constant throughout the task, meaning the torque-velocity relationship remained consistent throughout the course of the task.

An incremental fatigue neuromuscular behavior type setting was used to simulate how a neuromuscular actuator might perform after many repetitions. Physiological fatigue was simulated by incrementally dropping the \( a_{s,Tc} \) and \( a_{s,Te} \) scaling parameters over separate trials (each representing a number of completed repetitions). The baseline (non-fatigued) torque-
velocity relationship was taken to be the ideal neuromuscular behavior type setting. Calculated fatigue scaling parameters, each representing a number of repetitions, were used to adjust $a_{s,Tc}$ and $a_{s,Te}$, and thus the torque-velocity relationship. The fatigue scaling parameters were estimated using concentric and eccentric isokinetic knee extension force data produced during fatigue inducing contractions in a study by Binder-Macleod and Lee (214). Force levels and equivalent torque levels corresponding to 20 and 40 repetitions were estimated from the data. These values were normalized relative to the starting force/torque. The 20- and 40-repetitions scenarios are meant to represent fatigue levels during the so-called fatigue phase, which is characterized by a steep decrease in force/torque output (185). After a slightly higher number of repetitions, approximately 50, the endurance phase is reached. This phase is characterized by no additional decrease in the concentric force/torque output. The normalized values calculated from the fitted data were used as the fatigue scaling parameters. The fatigue scaling parameters for the 20-repetitions scenario are 0.78 and 0.95 for concentric and eccentric torque, respectively. The fatigue scaling parameters for the 40-repetitions scenario are 0.66 and 0.89 for concentric and eccentric torque, respectively. During the 20-repetitions fatigue simulation scenario, the changes in the torque-velocity curve amount to a concentric torque decrease of 24% and an eccentric torque decrease of 7%. During the 40-repetitions fatigue simulation scenario, the changes in the torque-velocity curve amount to a concentric torque decrease of 36% and an eccentric torque decrease of 14%.

It is important to note that fatigue varies between individuals and the level of resistance which they are working against. According to Lindstrom et al., older men (test subjects in their 70s) fatigue the most, older women (test subjects in their 70s) fatigue the least, while young men (test subjects in their 20s and 30s) perceive the greatest fatigue (187). The behavior simulated here is representative of one fatiguing pattern to demonstrate performance of the control system in its ability to accommodate a simulated neuromuscular actuator. A second fatiguing pattern was simulated to demonstrate the control system’s ability to accommodate a rapidly fatiguing
neuromuscular actuator. A time-dependent fatigue neuromuscular behavior type setting was used to simulate the rapidly fatiguing neuromuscular actuator. The three scaling parameters ($a_{s_{Te}}, a_{s_{Te}},$ and $a_{s_{V}}$) changed continuously throughout the task according to the Eq. (25) based on isometric fatigue research (188).

$$a_{n_{_\text{fatigued}}} = a_{n_{_\text{initial}}} \left( 0.15 + 0.85 \left( 1 - \sin \left( \frac{\pi t}{2\Delta t_{F_{n}}} \right) \right) \right)$$

(25)

The $n$ is a placeholder representing either the concentric torque ($s_{Te}$), eccentric torque ($s_{Te}$), or velocity ($s_{V}$). The $a_{n_{\text{initial}}}$ represents the initial scaling parameter before fatigue begins. Task time is represented by $t$, while time to fatigue is represented by $\Delta t_{F}$. The times to fatigue were set as follows: $\Delta t_{F_{c}} = 40$ s, $\Delta t_{F_{e}} = 60$ s, and $\Delta t_{F_{V}} = 100$ s.

During the time-dependent fatigue simulation scenario, the changes in the torque-velocity curve are continuously changing. The changes in torque output at the desired task velocity are equivalent to a concentric torque decrease of up to 40% and an eccentric torque decrease of up to 44%, depending on task time.

Two erratic neuromuscular behavior type settings were used to simulate a neuromuscular operator with erratic volitional control. One type of erratic behavior was simulated by changing the concentric torque, eccentric torque, and velocity scaling parameters according to a sine wave function as shown in Eq. (26).

$$a_{n_{_\text{erratic-sine}}} = a_{n} + 0.05 \sin \left( \frac{1}{2\pi} t \right)$$

(26)

The $n$ is a placeholder representing either the concentric torque, eccentric torque, or velocity. Task time is represented by $t$.

During the erratic-sine wave simulation scenario, the changes in the torque-velocity curve are continuously changing, representing changes in the relationship between torque output and velocity that manifest as variations in simulated neuromuscular actuator behavior. The changes in


torque output at the desired task velocity are equivalent to a concentric torque increase of up to 24% and an eccentric torque increase of up to 32%, depending on task time.

The second type of erratic neuromuscular behavior was simulated by suddenly increasing or decreasing the concentric torque, eccentric torque, and velocity scaling parameters for a short period of time. A positive pulse with a magnitude of 0.05 was applied during the concentric phase and a negative pulse with a magnitude of 0.05 was applied during the eccentric phase. The scaling parameters changed within 0.1 s for a period of 1.5 starting at 5.5 s and 16 s. During the erratic-pulse simulation scenario, the torque-velocity relationship changes suddenly four times as the two pulses are applied and removed. The changes when away from the default curve amount to a concentric torque increase of 24% and an eccentric torque decrease of 32%.

Variation in scaling parameters affects the relationship between torque and velocity of the simulated neuromuscular actuator. Therefore, the simulated neuromuscular actuator’s response to a given force resistance changes according to the changing torque-velocity relationship. The HQDS determines the motor command signal according to the latest torque-velocity relationship in order to model time-varying neuromuscular actuator response to force resistance stimuli.

Within the HQDS, the plateau characteristic of eccentric torque beyond a certain region of low velocity is applied by using dynamic saturation. The scaled peak eccentric torque cannot exceed that of scaled peak concentric torque at each joint angle. Thus eccentric torque saturates at the isometric level (peak concentric equal to the peak isometric). A controlled starting resistance is provided for each type of neuromuscular behavior simulation so that the simulated neuromuscular actuator is better able to start the task with controlled acceleration, thus avoiding an initial velocity spike. This also avoids the need for harsh braking which can cause torque spiking at the onset of the exercise. This type of isometric preloading is typically referred to as isometric preactivation (183). The level of isometric preloading is automatically set to the scaled torque corresponding to the desired task velocity at the starting position. Additional code is designed to appropriately simulate reactions to stimuli during the isometric hold. If the torque
resistance felt by the simulated neuromuscular actuator exceeds its scaled isometric torque capability, then the HQDS command will change in response. This results in movement (representing dropping of the lower leg from the isometric hold position), triggering the controller to adjust the resistance to a level at which the simulated neuromuscular actuator can maintain isometric behavior.
7. **HAPTIC CONTROL OF PMA ISOKINETIC STRENGTH TRAINING SYSTEM**

7.1. **Control Structure**

The basic structure of the control system is shown in Figure 24. Each block represents a set of control calculations performed at each iteration with data flow between blocks as shown. The Trajectory Generator is the source of desired outputs used at each iteration of control calculations. Trajectory Generator outputs include desired PMA displacement and velocity, angular acceleration of the motor (simulated lower leg angular acceleration), velocity of motor (simulated lower leg angular velocity), and motor position (simulated joint angle).

The Task Motion Controller (TMC) block runs calculations to determine the proper torque/force resistance command to send to the PMA Motion Controller (PMC) based on desired motor acceleration, velocity, and position as well as experimental velocity and position outputs from the simulated neuromuscular actuator. The PMC block runs calculations to determine the correct pressure to command to the proportional pressure regulator with the goal of producing the correct pressure change and thus the correct PMA force and displacement outputs to correctly track motor movement. Closed loop pressure control performed by the PPR is contained within the PPR+PMA block. The combination of the TMC and PMC provides control for both the PMA force output and PMA displacement. Closed loop control of the system is based on the position, velocity and torque outputs of the simulated neuromuscular actuator (motor).

The Human Quadriceps Dynamic Simulator (HQDS) block runs calculations to determine motor commands that will best simulate a neuromuscular actuator’s response to the system feedback. Closed loop motor control is contained within the Motor/HumanSim block.
LabVIEW was used as the controller and signal transfer interface. Data collection and signal processing also took place within LabVIEW.

![Diagram of Haptic PMA control system diagram](image)

**Figure 24. Haptic PMA control system diagram**

7.2. Task Motion Controller (TMC)

The control law used to calculate the torque resistance command within the Task Motion Controller is based on the system’s set of combined moments or torques as shown in Figure 25. The moments are summed and set equal to moment of inertia times the simulated joint angular acceleration as shown in Eq. (27). A nonlinear adaptive control method using full state feedback is applied to the resulting second order nonlinear equation.

![Diagram illustrating system forces](image)

**Figure 25. Diagram illustrating system forces**
In the sum of moments equation, Eq. (27), $T_{h,n}$ is the predicted estimate of human neuromuscular actuator produced torque which is a nonlinear function of knee flexion angle, $\theta$, and angular velocity, $\dot{\theta}$. The $n$ is a place holder representing either the concentric or eccentric based torque estimate as defined by Eq. (28) or (29). Peak isometric torque is represented by $T_{o,max}$, and peak angular velocity by $V_{o,max}$. The terms $a_{Tc}, a_{Te}$, and $a_{V}$ within Eq. (28) and (29) are scaling parameters based on predicted estimates of system outputs. They allow for the pre-trial adjustment of predicted neuromuscular actuator torque estimates. $T_R$ is the torque resistance command which will ultimately be calculated by the controller within the TMC. The middle term of Eq. (27), $T_{leg}$, is the estimated torque needed to overcome the weight of the lower leg and foot. Gravitational acceleration is represented by $g$, mass of the lower leg and foot by $m$, and the distance between the center of rotation and the combined center of mass of the lower leg and foot by $l_{com}$. The angle $\phi$ is the angle at which the center of mass of the combined lower leg and foot is located at 1.57 rad (90°), the bent knee position. The combined center of mass is located off of the longitudinal axis of the lower leg because of the added mass of the foot. The terms $c$ and $b$ are included as coefficients of the human neuromuscular actuator torque term and leg torque term,
respectively. These coefficients are used to adjust the terms as adaptation takes place during the course of the task.

A nonlinear adaptive control method based on sliding mode control is applied to Eq. (27) using methodology described in (215). Sliding mode control is a robust nonlinear control method which can handle parameter uncertainty (inaccuracies of terms included in the model) and disturbances. A weighted combination of velocity and position error is defined according to Eq. (30). The combined error, \( s \), is also referred to as the time-varying sliding surface.

\[
\begin{align*}
\dot{s} & = \dot{e} + \lambda_c e = \dot{\theta} + \lambda_o \tilde{\theta} \\
\theta & = \theta_{out} - \theta_d \text{ where } out \text{ signifies the output of the system and } d \text{ signifies the time dependent value within the desired profile. Tracking the states of theta (simulated joint position) is equivalent to remaining on the surface } s. \text{ The goal is to keep } s \text{ at zero and remain on the sliding surface. The system’s motion on the sliding surface is defined as } \dot{s} = 0. \text{ An expression for } \dot{s} \text{ is shown as Eq. (31).}
\end{align*}
\]

\[
\dot{s} = \ddot{\theta} - \dot{\theta}_d + \lambda_o \tilde{\theta} \tag{31}
\]

Combining Eq. (31) with an expression for \( \dot{\theta} \) derived from Eq. (27), \( \dot{s} \) can now be written as Eq. (32).

\[
\dot{s} = -\dot{\theta}_d + \lambda_o \tilde{\theta} - c_\theta \frac{T_h}{I} + b \frac{T_{leg}}{I} + T_R \frac{R}{I} \tag{32}
\]

Eq. (32) is set equal to zero and rearranged for \( T_R \) to form an approximate control law that helps the system remain on the sliding surface. An addition term (-k\( s \) where \( k > 0 \)) is added to help constrain system trajectories to point towards the sliding surface (see Eq. (33)).

\[
T_R = \hat{I} \left( \dot{\theta}_d - \lambda_o \tilde{\theta} \right) - k \left( \dot{\theta} + \lambda_o \tilde{\theta} \right) + cT_h(\theta, \dot{\theta}, t) - bmgI_{com} \cos(\phi - \theta) \tag{33}
\]

\( I, c \) and \( b \) within the control law are now replaced with their estimated values \( \hat{I}, \hat{c} \), and \( \hat{b} \) as shown in Eq. (34).
The tracking error dynamics expression for this control law, $I\dot{s} + ks$, is equal to zero if parameters $I$, $c$, and $b$ are known, giving exponential convergence of $s$, and thus convergence of $e$. These parameters are unknown in our case and will be adapted so that they converge to appropriate values. By defining parameter errors as shown in Eq. (35) through Eq. (37), the tracking error dynamics expression becomes Eq. (38).

$$\tilde{I} = \hat{I} - I$$

$$\tilde{c} = \hat{c} - c$$

$$\tilde{b} = \hat{b} - b$$

$$I\dot{s} + ks = \tilde{I}\left(\dot{\theta}_d - \lambda_o \dot{\theta}\right) + \tilde{c}T_h(\theta, \dot{\theta}, t) - \tilde{b}mgl_{com}\cos(\varphi - \theta)$$

The tracking error dynamics can be converted into the Laplace domain as shown in Eq. (39) where $p$ is the Laplace variable. Note that Eq. (39) contains time and frequency domain notation, which is common in adaptive literature (215).

$$s = \frac{1}{p + \frac{k}{I}} \left[\tilde{I}\left(\dot{\theta}_d - \lambda_o \dot{\theta}\right) + \tilde{c}T_h(\theta, \dot{\theta}, t) - \tilde{b}mgl_{com}\cos(\varphi - \theta)\right]$$

For an error signal of the form shown in Eq. (39) with a transfer function that is strictly positive real, the estimated parameters $\hat{I}$, $\hat{c}$, and $\hat{b}$ should be varied according to Eq. (40) through Eq. (42) where $\gamma$, the adaptation gain, is a value between 0 and 1 used to regulate the speed of adaptation.

$$\dot{I} = -\gamma_I sgn(I)\left(\dot{\theta} + \lambda_o \dot{\theta}\right)\left(\dot{\theta}_d - \lambda_o \dot{\theta}\right)$$

$$\dot{c} = -\gamma_c sgn(I)\left(\dot{\theta} + \lambda_o \dot{\theta}\right)T_h$$

$$\dot{b} = -\gamma_b sgn(I)\left(\dot{\theta} + \lambda_o \dot{\theta}\right)$$
\[ \dot{b} = -\gamma_b \text{sgn}(I) \left( \dot{\theta} + \lambda_c \ddot{\theta} \right) mgl_{\text{com}} \cos(\phi - \theta) \]  

(42)

These adaptation laws ensure that \( s, \tilde{I}, \tilde{c}, \) and \( \tilde{b} \) are globally bounded. Global tracking convergence (\( s \to 0 \) as \( t \to \infty \)) can be shown using the Lyapunov function candidate, Eq. (43), and its derivatives along system trajectories, Eq. (44) and (45).

\[ V = |I| s^2 + \left[ \frac{1}{\gamma_I} \tilde{I}^2 + \frac{1}{\gamma_c} \tilde{c}^2 + \frac{1}{\gamma_b} \tilde{b}^2 \right] \]

(43)

\[ \dot{V} = -2k |s| s^2 \leq 0 \]

(44)

\[ \ddot{V} = -4k |s| \dot{s} \]

(45)

The boundedness of \( s, \tilde{I}, \tilde{c}, \) and \( \tilde{b} \) implies boundedness of \( \dot{s} \), and therefore boundedness of \( \dot{V} \).

Boundedness of \( \dot{V} \) indicates uniform continuity of \( \dot{V} \). With uniform continuity of \( \dot{V} \) and \( \dot{V} \leq 0 \), invariant set theorem and Barbalat’s lemma (Lyapunov-like analysis) (215) tell us that \( s \to 0 \) as \( t \to \infty \). With \( s \) acting as the combined error, this also indicates that both position and velocity tracking error will converge to zero.

On-line adaptation of model parameters is designed to maintain consistent performance of the system in the presence of uncertainty or system variation (215). Adaptation of \( \tilde{I} \) acts to adjust the contribution of the inertial term according to system outputs. It has a starting value equal to the estimated system inertia. For the motor used in this study, it is 0.00104 kg\(\cdot\)m\(^2\). For a lower leg and foot appropriately scaled according to the neuromuscular actuator and HQDS previously described, it is 0.071 kg\(\cdot\)m\(^2\). Adaptation of \( \tilde{c} \) acts to adjust the predicted estimate of the human neuromuscular actuator torque output, which is useful if the predicted value is inaccurate or changes with time. As the most significant term, it also helps to account for differences between actual and predicted model PMA behavior and changes in PMA behavior.
due to sequential use, age or environmental changes. Adaptation of $\hat{b}$ acts to adjust the torque due to gravitational effects. Although not significant during simulation, this term will be help in finding the proper gravity contribution for different human operators during the task. Initial lower leg/foot mass and center of mass calculations may not be accurate for a true human operator, affecting the accuracy of the leg torque term within the control law. Center of mass may even change during the task due to slight changes in the knee’s center of rotation. Therefore, adaptation on the term has potential importance and value. The starting value of $\hat{c}$ and $\hat{\hat{b}}$ is 1. Adaptation gains used for $\hat{b}$, $\hat{c}$ and $\hat{\hat{b}}$ are 0.02, 0.3, and 0.3, respectively.

An adaptation dead zone was implemented for $\hat{c}$ and $\hat{\hat{b}}$, acting to stop adaptation if the combined error falls below a certain level (see Eq. (46)). The dead zone helps to reduce the negative impact or possible instability caused by small disturbances based on the assumption that small tracking errors contain mostly noise and disturbance (215).

$$\hat{\eta}_{\text{deadzone}} = \begin{cases} \hat{\eta} & |s| \geq 0.05 \\ 0 & |s| < 0.05 \end{cases} \quad (46)$$

In its attempt to improve performance with parametric uncertainties present in the model, sliding mode control may result in high control activity. The high control activity can excite unmodeled dynamics, and thus a balance has to be achieved between tracking error performance and the level of control activity. Control activity is very sensitive to the control bandwidth and large changes in control activity can be seen with small adjustments in bandwidth. In this study, a control bandwidth, $\lambda_o$, of 2 and a gain, $k$, of 35 were found to provide the best balance between tracking performance and level of control activity.

7.3. PMA Motion Controller (PMC)

The operation of the PMA Motion Controller (PMC) is based on the three-element phenomenological model consisting of a spring element, damping element and contractile
element arranged in parallel. An illustration of the model and governing equations of motion are shown in section 3. This model has been shown to sufficiently predict force and length output from the PMA (18) (16) and characterization of the model has already been performed (19).

As previously explained, $K_{PMA}$, $B_{PMA}$ and $F_{ce}$ are pressure dependent terms. By expanding these terms (corresponding to the spring element, damping element, and contractile element force) and rearranging governing equations, an inverse of the phenomenological model was derived. This is used as the basis of the controller calculations used to calculate the PMA pressure. In the inverse model, shown as Eq. (47), pressure ($P$) is the output, whereas force resistance command ($F_R$), PMA displacement ($x$), and PMA velocity ($\dot{x}$) are the inputs. $F_R$ is the resistance command calculated by the TMC. The PMA velocity and displacement terms are calculated at each iteration by the Trajectory Generator. The coefficients in the inverse model refer to model characterization values (see Table 4 in section 3).

$$P = \frac{b_f - b_y \dot{x} - b_k x - F_R}{a_k^2 x + a_k x - a_f}$$

(47)

Within the PMC, there is internal feedback to adjust the characterization values based on the initially calculated pressure command at each iteration and the current type of PMA operation (either contraction or relaxation).

In addition to inverse model based control within the PMC, there is an additional layer of control which is designed to improve performance during the isometric hold phase of the task. This performance improvement is only necessary if there is an interest in controlling the isometric resistance to a desired level (either a predefined isometric resistance or one derived from the isometric torque-position relationship). During the isometric phase, the PMA remains static or nearly static ($\dot{x} \approx 0$), and there is a small displacement command based on the PMA configuration design ($x \approx 0$). In addition, there is nothing to drive correction originating from the TMC if motor position and velocity are correct. TMC operates based on motor position and
velocity errors. The motor position and velocity may be correct during the hold phase while the steady state force does not meet the desired isometric force. Velocity remains near zero during the hold phase unless the PMA resistance triggers a change in simulated neuromuscular actuator behavior. The inverse model based on TMC input fails to calculate the pressure needed to achieve the ideal isometric resistance. A PID controller was used to adjust the force output of the PMA. The PID controller operates by adjusting the force resistance command according to force error defined as the target force (as defined by the simulated neuromuscular actuator’s torque-position curve or predefined isometric hold force) minus the experimental force output (converted from torque to force using the 0.027 m moment arm of the motor shaft-spool-cable assembly). The PID adjustment based on force error is only active during the isometric hold phase of the task.

In general, using force feedback is not necessarily useful for this system. With no desired value with which to compare torque feedback, error cannot be calculated and used to make system adjustments. There is no desired force because with a true human operator, the torque-velocity-position curve will only be estimated based on prior testing. It is not a known value to which outputs should be compared for true human operators. Also, there is no direct way to determine the true resistance produced by the PMA. Here we are only working with torque produced by the motor, which will be produced by the human upon human implementation. This torque measurement is a combination of PMA resistance, inertia effects, and any torque required to move mass. It is also important to note that the torque output will not follow the control variable, \( T_R \), which is the torque equivalent of the force resistance command. There are major adjustments made to \( F_R/T_R \) according to motor position and velocity tracking errors in order to track desired movement.

The fixed proportional, integral and derivative gains used in the supplemental isometric PID controller were as follows: 0.5, 2, and 0.2. Nonlinear and adaptive methods for on-line PID gain adjustment were tested, but satisfactory results were achieved after tuning with fixed gains.
Simpson’s Rule, a discrete integration method, is used within LabVIEW for integration of error. The 2nd order central method of discrete differentiation is used to take the derivative of error.

It has been mentioned that due to the PMA configuration design, PMA displacement is near zero during the isometric hold phase. The inverse model would most likely perform better during this phase with greater displacement. However, that was not an option in this case because by adding PMA displacement to this position, the PMA displacement would have to be shifted throughout the task. This would result in a reduction of available peak resistance force throughout the task. It is not advantageous to simply stay away from controlling the PMA near nominal length because this is where the highest potential force output is produced.

7.4. Simulation Studies

Simulations were performed within the MATLAB 7.11.0 (R2010b) Simulink environment prior to the controller’s implementation on the DTS. Simulations of the PMA-human quadriceps dynamic simulator interaction indicated that effective, robust simulated neuromuscular actuator position control was possible at different task velocities despite force and velocity disturbances, lag due to time constants, as well as time delays. Saturation limits were implemented to control parts of the system within realistic physical limits. High inertial loading of the system, equivalent to a lower leg in normal environment, was also simulated successfully. The torque required to lift the leg as a function of joint angle is included as a term within the TMC control law. It does not, however, play a role when no torque is required to move the leg itself. Once a human is operating the device (in a 1-g environment), the gravity term plays a role and is adapted based on system outputs. Simulations indicate that the controller is effective and robust, able to handle large disturbances. With an active gravity term in addition to a larger inertial term, both of which adapt based on system outputs, simulation studies indicated that the impact of disturbances on the system is reduced.
7.5. Experimental Procedure

An evaluation of motor performance was first completed in two steps. First, an evaluation of motor step response was performed. A command voltage profile was sent to the motor corresponding to steps in velocity equaling \(-0.314\ \text{rad/s}\) at 5 s, 0 rad/s at 10 s, 0.314 rad/s at 15 s, and finally 0 rad/s at 20s. The second part of the motor performance evaluation used the experimental setup and some of the control structure previously described. This test was intended to show the capability of the motor to follow a desired motion profile against PMA resistance while no resistance adjustments are being made by the TMC (no haptic control). The HQDS was not used. Instead, velocity was commanded directly to the motor according to a 0.314 rad/s desired motion profile using the Trajectory Generator. The resistance command followed an accommodating profile; peaking between 10 and 12 Nm at the simulated joint angle where peak simulated scaled human torque production typically occurs at the commanded 0.314 rad/s velocity. This level of resistance is similar in magnitude to what is produced during closed loop testing. The PMC operated as previously described, with the exception of replacing the TMC output with the predefined resistance command.

The closed loop control system was then tested using the complete system, including the TMC and HQDS. The experimental trials tested several desired task velocities and several simulated neuromuscular behavior types. Different velocity disturbances were applied to further test robustness of the control system. The task velocities tested include 0.175 rad/s (10 °/s), 0.314 rad/s (18 °/s), and 0.524 rad/s (30 °/s). Five repetitions at each velocity were run. The simulated neuromuscular behavior types included ideal (baseline), fatigued (after 20 repetitions and 40 repetitions), fatigue with rapid onset, erratic with sine wave variation of torque-velocity curve parameters, and erratic with positive/negative pulse variation of torque-velocity curve parameters. Five repetitions of each simulated neuromuscular behavior type were run at the 0.314 rad/s desired task velocity.
The simulated neuromuscular behavior type was set within the HQDS before each trial or set of trials. The HQDS produces motor velocity commands according to torque and position feedback applied to the simulated neuromuscular actuator time-varying torque-velocity-position performance envelope defined by Eq. (17) through (24) (specific equations are dependent on neuromuscular behavior type).

Velocity disturbances added to the velocity calculation for the baseline neuromuscular behavior type included random noise with a 0.05 rad/s amplitude (equivalent to 2.86 °/s) and a velocity pulse of -0.1 rad/s (equivalent to -5.73 °/s) from 5 to 10 s and from 15 to 20 s.

The time dependent desired motion profiles were created prior to the trials (see Appendix G). The desired motor velocity and position profiles were based on desired task speed, but acceleration to desired speed was included as a consideration. PMA displacement must remain equal to motor displacement in order to maintain proper force transfer between the PMA and motor. PMA displacement was therefore calculated based on desired motor shaft position. The motor rotation can be related to arc length of wrapping/unwrapping cable around the motor shaft spool allowing for the calculation of desired PMA displacement. Knowing PMA displacement over time allows us to calculate desired PMA velocity. Interpolation between specified points in the desired motion profiles was performed within LabVIEW so that the correct desired point, corresponding to the exact task time, was used within the controller.

The controller elements described were implemented using the LabVIEW interface and a data acquisition card (National Instruments PCI-6025E). Data from sensors was sent through a fifth order low pass finite impulse response (FIR) filter before the voltages were converted to proper units for use in control calculations. The advantage of using an FIR filter is that this type of filter keeps the delay for all frequency components the same. Voltages are collected and commanded at a rate of 1 kHz with a 100 sample buffer size using a NI 6025E data acquisition card.
Calibrations had to be completed to relate incoming sensor voltage to meaningful units that could be used within the control calculations. The relationship between LVDT voltage and PMA displacement \( x \) (in mm) is defined by Eq. (48). The relationship between rotational potentiometer voltage and motor shaft position \( \theta_{out} \) (in rad) is defined by Eq. (49). This equation corresponds to a rotational potentiometer voltage of 0.722656 at 1.57 rad of flexion, corresponding to a motor position of 9682 counts. The relationship between the voltage representing motor current feedback and motor torque \( T_{out} \) (in Nm) is defined by Eq. (50). The \( \text{Gain}_{IFB} \) as set in the motor configuration was 0.7125 V/amp. The relationship between the voltage representing motor velocity feedback and simulated joint angular velocity (in rad/s) is defined by Eq. (51). The \( \text{Gain}_{vel} \) as set in the motor configuration was 410 V/kRPM.

\[
x_{out} = 10.206V - 13.058
\]  \hspace{1cm} (48)

\[
\theta_{out} = -1.243027V + 2.467250
\]  \hspace{1cm} (49)

\[
T_{out} = 2.85 \frac{|V|}{\text{Gain}_{IFB}}
\]  \hspace{1cm} (50)

\[
\omega_{out} = \frac{V + 0.0295 \left( \frac{100 \pi}{3} \right)}{\text{Gain}_{vel}}
\]  \hspace{1cm} (51)

Voltage command outputs to the proportional pressure regulator were calculated based on a desired pressure \( P \) in kPa as shown in Eq. (52). Voltage command outputs to the motor driver were calculated based on a desired velocity \( \omega \) in rad/s as shown in Eq. (53). The \( \text{Gain}_{AV} \) as set in the motor configuration was 0.001333 kRPM/V. See Appendix D for more information on calibration procedures. See Appendix C for more information regarding motor configuration.

\[
V_{PPR} = 0.016683P + 0.1589
\]  \hspace{1cm} (52)

\[
V_{motor} = \frac{3}{\text{Gain}_{AV} 100 \pi} \omega
\]  \hspace{1cm} (53)
7.6. Statistical Analysis

Statistical analysis was performed to evaluate the following system parameters: motor position (simulated joint angle), motor velocity (simulated angular velocity of lower leg), motor torque (measure of transferred PMA force resistance), and PMA displacement. Root Mean Square Error (RMSE) was calculated for each parameter for each trial according to Eq. (54). The value of RMSE for each trial was used as a measure of the error between the desired values and the experimental outputs of the system, and thus a measure of system performance. It is beneficial to use RMSE over the mean, standard deviation or mean error because RMSE is a measure of not only variance of the errors, but also the bias. However, it is particularly sensitive to occasional large errors. The variable \( n \) in Eq. (54) is the number of compared output/desired data points used to calculate the RMSE. The desired values of motor position, motor velocity, and PMA displacement are defined in the desired motion profiles (as listed in Appendix G). The desired torque is not a predetermined value. The target torque is based on the torque-velocity-position relationship for the specific simulated neuromuscular actuator set within the HQDS. The target torque is the torque corresponding to the desired velocity (as defined in the desired motion profile) and the actual position output. This value is not to be confused with the control variable, \( T_R \), which is the torque command determined by the TMC.

\[
RMSE = \sqrt{\frac{\sum_{i=1}^{n} (\text{output}_i - \text{desired}_i)^2}{n}}
\]  

The calculated RMSE provides one value which can be conveniently used to compare control performance for different phases of the task (concentric, eccentric, and the entire phase), task velocities (0.175 rad/s, 0.314 rad/s, and 0.524 rad/s), and simulated neuromuscular behavior types (simulated fatigue and erratic operational behavior).

There is some interest in studying the differences in control performance across the simulation scenarios and phases of the task in order to identify control limitations. This will shed
light on what is going on in the system and how to potentially improve system performance. To compare the effects of simulation scenario, phase, and the interaction of simulation scenario and phase on the control performance, an ANOVA was performed using JMP 9.0.0 (SAS Institute Inc.). The control performance results (in terms of RMSE) for the position, torque, and PMA displacement parameters within each subgroup were compared. The coefficient of variation of the RMSE (CV RMSE) was compared for the velocity parameter. The CV RMSE of velocity over the desired isokinetic range of motion was calculated for each trial according to Eq. (55) in order to normalize the RMSE for comparison across the different simulations scenarios. This was important for comparison of the velocity due to the difference in velocity outputs across different scenarios.

\[
CV \text{ RMSE} = \frac{RMSE}{\text{output mean}} \tag{55}
\]

The null and alternative hypotheses for the ANOVA are listed as Eq. (56) through (58) where \( \tau \) represents the effect of phase (concentric or eccentric) and \( \beta \) represents the effect of simulation scenario (eight in total representing different simulated neuromuscular behavior types and task speeds). Finding proof to reject \( H_o \) of Eq. (56) signifies that simulation scenario has an effect on control performance. Finding proof to reject \( H_o \) of Eq. (57) signifies that phase, whether operating in concentric or eccentric mode, has an effect on control performance. Finding proof to reject \( H_o \) of Eq. (58) signifies that the interaction of simulation scenario and phase has an effect on control performance, meaning that the effect of simulation scenario on the control performance depends on phase (or vice-versa).

\[
H_o : \tau_1 = \tau_2 = 0
\]

\[
H_{A \tau} : \text{at least one } \tau_i \neq 0 \tag{56}
\]

\[
H_o : \beta_1 = \beta_2 = ... = \beta_8 = 0
\]

\[
H_{A \beta} : \text{at least one } \beta_j \neq 0 \tag{57}
\]
\[
H_o : (\tau_\beta)_{ij} = 0 \quad \forall \ i, j
\]
\[
H_A : \text{at least one } (\tau_\beta)_{ij} \neq 0
\]  

(58)

After identifying differences in controller performance between simulation scenarios and phases, the results were evaluated to determine controller limitations and possible explanations for controller performance.

### 7.7. Results

#### 7.7.1. Baseline Motor Performance

The motor’s response to the 0.314 rad/s step command velocity profile against no load is shown in Figure 26. The time constant is calculated to be 0.18 s, with a settling time (within 10% of desired value) of 0.36 s. There is oscillatory behavior around the command value throughout the length of the command, although this behavior does not result in significant position error. After settling, the RMSE is 0.015 rad/s, based on the error between the actual velocity and the command. This oscillatory behavior may be attributable to the method of closed loop motor velocity control (and associated tuning), low motor damping, or signal processing error. It will become important as we study the velocity outputs of the closed loop haptic control system.
Baseline motor performance testing was completed to show the capability of the motor to follow a desired motion profile against PMA resistance while no resistance adjustments are being made by the TMC (no haptic control). PMA resistance was a predetermined profile similar in magnitude to what is produced during haptic control testing (position dependent torque profile with a maximum of 10 to 12 Nm). Motor velocity is closed loop controlled according to the desired motor velocity, making this a representation of the best case scenario of motor control performance against resistance with no haptic control. Figure 27 shows the desired and experimental position and velocity outputs.
Figure 27. Baseline motor performance (automatic command of desired motion profile with no haptic control)

The average RMSE values of motor shaft position, velocity, torque, and PMA displacement for concentric, eccentric and the entire task from five trials are shown in Table 9.

Table 9. Baseline motor performance results for independent motor control; no haptic control applied

<table>
<thead>
<tr>
<th>Phase</th>
<th>Position [rad]</th>
<th>Velocity [rad/s]</th>
<th>Torque [Nm]</th>
<th>PMA Displacement [mm]</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Avg RMSE</td>
<td>Std Dev</td>
<td>Avg RMSE</td>
<td>Std Dev</td>
</tr>
<tr>
<td>Motor Control Only</td>
<td>Concentric</td>
<td>0.123</td>
<td>0.003</td>
<td>0.094</td>
</tr>
<tr>
<td></td>
<td>Eccentric</td>
<td>0.083</td>
<td>0.001</td>
<td>0.071</td>
</tr>
<tr>
<td></td>
<td>Entire Task</td>
<td>0.088</td>
<td>0.001</td>
<td>0.072</td>
</tr>
</tbody>
</table>
7.7.2. Haptic Control Performance

7.7.2.1. Performance Evaluation using RMSE

The RMSE values for all trials are shown in Figures 28 through 31. Each figure contains information about one parameter investigated; either motor position (simulated joint angle), angular velocity, torque or PMA displacement. The individual RMSE values for each repetition are displayed as points. Range bars are displayed to show the minimum and maximum RMSE value. In addition to displaying differences among repetitions, Figures 28 through 31 also clearly display differences in RMSE between the different simulation scenarios (each shown as a separate column) and different phases of the task (shown within each simulation scenario column).

Figure 28. Position RMSE for each simulation scenario and each task phase
Figure 29. Velocity RMSE for each simulation scenario and each task phase

Figure 30. Torque RMSE for each simulation scenario and each task phase
**Figure 31. PMA Displacement RMSE for each simulation scenario and each task phase**

Table 10 lists RMSE averages and standard deviations for each simulation scenario (see left column) and each parameter (see top row). The RMSE values are specified for the concentric and eccentric phases of the task, as well as the entire task. In addition, a subset of the data is evaluated as designated by the Concentric, $\alpha = 0$ and Eccentric, $\alpha = 0$. These concentric/eccentric subsets correspond to data where there is no acceleration or deceleration specified in the desired motion profile. It represents the true region of desired isokinetic movement. For the 0.175 rad/s motion profile, the region is between 1.52 and 0.14 rad (87 to 8˚ of flexion). For the 0.314 rad/s motion profile, the region is between 1.40 and 0.24 rad (80 to 14˚ of flexion). For the 0.524 rad/s motion profile, the region is between 1.31 and 0.35 rad (75 to 20˚ of flexion). For concentric and eccentric phases individually, position RMSE values range from 0.009 to 0.030 rad, velocity RMSE from 0.018 to 0.075 rad/s, velocity RMSE for isokinetic range from 0.015 to 0.044 rad/s, torque from 0.539 to 1.46 Nm, and PMA displacement RMSE from 0.258 to 0.616 mm, depending on the simulation scenario and phase of the task.
<table>
<thead>
<tr>
<th>Simulation Scenario</th>
<th>Phase</th>
<th>Position [rad]</th>
<th>Velocity [rad/s]</th>
<th>Torque [Nm]</th>
<th>PMA Displacement [mm]</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.314 rad/s</td>
<td>Concentric</td>
<td>0.0099 0.0002</td>
<td>0.0184 0.0020</td>
<td>0.7739 0.0237</td>
<td>0.3812 0.1156</td>
</tr>
<tr>
<td></td>
<td>Concentric, α = 0</td>
<td>0.0090 0.0001</td>
<td>0.0145 0.0024</td>
<td>0.7243 0.0143</td>
<td>0.3815 0.1262</td>
</tr>
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<td>0.0143 0.0013</td>
<td>0.0273 0.0022</td>
<td>0.6971 0.0874</td>
<td>0.3920 0.0207</td>
</tr>
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<td></td>
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<td>0.0183 0.0022</td>
<td>0.5702 0.0809</td>
<td>0.3922 0.0239</td>
</tr>
<tr>
<td></td>
<td>Entire Task</td>
<td>0.0125 0.0015</td>
<td>0.0223 0.0011</td>
<td>0.7159 0.0448</td>
<td>0.3735 0.0609</td>
</tr>
<tr>
<td>0.314 rad/s Rapid Fat</td>
<td>Concentric</td>
<td>0.0147 0.0014</td>
<td>0.0286 0.0017</td>
<td>1.0414 0.0973</td>
<td>0.2880 0.0524</td>
</tr>
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<td>Concentric, α = 0</td>
<td>0.0133 0.0021</td>
<td>0.0162 0.0019</td>
<td>0.8331 0.0819</td>
<td>0.2783 0.0622</td>
</tr>
<tr>
<td></td>
<td>Eccentric</td>
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<td>0.0428 0.0024</td>
<td>0.7716 0.0288</td>
<td>0.4850 0.0595</td>
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<td>Eccentric, α = 0</td>
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</tr>
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<td>Entire Task</td>
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<td>0.0318 0.0021</td>
<td>0.8340 0.0552</td>
<td>0.3805 0.0507</td>
</tr>
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<td>Concentric</td>
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<td>0.0562 0.0026</td>
<td>1.1963 0.0488</td>
<td>0.2653 0.0994</td>
</tr>
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<td></td>
<td>Concentric, α = 0</td>
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<td>0.0262 0.0045</td>
<td>1.2908 0.0373</td>
<td>0.2512 0.1300</td>
</tr>
<tr>
<td></td>
<td>Eccentric</td>
<td>0.0219 0.0015</td>
<td>0.0751 0.0057</td>
<td>0.9637 0.2106</td>
<td>0.6164 0.0582</td>
</tr>
<tr>
<td></td>
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<td>Entire Task</td>
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<td>0.0537 0.0025</td>
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<td>0.4973 0.0568</td>
</tr>
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<td>0.314 rad/s 40rep Fat</td>
<td>Concentric</td>
<td>0.0241 0.0012</td>
<td>0.0383 0.0018</td>
<td>0.8541 0.0926</td>
<td>0.3207 0.0448</td>
</tr>
<tr>
<td></td>
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<td>0.0238 0.0020</td>
<td>0.0314 0.0034</td>
<td>0.9053 0.0668</td>
<td>0.2752 0.0669</td>
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<td></td>
<td>Eccentric</td>
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<td>0.0400 0.0041</td>
<td>0.6068 0.0982</td>
<td>0.4544 0.0574</td>
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<td></td>
<td>Eccentric, α = 0</td>
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<td>0.0251 0.0035</td>
<td>0.5610 0.0954</td>
<td>0.3448 0.0382</td>
</tr>
<tr>
<td></td>
<td>Entire Task</td>
<td>0.0170 0.0007</td>
<td>0.0341 0.0012</td>
<td>0.6991 0.0611</td>
<td>0.3948 0.0596</td>
</tr>
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<td>Concentric</td>
<td>0.0305 0.0020</td>
<td>0.0459 0.0024</td>
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<td>0.4431 0.0374</td>
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<td></td>
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<td>0.0416 0.0045</td>
<td>0.9302 0.0299</td>
<td>0.4192 0.0356</td>
</tr>
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<td></td>
<td>Eccentric</td>
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<td>0.8003 0.0536</td>
<td>0.4770 0.0739</td>
</tr>
<tr>
<td></td>
<td>Eccentric, α = 0</td>
<td>0.0095 0.0017</td>
<td>0.0233 0.0033</td>
<td>0.8316 0.0493</td>
<td>0.3961 0.1422</td>
</tr>
<tr>
<td></td>
<td>Entire Task</td>
<td>0.0191 0.0009</td>
<td>0.0357 0.0008</td>
<td>0.8291 0.0712</td>
<td>0.4842 0.0529</td>
</tr>
<tr>
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<td>Concentric</td>
<td>0.0195 0.0020</td>
<td>0.0282 0.0013</td>
<td>0.9314 0.0556</td>
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<tr>
<td></td>
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<td>0.0159 0.0022</td>
<td>0.7799 0.0596</td>
<td>0.2408 0.0778</td>
</tr>
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<td></td>
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<td>0.0343 0.0025</td>
<td>0.5391 0.0396</td>
<td>0.5515 0.1054</td>
</tr>
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<td></td>
<td>Eccentric, α = 0</td>
<td>0.0141 0.0020</td>
<td>0.0200 0.0029</td>
<td>0.5514 0.0170</td>
<td>0.5420 0.1294</td>
</tr>
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<td></td>
<td>Entire Task</td>
<td>0.0148 0.0015</td>
<td>0.0281 0.0010</td>
<td>0.7108 0.0403</td>
<td>0.3954 0.0496</td>
</tr>
<tr>
<td>0.314 rad/s Erratic-pulse</td>
<td>Concentric</td>
<td>0.0171 0.0010</td>
<td>0.0340 0.0013</td>
<td>1.1056 0.0288</td>
<td>0.4368 0.0800</td>
</tr>
<tr>
<td></td>
<td>Concentric, α = 0</td>
<td>0.0141 0.0011</td>
<td>0.0293 0.0015</td>
<td>1.2282 0.0409</td>
<td>0.4896 0.1235</td>
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<td></td>
<td>Eccentric</td>
<td>0.0206 0.0013</td>
<td>0.0478 0.0026</td>
<td>0.9909 0.0937</td>
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<td></td>
<td>Eccentric, α = 0</td>
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<td>Entire Task</td>
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<td>0.0374 0.0007</td>
<td>1.0783 0.0417</td>
<td>0.4343 0.0447</td>
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Table 10. Average RMSE and corresponding standard deviation for each simulation scenario and task phase
It is possible to evaluate differences in RMSE values using the displayed and tabulated data, but analysis of variance was used in a formal test for finding potential effects of simulation scenario, phase, and the interaction between simulation scenario and phase on control performance as measured by RMSE. The objective of this analysis was to identify whether differences in control performance did indeed occur during different simulation scenarios and during different phases of the task so that system behavior could be studied further with the identification of potential strengths and limitations of the system.

The performed trials included five repetitions of eight different simulation scenarios, each with two phases (concentric and eccentric). The entire task RMSE was not included in this analysis as the isokinetic concentric and eccentric phases are of greater interest than the combined concentric, isometric and eccentric. Also, preliminary analysis (using t-tests) revealed a statically significant difference between the concentric and eccentric phases within many of the scenarios.

It is important to note that the trials were not randomized so caution must be used when interpreting the results. Randomizing the trials was not practical as the eccentric always follows the concentric phase. Also, the majority of trials were run in sequential repetitions, as this is the way that the device will be used for strength training purposes. Assumptions that the model and the errors are normally and independently distributed with mean zero and constant but unknown variance were not violated as evident by the scattered residual vs. predicted plots and the normal distribution analyses of residuals produced during the statistical study (see Appendix I). In this application, statistical analysis is not crucial to the study and is only used as a reference in finding the strengths and limitations of the system.

ANOVA results indicate that first, the model is significant for each of the parameters \((p<0.0001 \text{ where } \alpha=0.05)\), and second, the simulation scenario, phase, and interaction between the two have a significant effect \((p<0.0001)\) for each parameter except for PMA displacement. Simulation scenario does not have a significant effect on PMA displacement \((p=0.0962)\). See Appendix I for a more complete listing of results. Tukey’s HSD tests and Student’s t tests (for the
two-level phase comparison) were used to evaluate the differences in the significant effects between simulation scenario, phase and the interaction. The fact that the interaction has a significant effect is not surprising due to the time dependent changes in simulation scenario parameters which affect the system differently during the concentric and eccentric phases. The focus here will be to identify significant differences between simulation scenarios and the phases, not necessarily the interaction subgroupings that are significantly different.

For the position parameter, the 0.175 rad/s scenario was significantly different from the other scenarios, as was 0.524 rad/s. The 0.175 rad/s scenario had the best position tracking performance (in terms of low RMSE) while 0.524 rad/s had the worst. The rest of the scenarios were positioned between these extremes in two groups (fatigued-rapid, 0.314 rad/s, erratic-pulse and fatigued-40, fatigued-20, erratic-sine wave). Erratic sine wave and fatigued-rapid were not significantly different. Position tracking performance during the eccentric phase was better than during the concentric phase.

For the velocity parameter, the 0.314 rad/s, fatigued-rapid, and 0.524 rad/s scenarios were significantly different than the erratic-pulse, erratic-sine wave, and fatigued-40. The 0.175 rad/s and fatigued-20 scenarios were positioned between these groups, but were not significantly different from the erratic-sine wave and fatigued-40. The first group, 0.314 rad/s, fatigued-rapid, and 0.524 rad/s had the best velocity tracking performance, while the second group listed had the worst. Better velocity tracking performance took place during the concentric phase.

For the torque parameter, the scenarios were separated into four non-overlapping significantly different groups. The 0.175 rad/s, fatigued-rapid, and fatigued-20 scenarios were grouped into the lowest RMSE category. Fatigued-40 and 0.314 rad/s were grouped into the next lowest. Erratic-sine wave and 0.524 rad/s were next lowest, while erratic-pulse had the highest RMSE. The target torque was followed more closely during the eccentric phase.
For the PMA displacement parameter, there were no statistically significant differences between simulation scenarios. Better displacement tracking performance took place during the concentric phase.

See the least squares means results (Figure 32) for an illustrative comparison between simulation scenario and phase for the four parameters studied.

*Figure 32. LS Means according to simulation scenario and task phase (found using JMP 9.0.0 (SAS Institute))*. 

124
7.7.2.2. Time Dependent Performance Results

The RMSE values for each of the parameters give only a limited view of the controller performance during each trial. Time dependent data is helpful in gaining a better understanding of the functioning of the system. Motor position (simulated joint angle), angular velocity, torque, and PMA displacement results plotted as a function of time for one repetition of each simulation scenario are shown in the following figures: Figures 33 to 36 for 0.175 rad/s, Figures 37 to 40 for 0.314 rad/s, Figures 41 to 44 for 0.524 rad/s, Figures 45 to 48 for fatigued-20 reps, Figures 49 to 52 for fatigued-40 reps, Figures 53 to 56 for fatigued-rapid, Figures 57 to 60 for erratic-sine wave, and Figures 61 to 64 for erratic-pulse. Calculation and comparison of coefficients of variation based on the standard deviation and mean RMSE of each subgroup (simulation scenario broken down by phase) shows that coefficients of variation for each subgroup vary between 0 and 15%, excluding a uniquely large spread of eccentric torque and velocity during the 0.524 rad/s scenario and larger variation (0 to 40%) in PMA displacement for several subgroups. This indicates that the figures displayed are a good representation of results over multiple repetitions. The standard deviations for position, velocity, and torque are less than 0.003 rad, 0.006 rad/s, and 0.13 Nm, respectively, excluding the large torque variation previously mentioned. The 0.524 rad/s results displayed are from a trial with torque results near the middle of the RMSE range. Trials chosen to be shown here have PMA displacement errors in the middle to upper end of the RMSE range. Each simulation scenario parameter set is from the same trial. Each of the position figures (Figures 33, 37, 41, 45, 49, 53, 57, and 61) display motor position output, desired motor position as defined in the desired motion profile (see Appendix G), and the position tracking error (defined as output minus desired). Each of the velocity figures (Figures 34, 38, 42, 46, 50, 54, 58, and 62) display motor velocity output, desired motor velocity as defined in the desired motion profile, and the velocity tracking error (defined as output minus desired). Each of the PMA displacement figures (Figures 36, 40, 44, 48, 52, 56, 60, and 64) display PMA displacement output, desired PMA displacement as defined in the desired motion profile, and the PMA
displacement error (defined as output minus desired). Each of the torque figures (Figures 35, 39, 43, 47, 51, 55, 59, and 63) display the motor torque output and the target torque, which is based on the torque-velocity-position relationship of the particular neuromuscular actuator being simulated as defined in the HQDS section. The torque was not a parameter that was tracked, as there are no predefined desired torque outputs.

Figure 33. Position results for 0.175 rad/s simulation scenario
Figure 34. Angular velocity results for 0.175 rad/s simulation scenario

Figure 35. Torque results for 0.175 rad/s simulation scenario
Figure 36. PMA displacement results for 0.175 rad/s simulation scenario

Figure 37. Position results for 0.314 rad/s simulation scenario
Figure 38. Angular velocity results for 0.314 rad/s simulation scenario

Figure 39. Torque results for 0.314 rad/s simulation scenario
Figure 40. PMA displacement results for 0.314 rad/s simulation scenario

Figure 41. Position results for 0.524 rad/s simulation scenario
Figure 42. Angular velocity results for 0.524 rad/s simulation scenario

Figure 43. Torque results for 0.524 rad/s simulation scenario
Figure 44. PMA displacement results for 0.524 rad/s simulation scenario

Figure 45. Position results for 0.314 rad/s, 20-repetitions fatigue simulation scenario
Figure 46. Angular velocity results for 0.314 rad/s, 20-repetitions fatigue simulation scenario

Figure 47. Torque results for 0.314 rad/s, 20-repetitions fatigue simulation scenario
Figure 48. PMA displacement results for 0.314 rad/s, 20-repetitions fatigue simulation scenario

Figure 49. Position results for 0.314 rad/s, 40-repetitions fatigue simulation scenario
Figure 50. Angular velocity results for 0.314 rad/s, 40-repetitions fatigue simulation scenario

Figure 51. Torque results for 0.314 rad/s, 40-repetitions fatigue simulation scenario
Figure 52. PMA displacement results for 0.314 rad/s, 40-repetitions fatigue simulation scenario

Figure 53. Position results for 0.314 rad/s, rapid fatigue simulation scenario
Figure 54. Angular velocity results for 0.314 rad/s, rapid fatigue simulation scenario

Figure 55. Torque results for 0.314 rad/s, rapid fatigue simulation scenario
Figure 56. PMA displacement results for 0.314 rad/s, rapid fatigue simulation scenario

Figure 57. Position results for 0.314 rad/s, erratic-sine wave simulation scenario
Figure 58. Angular velocity results for 0.314 rad/s, erratic-sine wave simulation scenario

Figure 59. Torque results for 0.314 rad/s, erratic-sine wave simulation scenario
Figure 60. PMA displacement results for 0.314 rad/s, erratic-sine wave simulation scenario

Figure 61. Position results for 0.314 rad/s, erratic-pulse simulation scenario
Figure 62. Angular velocity results for 0.314 rad/s, erratic-pulse simulation scenario

Figure 63. Torque results for 0.314 rad/s, erratic-pulse simulation scenario
Figure 64. PMA displacement results for 0.314 rad/s, erratic-pulse simulation scenario

The time dependent position data (Figures 33, 37, 41, 45, 49, 53, 57, and 61) shows that there is a small initial offset just as the task begins at 5 s. The initial PMA contraction is gradual to take up any slack in the force transfer cable. All slack is taken up at around 3 s in preparation for the 5 s concentric phase start. Resistance is introduced at around 3 s to prevent initial velocity overshoot as the task begins. The motor moves slightly around 3 s when it first feels resistance, making it necessary to produce torque in order to remain at the starting position. The time dependent position data also shows that there is higher error for regions of transition between concentric to isometric and isometric to eccentric. Errors during the isometric hold phase are most likely a result of resistance exceeding the isometric force capability of the simulated neuromuscular actuator as defined in the HQDS. The HQDS commands a non-zero velocity if the resistance felt is too high, analogous to a human operator lowering their leg if the resistance pulling it down exceeds their force output capability.

The time dependent velocity data (Figures 34, 38, 42, 46, 50, 54, 58, and 62) shows that the constant velocity regions are oscillatory in nature, but appear to oscillate around the desired
velocity except for spikes corresponding to changes in the torque-velocity-position relationship. In the 40 repetitions fatigued scenario, near the beginning of the task, the resistance drops dramatically to accommodate the drop in simulated neuromuscular actuator force output. The torque drop creates a situation where there is not enough resistance to constrain the simulated neuromuscular actuator to the correct velocity. The system then increases the resistance a bit too high, causing a velocity drop before correction takes place. In the erratic-pulse scenario, there are velocity spikes after the scaling parameter pulses causing a shift in the relationship between torque, velocity and position. The simulated neuromuscular actuator suddenly reacts differently to the same resistance level, and the system must adapt to accommodate for the changes. The motor commands are shown in the time dependent velocity data figures to illustrate the rate and magnitude of velocity command changes as the PMA resistance changes. The PMA control system appears to be able to mitigate the motor velocity changes as shown in the velocity output results.

The time dependent torque data (Figures 35, 39, 43, 47, 51, 55, 59, and 63) shows that the target torque is a complicated torque profile. It is important to remember that the control system does not function by tracking the target torque (torque corresponding to HQDS parameters, desired task velocity, and present position). Target torque is not something that will be known for an actual human. It is shown here to give a sense of how close the output resistance is to the resistance that corresponds perfectly to the desired velocity and position for the simulated neuromuscular actuator. It is clear that the output torque does not follow the target torque perfectly, particularly where there are rapid changes or local minima or maxima in the target torque curve.

The time dependent PMA displacement data (Figures 36, 40, 44, 48, 52, 56, 60, and 64) shows that there is a large amount of initial error as the PMA is contracting to take up slack in the force transfer cable. PMA displacement error seems to increase slightly at transition points when there is a change in PMA operation (relaxation to contraction). Error also seems to increase.
during the isometric hold (PMA displacement near \( x = 0 \)) if there is an elevated amount of torque or motor position activity.

For each simulation scenario, there is a unique set of torque resistance commands produced by the TMC. There is also a unique set of voltages commanded to the proportional pressure regulator (PPR) corresponding to pressure commands produced by the PMC. These command values, which in cooperation with the HQDS produce the results discussed here, are shown in Appendix F.

### 7.7.2.3. Accommodation of Different Neuromuscular Behavior Types

The torque outputs as seen in Figures 35, 39, 43, 47, 51, 55, 59, and 63, are different with each simulation scenario. Figures 65 and 66 show the torque outputs and target torques specifically for comparing different task velocities. With increasing task velocity, the concentric torque drops (10.2, 8.8, and 8.0 Nm for 0.175, 0.314, and 0.524 rad/s, respectively) while the eccentric torque rises (10.8, 12.0, and 13.4 Nm for 0.175, 0.314, and 0.524 rad/s, respectively). This follows the torque-velocity relationship seen in Figure 23. Torque drops with increased velocity during concentric exercise and increases with increased velocity during eccentric, at least for the range of low velocities tested here.

The torque output also changes according to the type of simulated neuromuscular behavior. Figures 67 and 69 illustrate differences between target torques for different simulation scenarios. Figures 68 and 70 illustrate differences in torque outputs for different simulation scenarios. For the fatigue scenarios (Figures 67 and 68), torque is reduced as the fatigue level increases and potential simulated quadriceps torque output decreases. For the rapid fatigue scenario, torque starts out near the 0.314 rad/s value during concentric (approximately 8.3 Nm), but then decreases below the 0.314 rad/s eccentric level (approximately 8 compared to 12 Nm). For the 20 and 40 repetitions scenarios, jumps in resistance can be seen as the system adjusts for the local minima and maxima in potential torque output and as it adjusts during the transition
points in the task (at approximately 5, 10 and 15 s). This behavior, at varying degrees, is seen in other simulation scenarios, and can be explained by the shape of the torque-velocity-position curve in these regions as well as the switch from concentric curve to eccentric curve taking place during phase transitions. This type of system reaction is also affected by the gains and bandwidth of the system. Choice of gain, $k$, and bandwidth, $\lambda_o$, was determined by position tracking results, not torque tracking because, as previously explained, the goal is to follow the desired motion profile.

During the erratic-sine wave simulated neuromuscular behavior trials (Figures 69 and 70), the force varied more drastically through the task as the HQDS scaling parameters varied according to a sine wave. During the erratic-pulse simulated neuromuscular behavior trials, there were four sudden changes in the HQDS scaling parameters causing an alteration of the torque-velocity curve with each change. These scenarios are intended to represent a neuromuscular actuator with reduced volitional control. In the concentric region we see accommodation of the simulated neuromuscular actuator with an increase in resistance at the onset of the step increase in torque potential of the erratic-pulse neuromuscular behavior type and in response to increased torque potential of the erratic-sine wave neuromuscular behavior type. In the eccentric region, we see accommodation with a decrease in resistance for the pulse and increased resistance for the sine wave. For the erratic-pulse simulation scenario, there is some oscillation in the eccentric resistance as the system adapts and pressure adjustments occur after the HQDS scaling parameters are returned to their original values for the final time.

After evaluating the position tracking results in relation to the torque output capabilities of each neuromuscular behavior type (each simulation scenario), a pattern seems to emerge. Using the 0.314 rad/s scenario as the baseline, scenarios with decreased torque output capability during the concentric phase (fatigued-20 repetitions, fatigued-40 repetitions, and 0.524 rad/s), where the resistance is lowered to accommodate the simulated neuromuscular actuator, have increased position error. For the scenario with increased torque output capability (0.175 rad/s),
position error decreases. The opposite is true during the eccentric phase. For scenarios with decreased torque output capability (0.175 rad/s, erratic-pulse, fatigued-rapid), position error decreases. For scenarios with increased torque output capability (erratic-sine wave, 0.524 rad/s), position error increases.

Figure 65. Target torques for different task speeds
Figure 66. Output torques for different task speeds

Figure 67. Target torques for fatigue simulation scenarios compared to 0.314 rad/s baseline
Figure 68. Output torques for fatigue simulation scenarios compared to 0.314 rad/s baseline

Figure 69. Target torques for erratic simulation scenarios compared to 0.314 rad/s baseline
7.7.2.4. Adaptation Results

Changes in neuromuscular actuator settings within the HQDS and other system changes such as velocity disturbances are designed to test the controller along with the adaptability of the controller. Parameters that are being adapted throughout the task, including $\hat{I}$, moment of inertia, and $\hat{c}$, the predicted human torque coefficient, are important in maintaining adequate tracking control throughout the course of the task. Without adaptation, position error for the 0.175 rad/s simulation scenario reached approximately 0.03 rad during the concentric phase and 0.05 rad during the eccentric phase. Position error for the 0.314 rad/s simulation scenario reached approximately 0.04 rad during the concentric phase and 0.10 rad during the eccentric phase. Position error for the 0.524 rad/s simulation scenario reached approximately 0.07 rad during the concentric phase and 0.13 rad during the eccentric phase. The adaptation process is reinitialized with an appropriate initial condition at the beginning of each phase. Figures 71 through 73 show

![Image](image1.png)

*Figure 70. Output torques for erratic simulation scenarios compared to 0.314 rad/s baseline*
the time course of the predicted human torque coefficient, \( \hat{\tau} \), whose starting value is 1, for different simulations scenarios. The predicted human torque coefficient adapts to adjust the contribution of the model prediction of human neuromuscular actuator torque when calculating the torque resistance command, \( T_R \), within the TMC. During the concentric phase, the value of \( \hat{\tau} \) drops, effectively reducing the contribution of the model prediction of human torque. This then changes the torque resistance command and in so doing changes the resistance produced by the PMA. The degree to which it drops depends on the neuromuscular behavior type being simulated. As seen in Figure 72, \( \hat{\tau} \) drops more substantially in the fatigue scenarios than in other simulation scenarios to automatically accommodate the reduced potential torque output of the simulated neuromuscular actuator, and drops most drastically for the 40-repetitions fatigue scenario, the weakest neuromuscular actuator type. As seen in Figure 73, \( \hat{\tau} \) for each of the erratic simulation scenarios seems to change more often than in other scenarios. During the eccentric phase, the value of \( \hat{\tau} \) increases. The exception to this is the rapid fatigue simulation scenario whose potential torque output decreases to the point that the system drastically reduces the contribution of the model prediction of human torque to produce the correct accommodating resistance.

![Figure 71. Time course of adapted predicted human torque coefficient (\( \hat{\tau} \)) for different task velocities](image)

**Figure 71. Time course of adapted predicted human torque coefficient (\( \hat{\tau} \)) for different task velocities**
7.7.2.5. Effect of Velocity Disturbance

Two types of velocity disturbances were added as an additional test of system robustness. The first was random noise with a 0.05 rad/s amplitude added to the HQDS calculated motor velocity command. The position RMSE values for the 0.314 rad/s desired task velocity are listed as follows: 0.0190 rad during the concentric phase, 0.0163 rad during the eccentric phase, and 0.0157 rad during the entire task. The second velocity disturbance was a 0.1 magnitude offset subtracted from the HQDS calculated motor velocity command between 5 and 10 s and
between 15 to 20 s. The position RMSE values for the 0.314 rad/s desired task velocity are listed as follows: 0.0095 rad during the concentric phase, 0.0221 rad during the eccentric phase, and 0.0282 rad during the entire task. The position results show that decent tracking remains intact despite the disturbances introduced.

7.8. Discussion

7.8.1. Controller Performance

Results from baseline motor performance testing indicate that closed loop haptic control improves tracking over simply commanding the desired velocity profile using closed loop motor control. The motor adjusts torque as necessary against the PMA provided resistance to move at the desired velocity, with the error as shown in Figure 27 and as reported in Table 9. The motor’s reaction to PMA resistance is not based on neuromuscular performance characteristics. It is simply a measure of motor closed loop control performance within the system setup (incorporation of filters and control command loops). With the implementation of the HQDS, the motor commands, and thus behavior in reaction to PMA resistance, are based on the torque-velocity-position performance envelope of the simulated neuromuscular actuator.

Evaluation of the velocity parameter reveals that there is a RMSE range of 0.0145 to 0.0440 rad/s with a CV RMSE range of 5.0 to 14.4% within the portion of the phase where desired velocity is constant (analogous to isokinetic lower leg movement). Oscillatory but stable velocity outputs are seen. This, however, does not prevent accurate position tracking. The system is able to sense and respond by adjusting PMA resistance in adequate time to maintain position accuracy. Position RMSE ranges from 0.0095 to 0.0305 rad. Accurate PMA displacement is necessary to ensure proper force transfer between the PMA and motor. The low PMA RMSE results (0.2576 to 0.6164 mm) indicate that there were no issues with force transfer during the course of the task. The torque data confirms this. There were no sudden drops in torque output.
that could not be attributed to effects of the HQDS. Torque outputs as compared to the theoretical target torque had a RMSE range of 0.5391 to 1.461 Nm.

Motor command changes are occurring at a fairly high frequency (10 Hz) within the HQDS, and there are potentially large changes in velocity with small changes in resistance. This can be seen in the command velocities shown in the time course velocity figures for each simulation scenario (see Figures 34, 38, 42, 46, 50, 54, 58, and 62). With a ±1 Nm offset in resistance for the task speed of 0.314 rad/s and default torque-velocity-position curve parameters, the velocity command change is -0.11/+0.13 rad/s during the concentric phase. During the eccentric phase, a ±1 Nm offset corresponds to a +0.22/-0.18 rad/s velocity command change. With the higher task speed of 0.524 rad/s, the ±1 Nm offset corresponds to velocity command changes of +0.14/-0.16 and +0.26/-0.22 rad/s. This illustrates the large velocity changes for differences in the resistances felt by the simulated neuromuscular actuator. Some of the velocity error can be attributed to these large jumps in velocity command magnitude, although the motor does not typically settle at these values as the controller is tracking motor position and velocity. PMA resistance changes, PMA damping ability, and the time constant of the motor all play a role in the rate and magnitude of the actual velocity changes seen in the results. The time constant is important in judging reaction of the motor to the velocity command. Although the motor cannot immediately achieve the commanded velocity (time constant of 0.18 s), there is a rapid reaction in response to command changes.

The oscillatory nature of the steady state motor readings as seen in the step response evaluation (Figure 26) may also contribute to the velocity error results, particularly when the PMA resistance is very close to the target torque and the velocity command is steady. Oscillatory velocity readings do not prevent accurate position tracking, but they do have an impact on the controller. Velocity error is a major component of the combined error signal, \( s \), defined in the TMC section. The combined error multiplied by \( k \) is a significant term in the control law (Eq. (34)), particularly with a gain setting of \( k=35 \). The combined error also plays a crucial role in the
adaptation process. Oscillatory velocity measurements, therefore, impact the torque resistance command, $T_R$, and in turn, the pressure command and the associated voltage command to the PPR (see Appendix F for control command signals).

Chattering, or high frequency oscillations in the control signal, is a common issue with the implementation of sliding control. It is usually undesirable because it results in high control activity and may excite high frequency dynamics (215). Chattering occurs in the control signal as the error signal rapidly switches back and forth across the sliding surface. This phenomenon was not present in this study.

As previously noted, there may be a link between level of resistance, phase of the task, and position tracking accuracy. With reduced task speeds and certain simulated neuromuscular behavior types, there is a higher torque resistance during the concentric phase and/or a lower torque resistance during the eccentric phase, relative to the baseline level torque required for a 0.314 rad/s task speed. These situations result in lower position RMSE. It is difficult to say why exactly this is happening. One possible explanation for this observation may be that the different PMA operation modes operate most accurately at different rates of pressure change. An increase in concentric resistance requires an increased rate of pressure change to achieve the range of torques needed throughout the phase of the task. A decrease in eccentric resistance requires a reduced rate of pressure change. During the concentric phase, the operation mode is relaxation where pressure is being bled to the environment. During the eccentric phase, the operation mode is contraction where pressure is increasing inside the PMA. Between the two operation modes, there are potential differences in force production, damping, and response time. Force production and level of damping is related to $B_{PMA}$, the coefficient of $\dot{x}$, within the three-element phenomenological model. It is characterized differently for contraction and relaxation.

Characterization studies outlined in (19), show that $B_{PMA}$ for contraction is averaged at 2.9 across the operating pressure range, although there is a higher amount of variability in the coefficient at lower pressures (varies within a 1Ns/mm range between 150 and 350 kPa). $B_{PMA}$ for relaxation is
defined as a piecewise function of pressure, with a value of 1.57 below 372 kPa and values ranging from 1.57 to 2.17 above 372 kPa. The eccentric phase theoretically has a larger time constant than relaxation mode due to the fact that you have to add pressure to a pressurized system instead of bleed to the lower atmospheric pressure. This idea is in agreement with the theoretically calculated time constant of $B_{PMA}/K_{PMA}$ according to the differences in $B_{PMA}$ outlined above. Past studies have shown that Festo MAS20-250N time constants vary from 0.05 to 0.13 s, depending on step pressure input and operation mode (contraction or relaxation). These results are similar to the theoretically calculated time constant of $B_{PMA}/K_{PMA}$ used previously in (16), although calculated with different PMA parameterization values. Due to the nature of PMA operation, relaxation mode is better able to handle increased rates of pressure change as compared to contraction mode. This may be a contributing factor in how rapidly the system is able to adjust resistance for better position tracking.

There is another possible explanation for decreased position RMSE in situations with a higher torque resistance during the concentric phase and a lower torque resistance during the eccentric phase. During the counterclockwise concentric phase, the motor must move in the opposite direction of the PMA provided resistance, and thus must provide torque to move opposite that of the PMA force vector. During the clockwise eccentric phase, the motor is moving in the same direction of the PMA provided resistance. Here the motor is providing braking torque against the PMA produced resistance. This difference in motor operation may be a contributing factor in performance differences seen at different levels of PMA provided resistance.

Improved position tracking at lower velocities could be a result of smaller accelerations and decelerations at lower speeds (0.3 rad/s$^2$ at 0.175 rad/s and 0.314 rad/s vs. 0.5 rad/s$^2$ at 0.524 rad/s). A higher acceleration was necessary at higher velocities to arrive at the correct speed in adequate time.

Velocity tracking was not shown to improve with decreasing velocity of the task (according to CV RMSE, a normalized version of velocity RMSE). There are several possible
explanations for this. First, only the isokinetic region was considered in the velocity CV RMSE comparisons, thus excluding the regions of acceleration and deceleration, unlike the position tracking evaluation. Second, the normalization process could be skewing the results. The normalization may cloud the comparison of velocity tracking, making the 0.524 rad/s scenario seem better or the 0.175 rad/s worse, due to the magnitude of the mean by which RMSE is divided. The degree to which velocity tracking error variance changes with velocity magnitude (level of heteroskedasticity) is not exactly known. Not enough speeds were studies to evaluate this effect, making it difficult to evaluate the normalization process. Third, there is a difference in duration of isokinetic movement at different task speeds. The length of time the motor is at a near constant speed for the slower speed, 0175 rad/s, is approximately 8 s as compared to 2 s for the 0.524 rad/s task speed. With a longer duration of isokinetic movement, there is more of an opportunity for increased error variance due to oscillatory velocity readings.

It is important to note that position may be a more reliable parameter on which to base tracking performance of the simulated neuromuscular actuator within the haptic PMA control system. The entire phase can be included in the evaluation, no normalization is necessary, there are no known signal processing issues, the calibration process was customized for the system (see Appendix D), and there were no large error spikes to potentially skew the RMSE value. Calculation of RMSE is particularly susceptible to sporadic spikes in error because the error is squared as part of its calculation, giving more weight to these points.

Increased position RMSE occurs near transition points, particularly at the end of the concentric and beginning of eccentric where deceleration and acceleration are occurring. Large time constants and system time delays potentially have more of an impact on the accommodation of the more rapid velocity changes occurring with higher accelerations. Faster changes in the desired motion profile require more rapid resistance adjustment. Motor behavior itself could also be causing position and velocity tracking error at transition points. Evaluation of the baseline
motor performance (see Figure 27) shows large overshoots in velocity and position despite the fact that the velocity command remains equal to the desired velocity.

Delay in system response causes a delay in the resistance adjustment to the simulated neuromuscular actuator, thus increasing the potential for position and velocity error. This may be one of the reasons why the eccentric phase is more sensitive to velocity changes than the concentric phase. As previously discussed, the eccentric phase has a larger time constant than the concentric phase. The portion of the task where acceleration/deceleration is 0 (as seen in Table 10 under the headings Concentric $\alpha=0$ and Eccentric $\alpha=0$) shows greatly reduced velocity error, particularly eccentric velocity error.

Delay in system response is evident in the initial PMA displacement ramp, where the PMA is contracting to get to the proper initial length against no load. There is a distinct difference in the PMA displacement results between this initial contraction and the contraction taking place during the eccentric phase. There is a greater amount of error associated with the initial contraction as compared to the incremental pressure changes taking place against a load during the eccentric phase. Operating the PMA in this manner (small incremental pressure changes), allows for greater accuracy and faster control. It is also important to note that there is greater PMA displacement error associated with the eccentric phase (mainly PMA contraction) as compared to the concentric phase (mainly PMA relaxation). The slower initial pressurization of the PMA will not be an issue for continuous repetitions as the PMA pressure will not be completely bled after each repetition is complete. Final PMA displacement of each repetition can be kept near the necessary starting displacement.

As seen in Figure 30, torque error is particularly high for the erratic-pulse scenario. In this scenario, a sudden disturbance is added into the system via changes in HQDS scaling parameters at four different instances during the task. The potential torque output capability at the desired velocity and position (target torque) also changes suddenly. Because torque RMSE is
based on the difference between experimental torque output and the purely theoretical target torque, larger errors occur in this scenario in comparison to other scenarios.

### 7.8.2. Velocity Tracking Goals

Isokinetic dynamometers serve as one tool which clinicians and researchers use for strength training and performance measurement of their patients. The objective of this work is not to create an isokinetic dynamometer capable of performance measurement. Rather, the ultimate goal is to control a PMA for use in a low-speed isokinetic strength training device which can accommodate different types of neuromuscular behaviors. The level of isokinetic dynamometer control accuracy, however, provides a benchmark with which to compare system performance. It was noted in (183) that researchers using isokinetic dynamometers as a device for performance measurement accept velocity that varies from 2 to 10% of the desired speed, due to the presence of velocity overshoot and oscillation. It is important to note that for clinical purposes, where results are not used to quantify human performance, higher errors may be acceptable. Percent errors for the trials run in this study range from 0.1 to 3.5%. Percent deviations range from 3.0 to 12.6%, although all except the erratic-pulse eccentric scenario are under 9.1%. These ranges show that the PMA haptic control system has potential for use in a device designed for human implementation.

### 7.8.3. Variability

Like most systems, the PMA control system operating with the simulated neuromuscular actuator has an amount of variability between repetitions for each simulation scenario. The uniquely large variability seen in the torque RMSE data for the 0.524 rad/s scenario (Figure 30) can be explained by varying degrees of overshoot above the peak target torque during the eccentric phase. The large amount of variability in PMA displacement between trials (see Figure 31) may be due to changes in behavior during different operation modes between phases.
(contraction during concentric phase, relaxation during eccentric phase). The variability seen in the PMA displacement results may also be a result of the strong dependence between the relative starting points of the PMA, LVDT, and motor. Slight changes in the starting positions could affect the accuracy of PMA displacement. The variability may also be due to hysteresis, as the contraction during eccentric phase is occurring after a contraction and relaxation cycle and many of the trials were run right after one another. As reported in Festo technical literature (24), hysteresis is specified to be $\leq 2.5\%$ of nominal length or 6.25 mm. This amount of hysteresis has the potential to affect repeatability of different system parameter outputs. Running the PMA sequentially potentially affects its behavior. Heat may build up and mechanical changes may occur affecting how the parts of the pneumatic muscle operate together, leading to changes in PMA performance. This is something that the control system must be able to handle. With human operation, repetitions will be a necessary component in the PMA control system operation for strength training or rehabilitation. There was no focus on studying hysteresis specifically, but the fact that repetitions were controlled with a high level of accuracy and relatively limited variability shows that hysteresis is not negatively impacting system performance. Variability remains small overall because RMSE values remain small for the different scenarios.

There are differences in variability among different simulations, different phases and different trials because there are interdependent changes occurring in the system throughout the course of the task. Motor behavior is changing according to how the neuromuscular actuator is modeled for the simulation being studied. The PMA displacement is changing, thus changing its potential force output and the relationship between pressure and force. The motor velocity direction is changing (counter clockwise for concentric phase, clockwise for eccentric phase), thus affecting the relationship between motor torque and PMA force production. The magnitude of motor velocity is in actuality changing despite the isokinetic nature of the task due to the acceleration and deceleration.
7.8.4. Limitations of the PMA Dynamic Test Station and Haptic Control System

The main limitations of the PMA system are its relatively slow response and its limited capability to maintain tracking accuracy with large simulated neuromuscular actuator accelerations, specifically clockwise accelerations above approximately 1 rad/s². However, this limitation can also be seen as an asset because it puts the neuromuscular actuator in control of the system. The PMA system is not able to handle large clockwise accelerations because of the single PMA configuration and the delay in system response. The response time of the system is dependent on the time-varying time constant of the PMA (0.05 to 0.13 s, depending on pressure change and operation mode), in combination with (but not in addition to) PPR response time (up to 0.22 s) and sensor data filter-related delay. Delay in system response due to large time constants and filter time delays could possibly allow slack to form during high clockwise accelerations, thus creating an interruption in force transfer. Theoretically, the neuromuscular actuator can stop the task at any point by dropping the lower leg clockwise, back towards the 1.57 rad (90°) starting position. The PMA resistance would never prevent this type of action (force production is in the opposite direction), and if dropped with a high enough acceleration, the leg would not feel any PMA resistance pulling the leg. This may be more ideal than working with stacked weights or an ankle weight where there is no way to prevent pulling of the leg.

This work focuses on the eventual use of PMA produced resistance for strength training, where there is a potential need for high torque outputs at slow speeds, and rehabilitation, where there is a potential need for low torque outputs at slow speeds. There is no intention to control PMAs for use within an isokinetic dynamometer capable of measuring human performance at high velocities. Low-speed training has been shown to produce greater strength gains than fast velocity training over a wide range of velocities (203). Also, with lower speeds, isokinetic movement is possible through a larger range of motion (less of the range of motion is needed for acceleration and deceleration). Velocities at the high end of the slow isokinetic exercise range (1.05 rad/s or 60°/s) may be manageable for the system if an appropriate acceleration is defined in
the motion profile. The issue is that with velocities above this range, accelerations exceeding that of the system capability are necessary to get to the desired task velocity before the task is completed. Otherwise there would be acceleration immediately followed by deceleration, without reaching the desired task speed. The adaptive nonlinear control is able to handle a certain amount of time delay and uncertainty in the system. The PMA system may work more effectively at higher velocities if time-varying time constants and time delays are specifically incorporated into the control law.

Another limitation of the system is the limited torque output of the motor and the limited potential force output of the PMA. The limited motor torque output of 22 Nm is about 10-20% of maximum human quadriceps torque (17). The potential force output of the PMA itself is also limited. The maximum potential output is 1200 N or 32 Nm of torque, and this decreases with increased displacement. When incorporating this type of PMA into an exercise device to be used by humans, multiple PMAs can be used simultaneously to increase the force output. The moment arm can also be increased somewhat, depending on what is practical for the mechanical design. The work done here is at a scaled level so that the limited motor torque and PMA force output does not restrict progress on developing effective PMA control.

7.8.5. Limitations of the Human Quadriceps Dynamic Simulator

The simulated neuromuscular actuator developed in this work is comprised of the HQDS and motor, and it functions by sensing and perceiving a stimulus (resistance in this case) and responding with a velocity output according to a time-varying torque-velocity-position relationship. Time-varying torque-velocity-position curves used within the HQDS are estimations of a neuromuscular actuator’s performance and by no means perfectly define human reaction, ability or performance. However, the creation and use of this model provides a safe, consistent, and reliable method in which to test the haptic control system and the capabilities of the PMA system. The model is fairly straightforward to implement and does not require significant
computing time. The HQDS responds to stimuli and changes with time, which was necessary to test the controller. In addition, considerations are in place to test the limits of the controller; namely different time-varying scaling parameter values within the HQDS and TMC and velocity disturbances to alter the HQDS calculated velocity command. There is also an inherent amount of error related to the closed loop motor control as previously discussed, adding a level of disturbance to the system.

Human response time, according to Phillips (188), is defined as the minimum time to generate an output response for an input stimulus. It is a combination of reaction time and movement time. Reaction time is dependent on the type of stimulus (visual, auditory or tactile), whether the operator is expecting the signal, and information content (choice of alternative actions). If we assume that knee extension is a simple task, with little or no information content, the reaction time can be estimated as 0.2 s. Movement time, defined as the time required to physically complete the response, varies significantly based on the task. Optimal control based human tracking models have been designed to incorporate cognition, information processing and response into modeling human performance (216). Within this type of model, human reaction time, defined as the time to process information from the receptor, appropriately transform the information, and communicate with the effector, is typically set to 0.15 s. The time constant of neuromuscular lag, defined as the time to get the muscles in motion, is typically set between 0.1 and 0.2 s.

The HQDS and motor respond to the system stimulus (level of resistance) every 0.1 s. Although the motor does not immediately achieve the commanded velocity (time constant of 0.18 s), this rate of response is as fast or faster than a typical human neuromuscular response. Like a true human operator, the simulated neuromuscular actuator bases its response on delayed and sometimes noisy observations of the output. The high frequency of the simulated neuromuscular actuator response, along with the large changes in response magnitude (velocity command) with small changes in stimulus (resistance), is representative of aggressive human neuromuscular
reaction in response to changes in a stimulus. Testing the system with an aggressive simulated response that is more difficult to control allows for more thorough testing of the haptic control system.
8. IMPLICATIONS OF RESEARCH AND FUTURE WORK

8.1. Potential Applications

PMAs have a high force to weight ratio and are a safe alternative to other types of actuators. The lightweight, clean, and compact PMA driven by pressurized air does not require gravity to produce resistance, and thus has potential for use as a microgravity exercise device used to counteract muscle atrophy and bone loss during prolonged spaceflight. There is also potential for development of a portable strength training or rehabilitation device. This potential exists because of PMA’s high force to weight ratio, relatively compact package, and low power needs (necessary only to run control functionality if air tank in use). Isokinetic strength training using PMA produced resistance could potentially develop as a smaller, less expensive alternative to isokinetic dynamometers (although with greatly reduced capability). Isokinetic dynamometers are expensive and large, requiring up to 30 to 60 square feet in floor space. They weigh up to 1300 lb, and require special electrical hookups (208/230VAC, 50-60Hz, 20A isolated dedicated service) (217).

8.2. Suggestions for Human Implementation

Several suggestions related to human implementation of controlled PMA produced resistance are listed here. If the PMA control system described here is implemented for a knee extension with a true human operator, settings for the predicted estimate of human quadriceps produced torque can be customized for each individual user. Peak torque and the angle at which peak torque occurs can be found using an isokinetic dynamometer. Finding the angle at which peak torque occurs allows one to create a new relationship between torque and joint angle.
(position) and velocity and joint angle. It is important to find the angle at which peak torque occurs at the desired task velocity. With increasing velocity, the peak torque angle will decrease (in terms of knee flexion angle) (208). Preliminary subject testing will also reveal the appropriate peak velocity for the individual, and an appropriate range of resistances based on human neuromuscular ability and desired speed. Finding the range of torques in which the operator is able to produce at the desired effort level (for cases of submaximal activation) will help in selecting human torque model scaling parameters ($a_{T_c}$, $a_{T_v}$, and $a_v$). This will allow for the convergence of the adaptive parameters within the TMC, theoretically leading to better control. Incorporation of visual or auditory feedback may be necessary to help the operator maintain a desired submaximal activation level or submaximal range. This type of feedback has been shown to affect isokinetic performance (180) (208). The start and stop angle can be set in the desired motion profile in addition to the desired task velocity and the desired accelerations and decelerations. In addition to predicted human torque model parameters (model scaling parameters, peak torque, peak velocity, and peak torque angle), control parameters such as gain, bandwidth, and adaptation gains can be adjusted to optimize the control operation for each human operator. Optimization of the parameters via a learning algorithm may help with this process.

Human implementation of controlled PMA produced resistance will require a new test device, complete with new torque, position, and velocity sensors. One design idea for a human test device can be seen in Appendix K. This configuration would be familiar to those who have used exercise equipment. There is no need to affix a complicated device to the leg, making the device more comfortable and better suited for a large range of human operators. There is no need to create a brace structure that can be used by people with varying anthropometric characteristics. Consideration should be taken to control body positioning during operation. The center of rotation of the lever arm should be aligned as close as possible to the axis of rotation of the knee. Angle of the hip and ankle joints should be controlled because this could potentially affect performance (208). If comparing different subjects, it is important to understand how device
characteristics affect each of the operators. When a human is performing a knee extension exercise, one must also be aware of the need for gravity compensation to account for gravitational effects on the torque output. There is a discrepancy in human torque production between extending the knee (working against gravity) and flexing the knee (working with gravity) so one must add or subtract the moment due to gravity according to phase and position.

Human implementation of controlled PMA produced resistance will require an increased level of PMA produced torque than what has been evaluated here. Adding additional PMAs in parallel will increase torque production. Using the same type of Festo fluidic muscle is beneficial because it has been characterized, but larger PMAs are available with higher potential force outputs. Increasing spool size will also increase the PMA produced torque by increasing the moment arm of the line of action of the PMA produced force.

8.3. Potential Control Improvement

There are several potential areas of control improvement. First, the frequency at which commands are sent to the PPR could be improved with an improved data acquisition card. The system is currently set up to send analog outputs at the maximum rate the NI 6025E data acquisition board allows. A more advanced data acquisition card would allow faster signal transfer, potentially increasing control system responsiveness. Second, the delay and lag in the system could be directly incorporated into the control law. This task is not trivial and would complicate the control law, thus lengthening computation time. As in any system, a balance has to be achieved between performance and necessary computing power. Third, further studies could be done to design a filter that would smooth high frequency PPR voltage commands without negatively impacting tracking accuracy. Preliminary studies revealed a negative impact on tracking performance with additional filtering of this signal. Fourth, incorporation of an iterative learning controller could potentially improve performance on a patient by patient basis, and at the same time make the system easier to use. The first repetition would be based on customized
torque-velocity-position relationships built from preliminary subject testing. The next repetition would use data from the first to improve controller performance. The parameters would be optimized over several iterations to improve control performance. The predicted human torque model component of the controller can easily be updated while keeping the control structure and functionality intact. One final area for potential improvement is the LabVIEW programming used to run the system. It could be made more user-friendly for operators unfamiliar with its programming functionality. If human implementation is pursued, software features should be added to improve performance and safety. See Appendix E for more information on LabVIEW programming.

8.4. Expanding System Functionality

There is also potential for expanded functionality of the system by changing the configuration of the PMA. The agonist-antagonist configuration of two PMAs around a pulley allows for torque production in either direction. A new control method would have to be applied to control the pressure differential between the two muscles. This would introduce the possibility of PMA produced resistance for knee extension and knee flexion strength training within the same device setup. With knee flexion the focus is on strengthening the hamstrings instead of the quadriceps. This would replace the eccentric phase of the task. Instead of the lower leg being pulled down by the resistance, the hamstrings would have to work at pulling the leg down to the original starting position. The new configuration would also introduce the possibility of resistance and assistance within the same device setup. There is also potential for spasm mitigation during extension because the direction of torque could potentially be changed at any point during the task. However, response time of the system may be a limiting factor for spasm mitigation.

With an agonist-antagonist pair of PMAs operating around a pulley, the need for highly accurate PMA displacement tracking will be reduced because there is no need to eliminate slack in a force transfer cable. The downside to operating an agonist-antagonist pair is that there is no
safety mechanism built into the way the PMA is transferring force. The neuromuscular actuator would no longer be a necessary element in the production of PMA resistance, and because of this, there is no guarantee the human operator will be able to stop applied resistance at any time they wish.

The PMA control system developed here was tested using a simulated neuromuscular actuator, whose model was based on simulated human quadriceps functionality. The control system could theoretically be applied to other joints, such as the elbow, with a change in the torque-velocity-position relationship. Potential torque outputs and velocities would be different, as well as peak torque angle. The relationship between concentric and eccentric torque-velocity production would also be different.
9. CONCLUSIONS

PMA, a clean and compact actuator with a high power to weight ratio provides “soft” actuation making it ideal for applications involving human interaction. A PMA control system was developed to produce controlled resistance against an opposing actuator in a dynamic system. The opposing actuator was designed to act as a simulated neuromuscular actuator, based on torque-velocity-position characteristics of human quadriceps. The PMA with a single line of force transfer was operated using a complete relaxation and contraction cycle, taking advantage of faster response with relatively small incremental pressure changes. The PMA was operated independently, meaning there was no PMA agonist/antagonist relationship and no spring or other device necessary for operation.

An effective combination of PMA displacement control (displacement tracking RMSE: 0.26-0.62 mm) and force/torque control (target torque tracking RMSE: 0.54-1.5 Nm) resulted in a successful demonstration of simulated isokinetic concentric and eccentric knee extension strength training. A nonlinear adaptive control method was incorporated into the system to create haptic (force feedback) control, which by adjusting PMA resistance, helps the simulated neuromuscular actuator maintain proper task speed. PMA control is based on an inverse three-element phenomenological model with pressure dependent parameters and nonlinear adaptive control, which operates on a combined position and velocity tracking error.

The simulated neuromuscular actuator was developed to test the controller prior to human implementation. A motor was used as the source of the torque, analogous to the torque produced by the quadriceps muscles in a knee extension exercise. The Human Quadriceps Dynamic Simulator (HQDS) controlled motor commands based on the torque-velocity-position relationship
of skeletal muscle and utilized system feedback including joint angle and PMA produced resistance force.

Simulated fatigued and erratic operational behavior was tested using the simulated neuromuscular actuator. Velocity disturbances were added to further test robustness of the system. Results show motor shaft position (simulated joint angle) control accuracies of 0.01 to 0.03 rad (position tracking RMSE), depending on the simulation scenario. Velocity control accuracies ranged from 0.01 to 0.04 rad/s (velocity tracking RMSE during isokinetic range of motion) with percent error ranging from 0.1 to 3.5%. There is evidence to suggest a portion of the velocity error variance is due to motor control behavior or signal processing and the rapid, large change in command velocity with small changes in resistance, which may or may not manifest in human operational testing. The control system interacting with the simulated neuromuscular actuator has been shown to work well at low isokinetic target velocities.

An important part of this work is the continuation of preceding work done at Wright State University, including model development and characterization. This work is an important step towards human implementation of controlled PMA produced resistance for isokinetic strength training and rehabilitation at Wright State University. The PMA haptic control system developed here, shown to accommodate simulated erratic operational behavior and simulated physiological fatigue, has a great amount of potential.
APPENDIX A

ADDITIONAL MOTOR CONTROL INFORMATION

The motor used within the Dynamic Test Station, PMA45N-00100-00, was manufactured by Pacific Scientific (now part of Danaher Motion). The motor driver, model PC 833-001-T, also manufactured by Pacific Scientific, controls motor behavior and has many different configuration options. Detailed information about these two pieces of equipment can be found in (218). Important information about motor control and configuration relative to this work and operation of the Dynamic Test Station is listed below.

The motor control modes used in this research are velocity mode-analog command and position mode-predefined moves. Others include position mode- step and direction, position mode- electronic gearing, velocity mode- frequency command, velocity mode- serial command, torque mode- analog command, and torque mode- frequency command. Torque mode- analog command has been used for past work with the DTS.

Position mode-predefined moves allows the user to define up to eight distinct moves. The available move options include holding a position, running at a certain specified velocity, traveling a certain predefined distance, traveling to a predefined position relative to a home position, and setting a home position.

Velocity mode- analog command allows the user commanded line voltage to set the motor shaft velocity. The output shaft velocity per input volt is set by the command gain (CmdGain) parameter in kRPM/V as shown in Eq. (A1). CmdGain acts to scale the analog voltage input, so its value is chosen based on the range of the desired velocity output. The analog voltage input command signal was limited to ±10V by the data acquisition card.
Motor Velocity \[ k_{RPM} \] = Analog In \[ V \] * CmdGain \[ k_{RPM}/V \]  
(A1)

Torque mode- analog command allows the user commanded line voltage to set the motor’s current. Controlling current is equivalent to controlling motor shaft torque. The output current amplitude in amps per input volt is set by the CmdGain parameter directly in Amp/V as shown in Eq. (A2). CmdGain acts in the same manner to scale the analog voltage input.

MotorCurrent \[ Amp \] = Analog In \[ V \] * CmdGain \[ Amp/V \]  
(A2)

Digital operation was made possible by adding electronics to the DTS that amplify 5V digital signals coming from the data acquisition card to 24V in order to be recognized by the driver. These digital signals can now be sent to the driver through six user mappable digital lines. The user can select what each line does using the motor driver configuration software. Functions include clearing analog voltage from a line, inhibiting clockwise or counterclockwise motion, switching gains back and forth between two values to alter motor speed or torque (depending on the mode), resetting drive faults, switching the active mode of operation, triggering a move in Position mode- predefined moves, and determining which move is active in Position mode- predefined moves. Each of the digital inputs can be defined as either active high or active low. If the mappable input function is configured as active high and the input is high, the mappable input function will be asserted. These digital lines can used even while operating in an analog mode.

Two analog outputs can be sent from the motor at a time. There are thirty outputs available to choose from. Gains were chosen to correctly scale the output signals between -5 and +5V. Potentially useful outputs include measured velocity, velocity error, position, position error, measured torque current, and measured non-torque current. The analog output signal is captured using LabVIEW.

Additional outputs from the motor can be used to monitor motor shaft position. The resolver feedback electronics are able to emulate an optical encoder using quadrature encoded TTL incremental position signals. One signal, from channel A, pulses as the motor shaft changes.
Each pulse represents an increment of angle change based on the EncOut parameter in the motor driver configuration file. The other signal, from channel B, also pulses, but 90° out of phase relative to channel A. If A leads B, the shaft is rotating in a clockwise direction. If B leads A, the shaft is rotating in a counterclockwise direction. LabVIEW is used to monitor both the number of pulses and the relative phase of A and B so that position is tracked and direction of rotation can be determined. Please note that the emulated encoder functionality was not used for final DTS testing. A rotational potentiometer was used. The resolver electronics are, however, used by the motor for motor control and analog output.

Basics of the motor control process are summarized in the schematic shown as Figure A1. More information on motor control structure and functionality can be found in (218). The left side of the control schematic is analogous to the somato-sensory system of a human, in that position and velocity information are sensed and fed back to the system. Decisions about the next action of the system are based on this feedback as well as the desired or command information. The right side of the control schematic is analogous to the musculoskeletal system of a human in that this is where the action of the system is produced and interaction between the actuator and the environment occurs. The feedforward (FFI) and feedback (FBI) interfaces are marked. Equations used within the controller are also marked and correspond to Eq. (A1) to (A5) listed below.

\[
\text{PosErr} = \text{PosCmd} - \text{PosFB} \quad \quad (A1)
\]

\[
\text{NetVelCmd} = (\text{PosErr} \times 2\pi \times Kpp) + \left( \frac{\text{dPosCmd}}{\text{dt}} \times \frac{Kvff}{100} \right) \quad \quad (A2)
\]

\[
\text{VelErr} = \text{NetVelCmd} - \text{VelFB} \quad \quad (A3)
\]

\[
\text{CurrentCmd} = Kvp \times \text{VelErr} \quad \quad (A4)
\]
\[
\omega = \text{VelFB} = \frac{K_{\text{eff}}}{J_{\text{TOT}}} \star \int \text{CurrentFB} \star dt = \left( \frac{\text{TorqueCmd}}{\text{CurrentCmd}} \right) \star \int \text{CurrentFB} \star dt
\]  

(A5)

Within these equations, \( Pos \) stands for motor shaft position, \( Cmd \) for command, \( FB \) for feedback, and \( Err \) for error. \( J_{\text{TOT}} \) is the total inertia of the system, and \( K_{\text{eff}} \) is the current (motor torque) command gain.

**SOMATO-SENSORY SYSTEM**:  
**MUSCULO-SKELETAL SYSTEM**

![Motor Control Schematic](image)

**Figure A1. Motor Control Schematic**

There are several loop gains that can be tuned to improve motor performance if necessary. Tuning is most effectively performed using the online control capability of the driver, where commands and gains can be changed in real time while the motor is operating. Within the velocity loop there is \( K_{vp} \), velocity loop proportional gain. This is based on the velocity loop bandwidth and the inertia ratio. If either of these is increased, \( K_{vp} \) may need to be increased to maintain adequate performance. ARF0 and ARF1 are anti-resonance low pass filters that filter out undesired high frequency velocity loop noise. ARF0 and ARF1 should be approximately 2 and
10 times the bandwidth, respectively. Kvi, integral gain, is used to reduce steady state error from the velocity loop. If this value is increased, stiffness and rejection of disturbances increases. If decreased, increased sponginess may occur. Within the position loop, there is Kpp, position loop proportional gain. Position error will be proportional to commanded speed by this value. If Kpp set too large relative to the bandwidth, velocity overshoot or oscillation may occur. Kvff, velocity feedforward gain, only helps in applications that require position synchronization (electronic gearing). The velocity loop bandwidth, \( f_{vc} \), ranges from 40 to 120 Hz ("spongy" to "stiff" performance). The recommended value for the motor is 75 Hz. Settings within the current loop (musculoskeletal section) include KIP (proportional current loop gain) and KII (current loop integral gain). These are factory set and should not be changed. Control gain settings used for this research are shown in Appendix C.

To more closely simulate quadriceps loading in a normal 1-g environment, the attachment of a scaled lower leg and foot to the motor shaft was investigated. The attachment created issues related to motor control. Stiffness of the motor mechanical system must be kept high in order to eliminate or reduce motor resonance issues and maintain high accuracy in controlling position. When attaching something to a motor shaft, it is difficult to make the connection perfectly stiff. The attachment of a scaled lower leg and foot also created a large inertia ratio (defined as the ratio of load inertia to motor inertia). Lower inertia ratios improve response times and reduce mechanical resonance (219). An inertia ratio greater than 10 can produce oscillations and extended settling times. The motor’s moment of inertia is equal to 0.00084 kg m\(^2\). The moment of inertia of parts attached to the motor shaft (spool around which the cable wraps and a blocking plate which helps to control cable wrapping) is 0.0002 kg m\(^2\). The ratio is therefore 0.238, well under the ratio required for acceptable motor performance and control. Attaching a scaled leg and foot increased the moment of inertia to 0.0710 kg m\(^2\) and the inertia ratio to 84.5, beyond the acceptable range for adequate motor control. Unstable behavior was seen while in position mode, despite tuning adjustments.
Position mode was utilized for the open loop control investigation. Velocity mode with operational switching into and out of position mode was utilized for the closed loop studies.

Position mode was required for positioning of the motor prior to the start of the task. It was also used to create a simulated hard stop at the 0.087 rad (5°) location so that no further extension was possible past this point (end of simulated neuromuscular actuator’s range of motion).
APPENDIX B

WIRING OF MAJOR COMPONENTS WITHIN DYNAMIC TEST STATION

Table B1. Wiring Between Motor Driver and Data Acquisition Card

<table>
<thead>
<tr>
<th>PC833 Motor Driver Terminal</th>
<th>NI PCI-6025E Terminal</th>
</tr>
</thead>
<tbody>
<tr>
<td>J2-1 Analog Command Input (+)</td>
<td>21 AO 1 Analog Out 1</td>
</tr>
<tr>
<td>J2-2 Analog Command Input (-)</td>
<td>23 AO GND Analog Out Ground</td>
</tr>
<tr>
<td>J2-3 I/O RTN</td>
<td>23 AO GND Analog Out Ground</td>
</tr>
<tr>
<td>J2-4 Analog Output 1 (DacMonitor 1)</td>
<td>11 AI 4 Analog In 4</td>
</tr>
<tr>
<td>J2-5 Analog Output 2 (DacMonitor 2)</td>
<td>13 AI 5 Analog In 5</td>
</tr>
<tr>
<td>J2-8 Encoder Output Channel A</td>
<td>47 PFI 8/CTR 0 SRC Count 0 Source</td>
</tr>
<tr>
<td>J2-10 Encoder Output Channel B</td>
<td>30 P0.6 Count Direction</td>
</tr>
<tr>
<td>J2-31 Input 1 (Fault Reset)</td>
<td>28 P0.5 D I/O</td>
</tr>
<tr>
<td>J2-32 Input 2 (Cwlnh)</td>
<td>25 P0.0 D I/O</td>
</tr>
<tr>
<td>J2-33 Input 3 (Cwlnh)</td>
<td>27 P0.1 D I/O</td>
</tr>
<tr>
<td>J2-34 Input 4 (Reg 1)</td>
<td>29 P0.2 D I/O</td>
</tr>
<tr>
<td>J2-35 Input 5 (Reg 2)</td>
<td>31 P0.3 D I/O</td>
</tr>
<tr>
<td>J2-36 Input 6</td>
<td>26 P0.4 Digital Source</td>
</tr>
<tr>
<td>J2-37 Enable Input</td>
<td></td>
</tr>
<tr>
<td>J2-38 Input RTN</td>
<td>33 Digital Ground</td>
</tr>
<tr>
<td>J2-39 +24VDC Output RTN</td>
<td></td>
</tr>
<tr>
<td>J2-40 +24VDC (Output)</td>
<td></td>
</tr>
</tbody>
</table>

Table B2. Wiring Between Sensors and Data Acquisition Card

<table>
<thead>
<tr>
<th>Other NI Card INPUTS</th>
<th>NI PCI-6025E Terminal</th>
</tr>
</thead>
<tbody>
<tr>
<td>LVDT</td>
<td>3 AI 0 Analog In 0</td>
</tr>
<tr>
<td>Load Cell</td>
<td>5 AI 1 Analog In 1</td>
</tr>
<tr>
<td>Pressure Transducer</td>
<td>7 AI 2 Analog In 2</td>
</tr>
<tr>
<td>Rotational Potentiometer</td>
<td>9 AI 3 Analog In 3</td>
</tr>
</tbody>
</table>
Table B3. Wiring Between PPR and Data Acquisition Card

<table>
<thead>
<tr>
<th>Other NI Card OUTPUTS</th>
<th>NI PCI-6025E Terminal</th>
</tr>
</thead>
<tbody>
<tr>
<td>Proportional Pressure Regulator</td>
<td>20 AO 0</td>
</tr>
</tbody>
</table>
APPENDIX B

DYNAMIC TEST STATION SETUP INSTRUCTIONS

Instructions for basic Dynamic Test Station setup and motor configuration are listed below.

1. Turn on DTS station computer.

2. Plug in and turn on 4 electrical boxes. These include the following: (1) main box with NI data acquisition terminal block, motor connection terminal block, and signal conditioner for force transducer/load cell, (2) Rotational Potentiometer amplifier, (3) 5 to 24V isolated converter for motor digital command, and (4) box containing power supplies.

3. Open the main regulator valve on the nitrogen tank. Turn the gauge control clockwise until the left gauge reads approximately 120 psi.

   *Note:* Remember to close the main regulator valve, and open the gauge control when finished running experiments. Bleed the system by commanding pressures to the PMA.

4. Plug in and turn on motor emergency shutoff box. Motor driver is on if green power light is lit. If red fault light is flashing, turn motor box off and then back on.

   *Note:* Keep motor disabled during the setup process. Only enable the motor once ready to run an experiment.

5. Open Pacific Scientific 800Tools.

6. Configure if necessary.

Notes on configuration are listed below.
If no changes are being made to the last configuration used, even if the system was shutdown, there is no need to reconfigure. However, if using analog command, the process of nulling the analog voltage on the line should be performed every so often to ensure accuracy of voltage command signals.

Default for the DTS system is Velocity Mode- Analog Command. Predefined Moves are set up so that digital position control can also be utilized.

To configure, follow steps below.

1. Click on Edit File.
2. Open DTS_VelMode_VelRange3,TrqRange2,newpos11.
3. In this configuration file, the settings under each tab are chosen as shown in Figures C1 through C9.
4. Under the Predefined Moves tab, Move# 4 to 7 are the same as Move#0, hold position.
5. Save to File and rename if changes were made.
6. Download to Drive.

If calibrating the rotational potentiometer, you can use the following configuration file: DTS_VelMode_VelRange2,PosRecord. The analog outputs with this configuration are position and velocity instead of the default torque and velocity as shown in Figure C10. The motor’s analog output position is used in the calibration process.

See Pacific Scientific Driver PC830 User Manual (218) for additional information on configuration options and the meanings and significance of different settings.
Figure C1. Digital I/O Settings for Velocity Mode Motor Setup in Pacific Scientific 800Tools Release 1.02, Pacific Scientific, 2002. Within this tab, the user can select the functionality for the six user mappable digital inputs.
Figure C2. Analog I/O Settings for Velocity Mode Motor Setup in Pacific Scientific 800Tools Release 1.02, Pacific Scientific, 2002. Within this tab, the user can select what two analog outputs are sent from the motor. Gains are set to scale the output (voltage output must be ±5V).
Figure C3. Loop Gains Settings for Velocity Mode Motor Setup in Pacific Scientific 800Tools Release 1.02, Pacific Scientific, 2002
Figure C4. Velocity Controller Settings for Velocity Mode Motor Setup in Pacific Scientific 800Tools Release 1.02, Pacific Scientific, 2002. Within this tab, the user can select the CmdGain which scales the voltage command signals.
Figure C5. Predefined Moves Move #0 Settings for Velocity Mode Motor Setup in Pacific Scientific 800Tools Release 1.02, Pacific Scientific, 2002. Within this tab, moves can be specified if the user is planning on switching from the analog mode to Position mode—predefines moves during the motor operation. The proper Digital I/O setup has to be completed before it is possible to switch between modes or use the predefine moves.
Figure C6. Predefined Moves Move #1 Settings for Velocity Mode Motor Setup in Pacific Scientific 800Tools Release 1.02, Pacific Scientific, 2002
Figure C7. Predefined Moves Move #2 Settings for Velocity Mode Motor Setup in Pacific Scientific 800Tools Release 1.02, Pacific Scientific, 2002
Figure C8. Predefined Moves Move #3 Settings for Velocity Mode Motor Setup in Pacific Scientific 800Tools Release 1.02, Pacific Scientific, 2002
Figure C9. Feedback Settings for Velocity Mode Motor Setup in Pacific Scientific 800Tools Release 1.02, Pacific Scientific, 2002. Within this tab, the user can set the resolution of the resolver’s emulated encoder.
Figure C10. Analog Output Position settings for Velocity Mode Motor Setup in Pacific Scientific 800Tools Release 1.02, Pacific Scientific, 2002
APPENDIX D

CALIBRATION PROCEDURES

Suggestions for calibrating the rotational potentiometer for the purpose of sensing motor shaft position are listed below.

- The first step is finding the appropriate starting motor count, which is a way of stating motor position.
- Command the motor to the appropriate motor count that represents 90° of flexion (or whatever angle is defined as the starting angle), while recording rotational potentiometer voltages.
- Configure the motor using the Pacific Scientific 800Tools program (Pacific Scientific Release 1.02) so that position is the motor’s first analog output. The motor’s position output is not used in the control system, but it is used for calibration purposes. This is why a different configuration file is necessary. The configuration file named DTS_VelMode_VelRange2,PosRecord can be used for calibration purposes.
- While the motor is disabled, run the LabVIEW program named SetupFile.vi. Find the motor position output that corresponds to the rotational potentiometer voltage previously recorded. The motor position output is shown in the DM1 box in units of Rev*10 if using the calibration file named here. The rotational potentiometer voltage is shown in the RotPot box.
- Treat the motor position value (in units of rev) as the starting point. Subtract increments of a full revolution from the starting revolution to create a set of revolutions. Three
increments of 1/16 each produce a set of four revolutions across the operating range. The operating range for a 90° rotation is 0.25 rev.

- With the motor still disabled and the LabVIEW SetupFile.vi still running, turn the motor shaft to find the rotational potentiometer voltage that corresponds to each revolution in the set.
- The revolution values can be converted into radians or degrees (depending on unit preference).
- The calibration curve can be constructed by fitting a line to the rotational potentiometer voltages as a function of angle.

Suggestions for calibrating the linear variable displacement transducer (LVDT) for the purpose of sensing PMA displacement are listed below.

- The LVDT can be calibrated using a set of calipers. First find a reference point for the caliper measurement. Using the LabVIEW program named SetupFile.vi, pressurize the PMA, record the LVDT voltage, and measure the actual displacement using the calipers. Repeat at different pressures to get several data points across the operating range. It is important to be consistent with the measurement procedure at the different pressures because the values do change over time, even when the pressure is held constant.
- Important note regarding setup and calibration: The relative starting position between the PMA and motor is very important to the force and displacement outcomes. Test basic model outcomes if major changes are made to the starting position of either the PMA or motor angle or if changes are made to the cable connecting the two.
APPENDIX E
LABVIEW DTS CONTROL INFORMATION

All LabVIEW Control files are located in C:\Documents and Settings\student\My Documents\LabVIEW Data\DTS LABVIEW CONTROLS. Data is automatically saved in the following folder: C:\Documents and Settings\student\My Documents\LabVIEW Data\DTS DATA COLLECTION.

The LabVIEW file SetupFile.vi can be used to calibrate, tare analog lines to the motor, clear analog lines to the motor or PPR, or clear digital lines to the motor. Data collected through these processes is automatically saved in C:\Documents and Settings\student\My Documents\LabVIEW Data\DTS SETUP DATA COLLECTION.

There are eight main LabVIEW files that work together to perform PMA DTS control. Each is described below.

E.1 Main DTS Control

This is the main control file. All the other LabVIEW files, also called subVIs, are either connected here or within one of the subVIs. As seen in the block diagram of the LabVIEW MainDTSControl.vi file (see actual file), the main While loop contains three sections. The first section collects sensor data, filters it, and converts it from voltages to proper units for use in calculations. The unit conversions are done within UnitConverter.vi. The second section contains controller subVIs including the Trajectory Generator, Task Motion Controller (TMC), PMA Motion Controller (PMC) and Human Dynamic Simulator (HDS). NOTE: The HDS is referred to as the HQDS within the body of the dissertation. This section also contains programming that
allows the user to override the HDS velocity command with the desired velocity. The third section sends the analog command voltages to the PPR and motor, and it contains code to automatically stop the loop at the proper time. The third section also contains a subVI, PosMode Operation, whose purpose is to control the motor using digital commands when necessary.

Outside of the main While loop there are Read from File functions which feed information into the system (from .txt files) including desired motion profiles, desired force profiles (if there is a need to override the controller calculated force command), and data used within the HDS to simulate different behaviors. Outside of the loop, there is also code used to write sensor data and outputs from control calculations to files. Settings of the While loop, such as effective period or rate of iteration calculations, can be adjusted within this file.

As seen in the front panel of the LabVIEW MainDTSControl.vi file, Figure E1, there are several set-up options including the following:

- **System Operation**: ON selection activates main case structure containing the While loop with control functionality.
- **Analog Output Operation**: ON selection activates the functionality of the analog output command to the PPR and motor. If OFF, no signals will be sent to the motor or PPR.
- **Desired Motion Profile**: Select desired motion profile from drop down menu of pre-loaded motion profiles. It is possible to add or change the desired motion profiles by creating or making edits to .txt files from which the desired profiles originate. Make sure to link the Read From Motion Profile block to the correct .txt file.
- **SlackSwitch**: This is a backup safety precaution that commands 0V to both the PPR and motor when the difference between the PMA displacement and motor displacement is deemed too large from which to recover.
- **VelCmdOverride**: Set this to ON only if you wish to command the desired velocity instead of the HDS calculated velocity which aims to simulate a neuromuscular actuator.
• **Position Mode:** ON selection activates the PosMode Operation subvi. If ON is chosen, select the desired type of position mode operation from the drop down menu. Options include the following:
  
  o **PosMode for to,hold,tf:** This triggers the motor to send commands designed to replicate hard stops during exercise. It positions the motor at the correct staring position, positions it at the correct hold location during the isometric phase, and then positions it at the correct stopping location once the task is over. The commands are set up to allow the motor to reach the desired locations on its own, but if it stops short of the desired position or goes past the desired position, the position mode operation will step in and adjust motor positioning so that the next phase begins at the correct starting location.
  
  o **PosMode for to,semi-hold,tf:** This acts similar to PosMode for to,hold,tf described above, but instead of maintaining the hold position throughout the isometric phase, it stops digital commands during the isometric hold.
  
  o **PosMode for to,tf:** This acts similar to PosMode for to,hold,tf and PosMode for to,semi-hold,tf described above, but there are no digital commands during the isometric hold.
  
  o **All PosMode:** This overrides analog velocity commands throughout the simulated task, allowing the motor to run in position mode throughout the task. Proper settings have to be chosen depending on the desired motor actions.
  
• **User select torque for isometric hold:** If ON is chosen, this allows the user to select the isometric force resistance within the PMC instead of relying on the torque-velocity relationship based isometric value. The torque-velocity isometric value used is the scaled isometric value, not the peak value.

• **Calculation Override:** If for testing purposes, there is a need to command 0V to either the PPR or motor (or both), these buttons can be set to ON.
• **Force Override:** If there is a desire to command force resistances directly into the PMA Motion Controller, this setting can be set to ON. Otherwise, the Task Motion Controller calculates the force command with the goal of maintaining isovelocity movement.

*Figure E1. Front Panel of the Main DTS Control LabVIEW VI (LabVIEW 8.0, National Instruments, 2005)*

As discussed in the body of the dissertation, data from sensors is sent through a fifth order low pass finite impulse response (FIR) filter before the voltages are converted to proper
units for use in control calculations. The advantage of using an FIR filter is that this type of filter keeps the delay for all frequency components the same. Filter adjustments can be made in the MainDTSControl block diagram.

Voltages are collected and commanded at a rate of 1 kHz with a 100 sample buffer size. Timing can be adjusted within the MainDTSControl block diagram, within the DAQ Assistant blocks.

E.2 Unit Converter

If calibrations are performed or gains are adjusted in the motor configuration setup (Analog I/O tab), the equations contained within the UnitConverter.vi file must be updated.

E.3 Trajectory Generator

The LabVIEW TrajectoryGenerator.vi file is designed to take the profile fed into it and the current task time, and, using interpolation, calculate the desired point to be used in the control system for the current iteration. If the timing of the system or desired profiles changes, adjustments may need to be made to ensure correct interpolation.

E.4 Task Motion Controller (TMC)

The calculations within the LabVIEW TaskMotionController.vi file can be separated into six parts, each corresponding to an element of the nonlinear adaptive control law (Eq. (34)). Three of the parts contain adaptation calculations. Simpson’s Rule, a discrete integration method, is used within the TMC for adaptation calculations. It is important to correctly set the initial conditions for this block within the block diagram if changes are made to the adaptation mechanism. It is also important to set or adjust the initialization command of the integration block based on when adaptation needs to start over. For example, if going from one phase of the task to another, you may want to reinitialize the adaptation calculation so that it starts near a predefined
initial condition instead of the last value found. A seventh section of code pertains to the creation of a boundary layer around the sliding surface designed to alleviate chatter or high frequency oscillations in the control signal. Chatter did not present itself as a problem in this work, so the boundary layer was not used. It was kept within the TMC for purposes of future work. The boundary layer works by replacing the control signal when error is within the boundary to prevent the control signal from switching back and forth across the surface, thus smoothing control activity (215). The drawbacks to the boundary layer method are that it may reduce control accuracy, and it may not be effective with a high level of measurement noise (220). A low pass filter in combination with a disturbance estimator may be more effective. See (220) for more information.

Additional code was created to make calculation adjustments to the force resistance command before it is sent to the PMC.

As seen in the front panel of the LabVIEW TaskMotionController.vi file, Figure E2, there are several set-up options including the following:

- **LegAttached:** If ON is chosen, the gravity and inertial terms contained within the sliding mode control calculations are adjusted based on lower leg characteristics. The mass, length, and center of mass values will have to be adjusted within the block diagram.
- **BoundaryThickness:** If ON is chosen, a boundary zone is created as previously discussed.
- **Scaling parameters for torque and velocity:** The scaling parameters values predicting human quadriceps torque output used within the nonlinear adaptive control calculation can be adjusted here.
- **Adaptation for c:** If ON is chosen, there will be NO adaptation of the predicted human torque coefficient, c. The initial c value has a default value of 1 but can be changed.
- **Adaptation for b:** If ON is chosen, there will be NO adaptation of the gravity term, b. The initial b value has a default value of 1 but can be changed.
Adaptation for $I$: If ON is chosen, there will be NO adaptation of the inertia term, $I$.

The front panel lists inputs and outputs of the subVI, but these are for informational purposes only. The inputs are fed in from other VIs. The outputs are sent to other VIs.

Figure E2. Front Panel of the Task Motion Controller LabVIEW VI (LabVIEW 8.0, National Instruments, 2005)

E.5 PMA Motion Controller (PMC)

The code within the LabVIEW PMAMotionController.vi file is based on the three-element phenomenological model previously discussed. There are five sets of calculations, each relating to a pressure range and PMA action (either contraction or relaxation). The calculation used for the PPR command is chosen using a baseline pressure calculation and desired velocity, which indicates in what mode the PMA is operating.

There are two subVIs within PMAMotionController.vi, PID_F_inPMC.vi and PID_x_inPMC.vi. Each aims to adjust either the force command or displacement command used for the PMC calculations. Incorporation of the PID_x_inPMC.vi was not found to improve performance. The PID based force adjustment, however, was found to improve the force output during the isometric phase of the task.
As seen in the front panel of the LabVIEW PMAMotionController.vi file, Figure E3, there are several set-up options including the following:

- **PID_x**: If ON is chosen, PID operating on displacement, x, becomes active.
- **PID_F**: If ON is chosen, PID operating on force command, FR_d, becomes active according to the timing settings.
- **PID_F start time**: ON will trigger the PID based adjustment of the force command to begin 5 seconds into the task. OFF will trigger the adjustment to begin once the hold phase begins.
- **PID_F end time**: ON will trigger the PID based adjustment of the force command to end at the end of the task. OFF will trigger the adjustment to end once the hold phase ends.
- **User selected torque resistance for isometric hold**: This is related to settings in the MainDTS.vi. If there is a desire to select the isometric hold resistance, one can enter the value here. The PID_F will adjust the command to achieve this resistance level vs. the default value based on the torque-velocity relationship.

![Front Panel of the PMA Motion Controller LabVIEW VI](image)

*Figure E3. Front Panel of the PMA Motion Controller LabVIEW VI (LabVIEW 8.0, National Instruments, 2005)*
Similar to the TMC, the PMC front panel lists inputs and outputs of the subVI, but these are for informational purposes only.

### E.6 PID_F in PMC

The code within PID_F in PMC.vi determines how to adjust FR_d according to force error between the target force (as defined by the simulated neuromuscular actuator’s force-velocity curve or predefined isometric hold force) and the actual force output. It is only active according to the settings found in the PMC subVI. The proportional (Kp), integral (Ki) and derivative (Kd) coefficients can be set on the front panel of the PID_F_inPMC.vi file. Time varying adjustments can also be made to the coefficients by setting the PID Gain adaptation or PID Gain Select to ON. If adaptation is ON, the PID coefficients are adjusted based on the magnitude and sign of the force error (F_{target}-F_{out}). The PID Gain Select set to ON activates a nonlinear scaling factor which adjusts the fixed gains according to changes in the magnitude of force error. A large force error will create a large change in the PID coefficients in an attempt to make a quick impactful change in the output. Small force errors will not have as large of an impact on PID coefficients or the outcome. Fixed gain PID worked well for our application and time-varying adjustments were not necessary.

Simpson’s Rule, a discrete integration method, is used within the PID_F_inPMC.vi for integration of error. The 2nd order central method of discrete differentiation is used to take the derivative of error.

Similar to the TMC and PMC, the PID_F front panel lists inputs and outputs of the subVI for informational purposes only.
E.7 **Human Dynamic Simulator (HDS)**

The functionality that this subvi provides is referred to as Human Quadriceps Dynamic Simulator (HQDS) within the body of the dissertation. As seen in the front panel of the LabVIEW HumanDynamicSimulator.vi file, Figure E4, there are several set-up options including the following:

- **Human Type**: There are several options for human type (simulated neuromuscular behavior type).
  - **Healthy**: For this simulated neuromuscular behavior type, $a_{Tc}$, $a_{Te}$, and $a_V$ are fixed according to front panel settings. The $a_{Tc}$, $a_{Te}$, and $a_V$ correspond to the $a_{s,Tc}$, $a_{s,Te}$, and $a_{s,V}$ discussed in the body of the dissertation.
  - **Fatigued_incremental**: Here, $a_{Tc}$, $a_{Te}$, and $a_V$ are multiplied by a fatigue parameter intended to further scale the torque-velocity relationship according to torque findings seen at 20 repetitions (during the fatigue phase) or at 40 repetitions (during the endurance phase).
  - **Fatigued_time dependent**: Here, $a_{Tc}$, $a_{Te}$, and $a_V$ are multiplied by a nonlinear function which alters the original scaling parameter scaling according to actual task time and a time to fatigue setting from the front panel. The nonlinear function originates from isometric behavioral findings (188). This is referred to as Fatigued-rapid in the body of the dissertation.
  - **Erratic**: For this simulated neuromuscular behavior type, the value of a sine wave function is added to $a_{Tc}$, $a_{Te}$, and $a_V$. This creates a set of time varying scaling parameters, and thus a time varying torque-velocity relationship. The amplitude, frequency, phase offset, and bias can be adjusted as desired. This is referred to as Erratic-sine wave in the dissertation.
Scaling Parameter Spike: For this simulated neuromuscular behavior type, a predefined value is either added to or subtracted from the $a_{Tc}$, $a_{Te}$, and $a_{V}$ scaling parameters for use in HDS calculations. The predefined value originates from a .txt file linked to the system within the MainDTS.vi file. The .txt file can be changed or replaced as desired. This behavior type is referred to as Erratic-pulse in the dissertation.

- Fatigue Level: This setting is only applicable to the Fatigued_incremental simulated neuromuscular behavior type. The fatigue level can be set so that the torque-velocity relationship is scaled to the equivalent output potential of the simulated neuromuscular actuator either after 20 repetitions or 40 repetitions.
- $TF_c$: This parameter, only used in the Fatigued_time dependent simulated neuromuscular behavior type, is the time to concentric fatigue setting.
- $TF_e$: This parameter, also only used in the Fatigued_time dependent simulated neuromuscular behavior type, is the time to eccentric fatigue setting.
- $TF_V$: This parameter, also only used in the Fatigued_time dependent simulated neuromuscular behavior type, is the velocity time to fatigue setting.
- $a_{Tc}$, $a_{Te}$, $a_{V}$: These are the concentric torque, eccentric torque, and velocity scaling parameters, respectively.
- Eccentric Plateau: The plateau effect of eccentric torque above a certain velocity can either be turned ON or OFF using this control. The plateau characteristic of eccentric torque beyond a certain region of low velocity is applied by using dynamic saturation. The scaled peak eccentric torque cannot exceed that of scaled peak concentric torque at each joint angle. Thus eccentric torque saturates at the isometric level (peak concentric equal to the peak isometric) at the current simulated joint angle of the system. This parameter can easily be adjusted within the block diagram.
• *Human Velocity Disturbance*: Velocity disturbances can be added to the motor velocity command. The disturbance options include a velocity pulse as defined by a .txt file, random noise with amplitude that can be adjusted within the block diagram, or a sine wave whose properties can be adjusted within the block diagram.

The target torque for each iteration is based on the torque corresponding to the desired velocity and actual joint angle output from the scaled torque-velocity-position relationship. Code is present in the block diagram to adjust target torque if necessary to prevent discontinuities while transitioning from the concentric to eccentric torque-velocity relationship.

Clockwise or counterclockwise motor velocity is commanded depending on the actual PMA resistance transferred to the motor. The HDS velocity command is the velocity corresponding to the torque and length output of the previous iteration. Additional code is designed to appropriately simulate reactions to stimuli during the isometric hold. There is also code designed to correctly assign the voltage signal with the proper sign (+ for clockwise velocity, - for counterclockwise velocity).

Make sure to update the velocity to voltage calculation if the gain is changed in the motor configuration (Velocity Controller tab). If, for example, the motor velocity output range needs to be increased beyond what the current gain setting allows, a new gain would be assigned during configuration (using Pacific Scientific 800 tools) and then incorporated into the block diagram.
E.8  PosMode Operation

The PosModeOperation.vi file determines when it is necessary to send digital signals to the motor and what signals are appropriate. The set of hi or lo digital signals on 6 channels corresponds to an assigned ‘move’ set up in the motor configuration. Absolute moves are used here to command the motor to specified locations. See motor configuration instructions and Pacific Scientific Driver PC830 User Manual for more information.

The PosMode option selected within the MainDTSControl.vi file determines the manner in which the signals are sent. Determination of the command signal depends on timing of the desired motion profile selected, the actual task time, and the actual motor position. The activation timing of the digital motor position commands can be tweaked within the block diagram. The specified motor locations must be changed within the Predefined Moves tab of the motor configuration set up.
APPENDIX F

CLOSED LOOP CONTROL COMMAND SIGNALS

Figure F1. Torque, Pressure, and PPR commands for 0.175 rad/s simulation scenario
Figure F2. Torque, Pressure, and PPR commands for 0.314 rad/s simulation scenario

Figure F3. Torque, Pressure, and PPR commands for 0.524 rad/s simulation scenario
Figure F4. Torque, Pressure, and PPR commands for 0.314 rad/s, 20-repetitions fatigue simulation scenario

Figure F5. Torque, Pressure, and PPR commands for 0.314 rad/s, 40-repetitions fatigue simulation scenario
Figure F6. Torque, Pressure, and PPR commands for 0.314 rad/s, rapid fatigue simulation scenario

Figure F7. Torque, Pressure, and PPR commands for 0.314 rad/s, erratic-sine wave simulation scenario
Figure F8. Torque, Pressure, and PPR commands for 0.314 rad/s, erratic-pulse simulation scenario
APPENDIX G

DESIRED MOTION PROFILES

The desired motion profiles for task velocities of 0.175 rad/s (10°/s), 0.314 rad/s (18°/s), and 0.524 rad/s (30°/s) are shown, in part, below. The profile within the DTS has a frequency of 20 samples per second and interpolation is performed within LabVIEW to find the correct data point corresponding to the exact time.

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APPENDIX H

ADDITIONAL CLOSED LOOP CONTROL PERFORMANCE RESULTS

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218
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| 4 | 0.314 rad/s, Rapid Fatigue | Concentric | 0.264240141 | 0.014546 | 0.0216 | 0.86305 | 0.312187 | 4.659389 |
| 5 | 0.314 rad/s, Rapid Fatigue | Concentric | 0.216302869 | 0.012744 | 0.02147 | 0.90991 | 0.314149 | 4.056669 |
| 1 | 0.314 rad/s, Rapid Fatigue | Concentric | 0.626752725 | 0.016493 | 0.01428 | 0.54047 | 0.314426 | 5.245424 |
| 2 | 0.314 rad/s, Rapid Fatigue | Concentric | 0.696267337 | 0.024387 | 0.01237 | 0.49304 | 0.313431 | 7.780657 |
| 3 | 0.314 rad/s, Rapid Fatigue | Concentric | 0.509625308 | 0.019775 | 0.01648 | 0.58611 | 0.315718 | 6.263510 |
| 4 | 0.314 rad/s, Rapid Fatigue | Concentric | 0.451300229 | 0.018781 | 0.0127 | 0.50684 | 0.316264 | 5.938390 |
| 5 | 0.314 rad/s, Rapid Fatigue | Concentric | 0.473728012 | 0.020386 | 0.01407 | 0.56887 | 0.315479 | 6.461911 |
| 1 | 0.314 rad/s, Erratic - sine wave Concentric | 0.415438592 | 0.030076 | 0.01571 | 1.12535 | 0.316152 | 9.513294 |
| 2 | 0.314 rad/s, Erratic - sine wave Concentric | 0.373495443 | 0.02793 | 0.01692 | 1.09655 | 0.318936 | 8.75725 |
| 3 | 0.314 rad/s, Erratic - sine wave Concentric | 0.354561215 | 0.029805 | 0.01715 | 1.09041 | 0.316152 | 9.771508 |
| 4 | 0.314 rad/s, Erratic - sine wave Concentric | 0.505691413 | 0.031152 | 0.01725 | 1.14413 | 0.318804 | 9.771508 |
| 5 | 0.314 rad/s, Erratic - sine wave Concentric | 0.534771151 | 0.027549 | 0.01865 | 1.07198 | 0.316152 | 8.713852 |
| 1 | 0.314 rad/s, Erratic - pulse Concentric | 0.446528955 | 0.02671 | 0.01969 | 1.53128 | 0.309783 | 8.622165 |
| 2 | 0.314 rad/s, Erratic - pulse Concentric | 0.413416635 | 0.026718 | 0.01513 | 1.33854 | 0.311677 | 8.572334 |
| 3 | 0.314 rad/s, Erratic - pulse Concentric | 0.404007164 | 0.027152 | 0.01812 | 1.35654 | 0.316696 | 8.573522 |
| 4 | 0.314 rad/s, Erratic - pulse Concentric | 0.450383878 | 0.024208 | 0.01897 | 1.38994 | 0.312874 | 7.737302 |
| 5 | 0.314 rad/s, Erratic - pulse Concentric | 0.350219863 | 0.02483 | 0.0149 | 1.35085 | 0.309783 | 8.015288 |
| 1 | 0.314 rad/s, Eccentric | 0.558825574 | 0.044918 | 0.01708 | 1.55192 | 0.303099 | 14.8196 |
| 2 | 0.314 rad/s, Eccentric | 0.42147353 | 0.042317 | 0.01547 | 1.45326 | 0.310121 | 13.6453 |
| 3 | 0.314 rad/s, Eccentric | 0.507221144 | 0.045611 | 0.01486 | 1.48822 | 0.303865 | 15.0103 |</p>
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APPENDIX I

ANOVA RESULTS EVALUATING EFFECT OF SIMULATION SCENARIO AND PHASE ON TRACKING ACCURACY

Results were found using JMP 9.0.0 (SAS Institute Inc.). This was a multiparameter test. Results for each parameter (position RMSE, velocity CV RMSE, torque RMSE, and PMA displacement RMSE) are listed separately.

I.1 Position RMSE ANOVA Results

Table I1. Analysis of Variance for Position RMSE

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<th>Mean Square</th>
<th>F Ratio</th>
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Table I2. Effect Tests Results for Position RMSE

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Figure I1. Residual by Predicted Plot for Position RSME

Figure I2. Residuals Distribution Analysis Results for Position RMSE

Table I3. LSMeans Differences Tukey HSD Results for Simulation Scenario

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<tr>
<td>0.314 rad/s, Erratic-pulse</td>
<td>0.01628715</td>
</tr>
<tr>
<td>0.175 rad/s</td>
<td>0.01217242</td>
</tr>
</tbody>
</table>

Levels not connected by same letter are significantly different.
Table I4. LSMeans Differences Student’s t Results for Phase

\( \alpha=0.050 \ t=1.99773 \)

<table>
<thead>
<tr>
<th>Level</th>
<th>Least Sq Mean</th>
<th>Concentric</th>
<th>0.02016421</th>
</tr>
</thead>
<tbody>
<tr>
<td>Eccentric</td>
<td>B</td>
<td>0.01603241</td>
<td></td>
</tr>
</tbody>
</table>

Levels not connected by same letter are significantly different.

Figure I3. LS Means Plot for Simulation Scenario

Figure I4. LS Means Plot for Phase
Table I5. LSMeans Differences Tukey HSD for Interaction between Simulation Scenario and Phase

\[ \alpha=0.050 \quad Q=3.56554 \]

<table>
<thead>
<tr>
<th>Level</th>
<th>Least Sq Mean</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.314 rad/s, 40rep Fatigue, Concentric</td>
<td>A 0.03049397</td>
</tr>
<tr>
<td>0.524 rad/s, Concentric</td>
<td>A 0.02813544</td>
</tr>
<tr>
<td>0.314 rad/s, 20rep Fatigue, Concentric</td>
<td>B 0.02408767</td>
</tr>
<tr>
<td>0.524 rad/s, Eccentric</td>
<td>B 0.02190556</td>
</tr>
<tr>
<td>0.314 rad/s, Erratic-sine wave, Eccentric</td>
<td>B 0.02058271</td>
</tr>
<tr>
<td>0.314 rad/s, Rapid Fatigue, Concentric</td>
<td>C 0.01950138</td>
</tr>
<tr>
<td>0.314 rad/s, Eccentric</td>
<td>C 0.01826057</td>
</tr>
<tr>
<td>0.314 rad/s, Erratic-pulse, Concentric</td>
<td>D 0.01736414</td>
</tr>
<tr>
<td>0.314 rad/s, Erratic-sine wave, Concentric</td>
<td>D 0.01713749</td>
</tr>
<tr>
<td>0.314 rad/s, Erratic-pulse, Eccentric</td>
<td>D 0.01521016</td>
</tr>
<tr>
<td>0.314 rad/s, Concentric</td>
<td>E 0.01468768</td>
</tr>
<tr>
<td>0.175 rad/s, Eccentric</td>
<td>F 0.01443893</td>
</tr>
<tr>
<td>0.314 rad/s, 20rep Fatigue, Eccentric</td>
<td>F 0.0135011</td>
</tr>
<tr>
<td>0.314 rad/s, Rapid Fatigue, Eccentric</td>
<td>F 0.01397893</td>
</tr>
<tr>
<td>0.175 rad/s, Concentric</td>
<td>G 0.00990591</td>
</tr>
<tr>
<td>0.314 rad/s, 40rep Fatigue, Eccentric</td>
<td>G 0.00953233</td>
</tr>
</tbody>
</table>

Levels not connected by same letter are significantly different.

I.2 Velocity CV RMSE ANOVA Results

Table I6. Analysis of Variance for Velocity CV RMSE

<table>
<thead>
<tr>
<th>Source</th>
<th>DF</th>
<th>Sum of Squares</th>
<th>Mean Square</th>
<th>F Ratio</th>
<th>Prob &gt; F</th>
</tr>
</thead>
<tbody>
<tr>
<td>Model</td>
<td>15</td>
<td>583.86837</td>
<td>38.9246</td>
<td>38.4226</td>
<td>&lt;.0001*</td>
</tr>
<tr>
<td>Error</td>
<td>64</td>
<td>64.83604</td>
<td>1.0131</td>
<td></td>
<td></td>
</tr>
<tr>
<td>C. Total</td>
<td>79</td>
<td>648.70441</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Table I7. Effect Tests Results for Velocity CV RMSE

<table>
<thead>
<tr>
<th>Source</th>
<th>Nparm</th>
<th>DF</th>
<th>Sum of Squares</th>
<th>F Ratio</th>
<th>Prob &gt; F</th>
</tr>
</thead>
<tbody>
<tr>
<td>Simulation Scenario</td>
<td>7</td>
<td>7</td>
<td>371.17864</td>
<td>52.3418</td>
<td>&lt;.0001*</td>
</tr>
<tr>
<td>Phase</td>
<td>1</td>
<td>1</td>
<td>17.05895</td>
<td>16.8390</td>
<td>0.0001*</td>
</tr>
<tr>
<td>Simulation Scenario*Phase</td>
<td>7</td>
<td>7</td>
<td>195.63077</td>
<td>27.5869</td>
<td>&lt;.0001*</td>
</tr>
</tbody>
</table>
Figure 15. Residual by Predicted Plot for Velocity CV RMSE

Figure 16. Residuals Distribution Analysis Results for Velocity CV RMSE

Table I8. LSMean Differences Tukey HSD for Simulation Scenario

<table>
<thead>
<tr>
<th>Level</th>
<th>Least Sq Mean</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.314 rad/s, Erratic-pulse</td>
<td>11.351977</td>
</tr>
<tr>
<td>0.314 rad/s, Erratic-sine wave</td>
<td>10.217800</td>
</tr>
<tr>
<td>0.314 rad/s, 40rep Fatigue</td>
<td>10.158620</td>
</tr>
<tr>
<td>0.175 rad/s</td>
<td>9.446502</td>
</tr>
<tr>
<td>0.314 rad/s, 20rep Fatigue</td>
<td>8.921246</td>
</tr>
<tr>
<td>0.314 rad/s</td>
<td>6.194561</td>
</tr>
<tr>
<td>0.314 rad/s, Rapid Fatigue</td>
<td>5.713498</td>
</tr>
<tr>
<td>0.524 rad/s</td>
<td>5.472880</td>
</tr>
</tbody>
</table>

Levels not connected by same letter are significantly different.
Figure 17. LS Means Plot for Simulation Scenario

Table I9. LSMeans Differences Student's t for Phase

\[ \alpha = 0.050 \quad t = 1.99773 \]

<table>
<thead>
<tr>
<th>Level</th>
<th>Least Sq Mean</th>
</tr>
</thead>
<tbody>
<tr>
<td>Eccentric A</td>
<td>8.8964113</td>
</tr>
<tr>
<td>Concentric B</td>
<td>7.9728596</td>
</tr>
</tbody>
</table>

Levels not connected by same letter are significantly different.

Figure 18. LS Means Plot for Phase
### Table I10. LSMeans Differences Tukey HSD for Interaction between Simulation Scenario and Phase

\[ \alpha = 0.050 \quad Q = 3.56554 \]

<table>
<thead>
<tr>
<th>Level</th>
<th>Least Sq Mean</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.314 rad/s, Erratic-pulse, Eccentric A</td>
<td>14.399832</td>
</tr>
<tr>
<td>0.314 rad/s, 40rep Fatigue, Concentric A B</td>
<td>12.840198</td>
</tr>
<tr>
<td>0.314 rad/s, Erratic-sine wave, Eccentric B C</td>
<td>11.198957</td>
</tr>
<tr>
<td>0.175 rad/s, Eccentric B C</td>
<td>10.641210</td>
</tr>
<tr>
<td>0.314 rad/s, 20rep Fatigue, Concentric C D</td>
<td>9.825982</td>
</tr>
<tr>
<td>0.314 rad/s, Erratic-sine wave, Concentric C D E</td>
<td>9.236643</td>
</tr>
<tr>
<td>0.314 rad/s, Erratic-pulse, Concentric D E F</td>
<td>8.304122</td>
</tr>
<tr>
<td>0.175 rad/s, Concentric D E F</td>
<td>8.251794</td>
</tr>
<tr>
<td>0.314 rad/s, 20rep Fatigue, Eccentric D E F G</td>
<td>8.016510</td>
</tr>
<tr>
<td>0.314 rad/s, 40rep Fatigue, Eccentric E F G</td>
<td>7.477041</td>
</tr>
<tr>
<td>0.314 rad/s, Eccentric E F G H</td>
<td>7.197425</td>
</tr>
<tr>
<td>0.314 rad/s, Rapid Fatigue, Eccentric F G H</td>
<td>6.337979</td>
</tr>
<tr>
<td>0.524 rad/s, Eccentric G H</td>
<td>5.902336</td>
</tr>
<tr>
<td>0.314 rad/s, Concentric H</td>
<td>5.191696</td>
</tr>
<tr>
<td>0.314 rad/s, Rapid Fatigue, Concentric H</td>
<td>5.089018</td>
</tr>
<tr>
<td>0.524 rad/s, Concentric H</td>
<td>5.043424</td>
</tr>
</tbody>
</table>

Levels not connected by same letter are significantly different.

### I.3 Torque RMSE ANOVA Results

#### Table I11. Analysis of Variance Results for Torque RMSE

<table>
<thead>
<tr>
<th>Source</th>
<th>DF</th>
<th>Sum of Squares</th>
<th>Mean Square</th>
<th>F Ratio</th>
<th>Prob &gt; F</th>
</tr>
</thead>
<tbody>
<tr>
<td>Model</td>
<td>15</td>
<td>5.0151763</td>
<td>0.334345</td>
<td>43.2551</td>
<td>&lt;.0001*</td>
</tr>
<tr>
<td>Error</td>
<td>64</td>
<td>0.4946946</td>
<td>0.007730</td>
<td></td>
<td></td>
</tr>
<tr>
<td>C. Total</td>
<td>79</td>
<td>5.5098709</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

#### Table I12. Effect Tests Results for Torque RMSE

<table>
<thead>
<tr>
<th>Source</th>
<th>Nparm</th>
<th>DF</th>
<th>Sum of Squares</th>
<th>F Ratio</th>
<th>Prob &gt; F</th>
</tr>
</thead>
<tbody>
<tr>
<td>Simulation Scenario</td>
<td>7</td>
<td>7</td>
<td>3.9799754</td>
<td>73.5572</td>
<td>&lt;.0001*</td>
</tr>
<tr>
<td>Phase</td>
<td>1</td>
<td>1</td>
<td>0.6902935</td>
<td>89.3052</td>
<td>&lt;.0001*</td>
</tr>
<tr>
<td>Simulation Scenario*Phase</td>
<td>7</td>
<td>7</td>
<td>0.3449074</td>
<td>6.3745</td>
<td>&lt;.0001*</td>
</tr>
</tbody>
</table>
Figure 19. Residual by Predicted Plot for Torque RMSE

Figure 110. Residuals Normal Distribution Analysis for Torque RMSE

Table I13. LSMeans Differences Tukey HSD for Simulation Scenario

\[ \alpha = 0.050 \quad Q = 3.13337 \]

<table>
<thead>
<tr>
<th>Level</th>
<th>Least Sq Mean</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.314 rad/s, Erratic-pulse</td>
<td>A 1.4272045</td>
</tr>
<tr>
<td>0.524 rad/s</td>
<td>B 1.0799843</td>
</tr>
<tr>
<td>0.314 rad/s, Erratic-sine wave</td>
<td>B 1.0482449</td>
</tr>
<tr>
<td>0.314 rad/s, 40rep Fatigue</td>
<td>C 0.9103740</td>
</tr>
<tr>
<td>0.314 rad/s</td>
<td>C 0.9064743</td>
</tr>
<tr>
<td>0.175 rad/s</td>
<td>D 0.7354993</td>
</tr>
<tr>
<td>0.314 rad/s, Rapid Fatigue</td>
<td>D 0.7352115</td>
</tr>
<tr>
<td>0.314 rad/s, 20rep Fatigue</td>
<td>D 0.7304453</td>
</tr>
</tbody>
</table>

Levels not connected by same letter are significantly different.
Figure I11. LS Means Plot for Simulation Scenario

Table I14. LSMeans Differences Student’s t for Phase

\[ \alpha = 0.050 \ t = 1.99773 \]

<table>
<thead>
<tr>
<th>Level</th>
<th>Least Sq Mean</th>
</tr>
</thead>
<tbody>
<tr>
<td>Concentric</td>
<td>A 1.0395704</td>
</tr>
<tr>
<td>Eccentric</td>
<td>B 0.8537892</td>
</tr>
</tbody>
</table>

Levels not connected by same letter are significantly different.

Figure I12. LS Means Plot for Phase
### Table I.5. LSMeans Differences Tukey HSD for Interaction between Simulation Scenario and Phase

<table>
<thead>
<tr>
<th>Level</th>
<th>Least Sq Mean</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.314 rad/s, Erratic-pulse, Eccentric</td>
<td>A 1.4609791</td>
</tr>
<tr>
<td>0.314 rad/s, Erratic-pulse, Concentric</td>
<td>A B 1.3934300</td>
</tr>
<tr>
<td>0.524 rad/s, Concentric</td>
<td>B C 1.1962850</td>
</tr>
<tr>
<td>0.314 rad/s, Erratic-sine wave, Concentric</td>
<td>C D 1.1056332</td>
</tr>
<tr>
<td>0.314 rad/s, Concentric</td>
<td>C D E 1.0413721</td>
</tr>
<tr>
<td>0.314 rad/s, 40rep Fatigue, Concentric</td>
<td>C D E 1.0204785</td>
</tr>
<tr>
<td>0.314 rad/s, Erratic-sine wave, Eccentric</td>
<td>D E F 0.9908566</td>
</tr>
<tr>
<td>0.524 rad/s, Eccentric</td>
<td>D E F G 0.9638636</td>
</tr>
<tr>
<td>0.314 rad/s, Rapid Fatigue, Concentric</td>
<td>D E F G 0.9313555</td>
</tr>
<tr>
<td>0.314 rad/s, 20rep Fatigue, Concentric</td>
<td>E F G H 0.8541087</td>
</tr>
<tr>
<td>0.314 rad/s, 40rep Fatigue, Eccentric</td>
<td>F G H I 0.8002695</td>
</tr>
<tr>
<td>0.175 rad/s, Concentric</td>
<td>G H I 0.7739004</td>
</tr>
<tr>
<td>0.314 rad/s, Eccentric</td>
<td>G H I 0.7715766</td>
</tr>
<tr>
<td>0.175 rad/s, Eccentric</td>
<td>G H I J 0.6970983</td>
</tr>
<tr>
<td>0.314 rad/s, 20rep Fatigue, Eccentric</td>
<td>I J 0.6067820</td>
</tr>
<tr>
<td>0.314 rad/s, Rapid Fatigue, Eccentric</td>
<td>J 0.5390676</td>
</tr>
</tbody>
</table>

Levels not connected by same letter are significantly different.

### I.4 PMA Displacement RMSE ANOVA Results

**Table I.6. Analysis of Variance Results for PMA Displacement RMSE**

<table>
<thead>
<tr>
<th>Source</th>
<th>DF</th>
<th>Sum of Squares</th>
<th>Mean Square</th>
<th>F Ratio</th>
<th>Prob &gt; F</th>
</tr>
</thead>
<tbody>
<tr>
<td>Model</td>
<td>15</td>
<td>0.7464977</td>
<td>0.049767</td>
<td>10.3921</td>
<td></td>
</tr>
<tr>
<td>Error</td>
<td>64</td>
<td>0.3064880</td>
<td>0.004789</td>
<td></td>
<td>&lt;.0001*</td>
</tr>
<tr>
<td>C. Total</td>
<td>79</td>
<td>1.0529857</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Source</th>
<th>Nparm</th>
<th>DF</th>
<th>Sum of Squares</th>
<th>F Ratio</th>
<th>Prob &gt; F</th>
</tr>
</thead>
<tbody>
<tr>
<td>Simulation Scenario</td>
<td>7</td>
<td>7</td>
<td>0.06142511</td>
<td>1.8324</td>
<td>0.0962</td>
</tr>
<tr>
<td>Phase</td>
<td>1</td>
<td>1</td>
<td>0.33890593</td>
<td>70.7694</td>
<td>&lt;.0001*</td>
</tr>
<tr>
<td>Simulation Scenario*Phase</td>
<td>7</td>
<td>7</td>
<td>0.34616668</td>
<td>10.3265</td>
<td>&lt;.0001*</td>
</tr>
</tbody>
</table>
Figure I13. Residual by Predicted Plot for PMA Displacement RMSE

Figure I14. Residuals Distribution Analysis for PMA Displacement RMSE

Table I18. LSMeans Differences Tukey HSD for Simulation Scenario

\[ \alpha = 0.050 \quad Q = 3.13337 \]

<table>
<thead>
<tr>
<th>Level</th>
<th>Least Sq Mean</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.314 rad/s, 40rep Fatigue</td>
<td>A 0.46009056</td>
</tr>
<tr>
<td>0.314 rad/s, Erratic-pulse</td>
<td>A 0.44607046</td>
</tr>
<tr>
<td>0.524 rad/s</td>
<td>A 0.44082905</td>
</tr>
<tr>
<td>0.314 rad/s, Erratic-sine wave</td>
<td>A 0.41411678</td>
</tr>
<tr>
<td>0.314 rad/s, Rapid Fatigue</td>
<td>A 0.40457785</td>
</tr>
<tr>
<td>0.314 rad/s, 20rep Fatigue</td>
<td>A 0.38753537</td>
</tr>
<tr>
<td>0.175 rad/s</td>
<td>A 0.38660304</td>
</tr>
<tr>
<td>0.314 rad/s</td>
<td>A 0.38650867</td>
</tr>
</tbody>
</table>

Levels not connected by same letter are significantly different.
Figures 115. LS Means Plot for Simulation Scenario

Table 119. LSMeans Differences Student’s t for Phase

<table>
<thead>
<tr>
<th>Level</th>
<th>Least Sq Mean</th>
</tr>
</thead>
<tbody>
<tr>
<td>Eccentric</td>
<td>A 0.48087852</td>
</tr>
<tr>
<td>Concentric</td>
<td>B 0.35070442</td>
</tr>
</tbody>
</table>

Levels not connected by same letter are significantly different.

Figure 116. LS Means Plot for Phase
Table I20. LSMeans Differences Tukey HSD for Interaction between Simulation Scenario and Phase

\[ \alpha=0.050 \quad Q=3.56554 \]

<table>
<thead>
<tr>
<th>Level</th>
<th>Least Sq Mean</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.524 rad/s, Eccentric</td>
<td>A 0.61636144</td>
</tr>
<tr>
<td>0.314 rad/s, Rapid Fatigue, Eccentric</td>
<td>A 0.55153472</td>
</tr>
<tr>
<td>0.314 rad/s, Eccentric</td>
<td>A B 0.48499021</td>
</tr>
<tr>
<td>0.314 rad/s, Erratic-pulse, Eccentric</td>
<td>A B C 0.47922983</td>
</tr>
<tr>
<td>0.314 rad/s, 40rep Fatigue, Eccentric</td>
<td>A B C 0.47704708</td>
</tr>
<tr>
<td>0.314 rad/s, 20rep Fatigue, Eccentric</td>
<td>A B C D 0.45439390</td>
</tr>
<tr>
<td>0.314 rad/s, 40rep Fatigue, Concentric</td>
<td>A B C D E 0.44313404</td>
</tr>
<tr>
<td>0.314 rad/s, Erratic-sine wave, Concentric</td>
<td>A B C D E 0.43679156</td>
</tr>
<tr>
<td>0.314 rad/s, Erratic-pulse, Concentric</td>
<td>A B C D E F 0.41291130</td>
</tr>
<tr>
<td>0.175 rad/s, Eccentric</td>
<td>A B C D E F 0.39202921</td>
</tr>
<tr>
<td>0.314 rad/s, Erratic-sine wave, Eccentric</td>
<td>A B C D E F 0.39144201</td>
</tr>
<tr>
<td>0.175 rad/s, Concentric</td>
<td>A B C D E F 0.38117687</td>
</tr>
<tr>
<td>0.314 rad/s, 20rep Fatigue, Concentric</td>
<td>A B C D E F 0.32067864</td>
</tr>
<tr>
<td>0.314 rad/s, Concentric</td>
<td>A B C D E F 0.28802714</td>
</tr>
<tr>
<td>0.524 rad/s, Concentric</td>
<td>A B C D E F 0.26529667</td>
</tr>
<tr>
<td>0.314 rad/s, Rapid Fatigue, Concentric</td>
<td>A B C D E F 0.25762097</td>
</tr>
</tbody>
</table>

Levels not connected by same letter are significantly different.
APPENDIX J: ADDITIONAL CONTROL OPTIONS

Several control methods were investigated for the system comprised of a PMA interacting with a simulated neuromuscular actuator. Several of these were adaptive techniques. Adaptive techniques were evaluated because of their potential ability to properly adjust in reaction to slow changes in the environment or system operation or discrepancies between actual and predicted human operator behavior. Some control techniques were developed to the point where they could be tested using Simulink models within MATLAB 7.11.0 (R2010b).

Model reference adaptive control (MRAC) and adaptive pole placement control (APPC) were each applied to a simplified three-element phenomenological model (with no functions of pressure, no use of characterization data, and inertial effects assumed to be negligible) as shown in Eq. (J1). With these assumptions, one is able to change the plant closed loop transfer function that is not strictly positive real (no zeros and relative degree of 2 as seen in Eq. (J2)) into a plant closed loop transfer function with a relative degree of one as seen in Eq. (J3).

\[
B_{PMA} \dot{x} + K_{PMA} x = \Delta F 
\]  

\[
x = \frac{1/m \Delta F}{p^2 + B_{PMA}/m p + K_{PMA}/m} 
\]  

\[
x = \frac{1/B_{PMA}}{p + K_{PMA}/B_{PMA}} \Delta F 
\]  

The intent was to use the developed adaptive control law to determine the change in force needed to produce the desired PMA displacement. However, this approach did not lend itself to
incorporation of the simulated neuromuscular actuator and HQDS model (whose behavior is
defined nonlinearly). For haptic control, the force resistance command should originate from
simulated neuromuscular actuator and HQDS model system outputs. Also, the control laws based
on this plant would operate using PMA position error. PMA position error is not the main
parameter of interest, as the objective is to track motor shaft position and velocity.

A form of adaptive pole placement, direct proportional plus integral control, as defined in
(221), was used to replace the inverse three-element phenomenological model within the PMC.
The simplified plant was used to develop a control law for $\Delta F$ (see Eq. (J4) through (J6)). A
pressure command was calculated at each loop iteration using Eq. (J7), based on the pressure-
dependent characterization of $F_{ce}$ and the force resistance command, $F_R$. Different adaptation
gains were selected through trial and error. $B_{PMA}$ was set to 2.9 during contraction and 1.87 during
relaxation (middle of the relaxation damping value range) for the simulation studies. Simulation
results showed that this method was not more effective than the pressure-dependent characterized
parameters within the inverse model.

\[
\Delta F = -\hat{k}_1 e - \gamma_2 \text{sgn}\left(\frac{1}{B_{PMA}} \int_0^t e(\tau) d\tau\right)
\quad (J4)
\]

\[
\hat{k}_1 = \gamma_1 e^2 \text{sgn}\left(\frac{1}{B_{PMA}}\right)
\quad (J5)
\]

\[
e = x - x_d
\quad (J6)
\]

\[
P = \frac{\Delta F + F_R - 44.6}{2.91}
\quad (J7)
\]

Parameter uncertainty in a linear or nonlinear system can be reduced using parameter
estimation. The values of the parameters are estimated based on the measurements of the input
and output signals of the system. A parameter estimation method is used for adaptation in the
TMC part of the control system. Parameter estimation applied to the PMC was also tested in
simulation by adapting the $B_{PMA}$ and $K_{PMA}$ terms of the simplified three-element
phenomenological model described above. Composite adaption, which uses tracking error and prediction error, (215) was applied to the model within the PMC. Simulation studies showed that parameter estimation of the $B_{PMA}$ and $K_{PMA}$ terms did not perform as well as the characterized inverse model.

PID controls with fixed, nonlinear, and adaptive gains were tested both in simulations and within the DTS. PID methods used to adjust the force command term from the TMC were effective at certain parts of the task and negatively impacted behavior during other parts of the task due to increased control activity. PID methods used to adjust the desired displacement term based on displacement feedback error were not shown to be effective.

Sliding mode control applied to the three-element phenomenological model for use within the PMC was simulated, as was sliding mode control developed to replace both the TMC and PMC functionality. The nonlinear adaptive control method based on sliding mode control used within the TMC was found to be most effective, and was therefore implemented within the PMA Dynamic Test Station.
APPENDIX K: DESIGN IDEA FOR PMA PRODUCED RESISTANCE STRENGTH TRAINING DEVICE

Figure K1. Starting and ending position (near 90° of knee flexion)

Cable spool bearing and fixture assembly (potential location for position and torque sensors)
Cable Spool
PMA
Lever arm assembly (mechanism which transfers torque between human and device)

PMA air inlet and PPR fixture assembly area (also provides surface area to fix device to ground)

Lever arm is adjustable to accommodate different human operators. Straps would be necessary to secure upper and lower leg(s) in the correction positions, as well as other parts of the body.

Figure K2. Isometric hold position (near 5° of knee flexion)

Orientation of PMAs can be changed from vertical to horizontal with the same force transfer mechanism. The vertical orientation saves space because vertical fixturing and stabilization is necessary with either orientation.
REFERENCES


