Evaluation of an Actuated Wrist Orthosis for Use in Assistive Upper Extremity Rehabilitation

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EVALUATION OF AN ACTUATED WRIST ORTHOSIS FOR USE IN ASSISTIVE UPPER EXTREMITY REHABILITATION

by

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Cerebral palsy (CP) is a neurological condition caused by damage to motor control centers of the brain. This leads to physical and cognitive deficiencies that can reduce an individual’s quality of life. Specifically, motor deficiencies of the upper extremity can make it difficult for an individual to complete everyday tasks, including eating, drinking, getting dressed, or combing their hair. Physical therapy, involving repetitive tasks, has been shown to be effective in training normal motion of the limb by invoking the neuroplasticity of the brain and its ability to adapt in order to facilitate motor learning. Creating a device for use with Activities of Daily Living (ADLs) provides an additional tool for task-based therapy with the goal of improving functional outcome.

A custom wrist orthotic has been designed and developed that assists flexion/extension of the wrist and rotation of the forearm, while leaving the hand open for the grasp and manipulation of objects. Actuated joints are driven with geared brushless DC motors on a lightweight, exoskeleton frame coupled to a passive arm that tracks positional changes within the task space. Control of actuation is accomplished with a custom mapping strategy, created from nominal movement profiles for 5 ADLs collected from healthy subjects. A simple relationship was created between position within the workspace and orientation necessary for task completion to determine needed assistance.

Validation of the design subjected the device to three different conditions, including robot guidance of the limb, co-contraction of the forearm, and the use of alternate approaches to complete the task. Co-contraction and alternate approach conditions were used to simulate characteristics of impaired subjects, including rigidity, spasticity, and lack of muscle control. Robot guidance achieved an average orientation error of 5° or less in at least 75% of iterations across all tasks, while co-contraction and alternate approach was able to do this in flexion/extension, but saw much higher errors in forearm rotation. Causes for performance deficiencies were attributed to lack of torque bandwidth at the motor and response delay due to signal filtering, aspects that will be corrected in the next iteration of the design.
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1 INTRODUCTION

The goal of the proposed research was to design, build, and evaluate a lightweight, low-clearance wrist orthosis capable of actively assisting flexion, extension, pronation, and supination for upper extremity rehabilitation. The target demographic for the device includes children ages 7 to 14 with hemiplegic Cerebral Palsy, for whom disease-induced motor deficiencies can make everyday living difficult. However, the flexibility of the device design extends its application to any group with similar impairments (stroke, etc.), where rehabilitation of wrist movement is useful in overcoming neurological damage.

The need for new tools in physical and occupational therapy has become more apparent due to the inconsistent functional improvement seen with current devices and therapies, where therapists use much of their time providing assistance to the patient. The need for a therapist to apply active assistance can introduce confounding variables such as delay of response between patient initiation and therapist assistance. Assistance protocols based on the application of precise forces at specific times can also be difficult to implement due to the often qualitative nature of the forces applied by a therapist. Regardless, functional improvement has been seen with extensive participation in physical therapy ([1], [2], [3], [4], [5]), though the degree of rehabilitation can vary widely across subjects. The use of robot-assisted physical therapy systems, such as the one proposed here, can provide faster and more precise assist responses due to direct feedback obtained from force and position sensors ([3], [6], [7], [8]). Robotic systems can also provide quantitative performance data that can be used to compare and contrast the effectiveness of different therapies ([6], [7], [8], [9], [10]).

1.1 Cerebral Palsy Background

Cerebral palsy (CP) is a neurological condition caused by damage to motor control centers of the brain. CP is the most common cause of severe physical disability in childhood,
affecting approximately 3 in 1000 children nationally and from 1.4 to 4 in 1000 worldwide, based on several studies referenced by the CDC ([11], [12]). Characteristic signs of CP include spasticity, movement disorders, muscle weakness, ataxia, and rigidity [11]. In addition to motor impairments, cognitive disorders, seizures, sensory impairment of the arms, hydrocephalus, autonomic dysfunction, impairment of visual perception, and learning disabilities have also been linked to CP [11]. Difficulties holding and detecting objects often develop as a result of diminished somatosensation and proprioception [11]. These impairments and disabilities result from neurological damage that occurs prenatally and/or during early childhood.

Physically, CP manifests as static lesions on the brain in which injury to upper motor neurons decreases cortical input to the reticulospinal and corticospinal tracts [11]. Deficits in sensation, sensorimotor processing, and coordinated movement in multiple muscle groups can form, depending on the lesion location [1]. Flexor synergies may also occur, involving shoulder internal

Figure 1: Example of a child with flexor synergy and left hemiparesis [11]
rotation, elbow flexion, forearm pronation, wrist flexion, finger flexion, and thumb-in-palm, as demonstrated in Figure 1. This can make opposing motions (flexion/extension, pronation/supination) about individual joints difficult to perform.

At the wrist, loss in range and/or types of movement due to CP results in functional deficits that impair subjects’ ability to perform daily tasks such as eating with a spoon, drinking from a cup, or combing their hair. Normal wrist movement has three degrees of freedom. Flexion/extension movements of the hand up and down, respectively, are mediated by rotation of the radiocarpal and midcarpal joints located between the eight carpal bones of the wrist [13]. The full range of movement is typically 130°-140°, with 65°-80° associated with flexion and 55°-70° with extension [13]. Inward and outward (relative to the body) rotation at the wrist is generated in the forearm from the proximal and distal radioulnar joints, and is referred to as pronation and supination respectively [14]. The zero reference position used to characterize pronation/supination is referred to as the “thumb up” position. Relative to the zero reference, pronation and supination of up to 75° and 85° can be achieved [14]. Greater range may be possible as a result of contributions from the elbow and upper arm. From this 160° range of total motion, 100° is classified as “functional”, evenly distributed between pronation and supination [14].

1.1.1 Classifications of CP

Due to the variation in functional deficits resulting from CP, different classifications are used to distinguish between the severity and topographical distribution of impairment. Levels of impairment range from mild to severe, based on the number of independent activities the child can perform without assistance [15]. Affected body parts vary, and are grouped functionally as either plegia (paralysis), or paresis (weakening) [15]. The extent across muscle groups can be limited to one affected limb (monoparesis or monoplegia) or can be extensive, involving all extremities (quadriparesis or quadriplegia) [15].
Children with hemiplegic CP were targeted as a desirable group for initial development and assessment of a robotic wrist therapy system due to the potential improvement it could provide to their ability to complete bimanual tasks. Hemiplegics often become “one-handed experts” with their normal functioning arm while the impaired arm acts only to stabilize or assist [16]. In a study by Mackey et al. [16], both hemiplegic and normally functioning children were asked to complete various tasks including touching hand to head, touching hand to mouth, and reaching out to touch a stationary object. Their results showed that hemiplegic children achieved significantly less supination (49°±39°; control, 77°±22°) and greater forward trunk flexion (43°±14°; control, 29°±16°) in the hand to head task, less supination and greater forward trunk flexion (42°±13°; control, 28°±15°) and shoulder flexion in the hand to mouth task, and significantly less elbow extension (24°±18°; p<.01) in the reach task. Interestingly, they also found that the movement of the normally functioning limb tended to match the movement of the hemiplegic limb during bilateral tasks. This could be a result of the brain attempting to maintain the symmetry of bilateral movements by slowing the normal arm down. In addition to reduced range of motion, time to complete tasks was greater for hemiplegic children. This could be attributed to reduced strength, fatigue, and lack of confidence in the movement, which cause the child to move slower in order to increase accuracy.

1.2 Current Therapies

With no cure available for Cerebral palsy, current treatment options focus around managing the condition and improving quality of life. Treatments are either pharmacologically or therapeutically based. Drug-based solutions, like the injection of the Botulinum toxin (Botox), can help reduce spasticity in the limbs [11]. Less invasive, unlicensed oral medications are also available and include anticonvulsants, which have movement-modifying effects, and antisdystonics [17]. These need to be used with caution, however, as many produce significant side effects and have shown limited evidence of their effectiveness [17]. Physical therapy has
shown the most improvement of motor skills through intensive upper-extremity training of bimanual performance, strength training, hippotherapy on muscle symmetry, and balance training for reactive balance [2]. Constraint induced movement therapy (CIMT) has been one of the most successful strategies at improving motor skills, especially in children [2]. It features “massed” or repetitive practice of tasks, and has shown retention rates after therapy of at least 6 months [1]. High levels of attention and motivation are required to make this therapy successful, which can be more challenging to achieve with children [1]. There are several questions that have yet to be answered with intense upper limb training paradigms, including optimal dosing strategy, optimal age to begin therapy, and possible adverse effects on the less affected extremity and associated cortical pathways [2]. Strength training has been studied in much greater detail, and has been shown to provide significant gains in a short period of time [2]. Continuous participation may be necessary to maintain the benefits, and it is unclear whether limb loading or functional practice provides the gain [2]. Regardless, the use of these therapy strategies shows a trend toward the use of task or function-based training to achieve the highest functional gain [2].

Surgical interventions have also shown promising results. For children with Cerebral palsy, wrist deformity occurs in pronation, flexion, and ulnar deviation, making it difficult to perform the opposing motions (supination, extension, radial deviation respectively) [18]. This can lead to issues with bimanual movements, and grasp and release functions [18]. The primary approach to correct these deformities involves transfer of a wrist flexor (flexor carpi ulnaris, FCU) to a wrist extensor (extensor carpi radialis brevis, ECRB) [18]. The tendon transfer removes the spastic FCU that over-promotes flexion and places it at the ECRB to help augment the patient’s weak wrist extensors. By performing this transfer over the ulna and not through the interosseous membrane, supination can be improved in addition to extension [18]. As a result of this surgery, however, certain movement limitations present themselves. The child is unable to perform passive wrist flexion or resistive wrist extension for 3 months, and is permanently unable to passively flex the wrist greater than 45° [18].
1.3 Neuroplasticity

Neuroplasticity refers to the ability of the brain and Central Nervous System (CNS) to reorganize in response to stimuli. This reorganization is an intrinsic property allowing adaptation to environmental changes, physiologic modifications, and experiences [19]. In a crossover study conducted by Byl et al. [20], stroke patients were asked to participate in 8-week programs of fine motor training and sensory discriminative training. The program for motor training featured stress-free hand strategies for functional activities, fine motor task practice, aerobic, strengthening and flexibility exercises, and mental rehearsal to reinforce motor learning. Data, processed categorically by fine motor skills, sensory discrimination, musculoskeletal performance, and functional independence, showed statistically significant gains from pre-treatment to post-treatment. Sixteen of the 18 subjects showed greater than 20% improvement in functional independence and upper limb function, as determined by culminated scores from several tests including the Wolf Motor Function Test, California Functional Evaluation, and tests for Graphesthesia, Kinesthesia, and Stereognosis. Ten subjects were evaluated again three months post-treatment and showed continued positive improvement in upper extremity fine motor skills, sensory discrimination, and functional independence. This maintained improvement shows the potential benefits that a training paradigm focused on neuroplasticity can provide for patients.

Functional organization of the primary motor cortex can also change in response to practice ( [20], [21]). Tyc and colleagues analyzed motor evoked potential (MEPs) responses to transcranial magnetic stimulation (TMS) in volleyball players and runners who were asked to aim at a target while sitting or perform a static wrist extension [21]. Cortical activity corresponding to the proximal medial deltoid and distal extensor carpi radialis were much larger in size for the volleyball players than the runners. This is consistent with the volleyball player’s regular use of their shoulders and upper extremity to strike a ball, while runners use their lower extremity most to propel them forward. Across groups, the amount of cortical activity was greater in the
hemisphere contralateral to the dominant arm. This study provides evidence that practice and repetitive motion have the ability to induce changes in the cortex.

It has been shown that motor control for everyday actions is task-specific. Bootsma et al. noted that the kinematics of reaching movements of objects with smaller widths coincided with a longer deceleration phase than a whole hand grasp. Object width was found to directly affect hand aperture, which defines the strategy used to grasp an object [22]. This allowed the individual to better account for potential spatial error, since the margin for error with a narrower hand

Table 1: Principles of experience-driven plasticity

<table>
<thead>
<tr>
<th>Principle</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>1. Use It or Lose It</td>
<td>Failure to drive specific brain functions can lead to functional degradation.</td>
</tr>
<tr>
<td>2. Use It and Improve It</td>
<td>Training that drives a specific brain function can lead to an enhancement of that function.</td>
</tr>
<tr>
<td>3. Specificity</td>
<td>The nature of the training experience dictates the nature of plasticity.</td>
</tr>
<tr>
<td>5. Intensity Matters</td>
<td>Induction of plasticity requires sufficient training intensity.</td>
</tr>
<tr>
<td>6. Time Matters</td>
<td>Different forms of plasticity occur at different times during training.</td>
</tr>
<tr>
<td>7. Salience Matters</td>
<td>The training experience must be sufficiently salient to induce plasticity.</td>
</tr>
<tr>
<td>8. Age Matters</td>
<td>Training-induced plasticity occurs more readily in younger brains.</td>
</tr>
<tr>
<td>9. Transference</td>
<td>Plasticity in response to one training experience can enhance the acquisition of similar behaviors.</td>
</tr>
<tr>
<td>10. Interference</td>
<td>Plasticity in response to one experience can interfere with the acquisition of other behaviors.</td>
</tr>
</tbody>
</table>
aperture was much smaller. Rizzolatti et al. showed neurophysiological evidence for selective
cortical activation for different types of grasp [23]. Bearing this in mind, Kleim and Jones
outlined a set of requirements for rehabilitation based on several principles derived from
neuroscience research (Table 1) that are geared toward inducing experience-dependent plasticity
during task-based therapy. These include using exercises resembling real life tasks, performing
sufficient repetitions, task-specificity geared toward functional outcome, and providing feedback
that induces an external focus of attention ([19], [24]).

The adaptive qualities of the brain are important to understand in order to develop
effective rehab strategies. It is common for individuals with brain damage to develop
compensatory behavioral strategies to perform daily activities in the presence of lost function
[24]. These strategies are seen as key drivers to what is considered a “typical” response to brain
damage. The ability of the brain to compensate can have a wide range of impacts when
rehabilitation training is introduced. Many times the behavioral changes can be adaptive, and
contribute greatly to functional outcome, while other times they can be maladaptive and interfere
with improvements in function [24].

With plasticity already a key consideration in task-based therapy, the integration of
robotic systems provides additional support intended to increase therapeutic outcome. Though it
is not fully understood to what extent robot-mediated therapy contributes to neurological
recovery, there has been evidence to suggest robot-mediated neuroplasticity [25]. Examples
include the modulation of contralateral alpha and beta frequencies from robot-assisted wrist
movement in cortical areas similar to those engaged during voluntary movement ([25], [26]), and
substantial overlap of neurological activity in elbow flexion/extension from voluntary movement
and torque-motor driven conditions ([25], [27]). Most importantly however, robot-assisted
therapy is an objective and reliable means of monitoring patient progress, allowing quantitative
evaluation of motor performance and tracking of results [3]. Training can also be delivered at a
much higher dosage and intensity [3]. Robotic systems are capable of providing rapid feedback, and corrective torques that can be used to modify an individual’s maladapted movement strategy. These features can help make the strategy promoted by the robotic system become more prominent. The ability of robotic therapy systems to provide faster feedback and more accurate assistance ties directly into neuroplastic principles involving feedback, repetition, and experience. These characteristics result in a powerful tool that can potentially change the clinical expectations of physical rehabilitation.
2 ROBOTICS AND POWERED DEVICES IN THERAPY

The use of robot-mediated therapy (RMT) to improve functional performance in the upper extremity for children with CP has begun to be explored, and includes many features conducive to positive functional outcome. While muscle strengthening and exercise-based therapy have been used to successfully enhance motor recovery, the key to functional recovery stems from neuroplasticity [3]. The ability of the cerebral cortex to reorganize can be enhanced through experience and behavioral demand [3]. Increasing the dosage and intensity of therapy help this reorganization by engaging cortical plasticity in the brain [3]. RMTs are able to facilitate this by providing visual and force feedback, interactively evaluating the patient’s movements, and providing corrective assistance [3].

2.1 RMTs in CP Rehabilitation

Evidence for the efficacy of new technologies in rehabilitation continues to be scarce, and it remains to be seen which indications and applications lead to the best results [4]. Conventional upper extremity therapies, which can be labor intensive, have led to the development and evaluation of a limited number of robotic systems on children with CP [4]. The InMotion2, developed by Interactive Motion Technologies Inc., was applied to 12 children between 4 and 12 years of age with upper limb hemiplegia caused by acquired brain injury or CP [4]. Testing was performed in conjunction with the child’s conventional therapy, and involved one-hour sessions, twice a week for eight weeks, performing repetitive, goal-directed, planar reaching movements. The robot consisted of a workstation with computer monitor for visual feedback, and 2 degree of freedom (DOF) arm with handle that the patient held and manipulated [5]. The monitor displayed visual targets that the patient would attempt to reach by manipulating their arm, using primarily the shoulder and elbow joints. If the patient could not properly grasp the handle, an assistive mitt or strap was used to comfortably secure the hand. Based on the child’s motor abilities, the robot was able to adapt and alter the amount of assistance it provided. Positive changes were observed
in assessments of the Quality of Upper Extremity Skills test (QUEST) and Fugl-Meyer scores following the conclusion of the study, and correlated with increased general use of the impaired arm based on questionnaires submitted by the parents.

A separate study, using the New Jersey Institute of Technology Robot-Assisted Virtual Rehabilitation System, evaluated its influence on functional outcomes for two subjects (7 year old girl, 10 year old boy), with hemiplegic CP [1]. The system used the commercially available Haptic Master robot (Moog™), outfitted with a custom 6 DOF ring gimbal to measure forearm rotation in addition to shoulder and elbow movements [4]. Force exerted by the user, and end-point position were measured in three dimensions to generate a reactive motion by the robot that acted as the interface between the subject and the virtual environment [1]. Activities focused on reaching and point-to-point movements were created in a virtual reality (VR) environment and displayed on a screen in front of the child to evaluate the speed and accuracy of limb movement. Objects within the virtual environment provided real-time force feedback when the virtual cursor made contact with them, promoting movement along a desired path. Resistance provided by the robot could be adjusted based on the level of impairment. The two subjects, both higher functioning, performed 9 total hours of training in 60 minute sessions, and showed improvement in kinematic and performance measures. These measures included upper extremity active range of motion and strength, upper extremity movement quality using the Melbourne Assessment of Unilateral Upper Limb Function, hand movement speed, and movement duration [1]. Subject 1 showed greatest gains in coordination and efficiency of movement while subject 2 showed greatest improvement in range of motion in the shoulder and elbow [1]. Both children also showed no problems with performing therapy tasks, as the virtual, game-oriented environment kept them interested and engaged [1]. One primary limitation of this study was the use of higher functioning subjects only. This limited full testing of the benefits of the system’s robotic assistance capabilities, and together with expanding therapy times beyond 60 minutes, was
identified as a primary objective for future study [1]. A subsequent study was performed on the
same system using 9 subjects with CP participating in nine, 60-minute sessions. The results of
this study showed improvement in the Melbourne Assessment, and several measurements of
reaching kinematics [28].

Evidence supporting the long-term efficacy of new technologies in rehabilitation, like
robotics, remains scarce [4]. For the few studies that have been performed, sample sizes have
been small resulting in mixed results. It is yet to be determined which combination of parameters
(duration, frequency, intensity) provide the best levels of functional improvement, and how those
results are affected in conjunction with other therapies (drugs, surgery, etc.) [4].

2.2 Other Areas of Rehabilitation Utilizing RMTs

Though robotic intervention in CP has not been widely studied, extensive work has been
done to develop robotic systems for improving functional gain following stroke. These robotic
devices have focused on characteristics such as being lightweight, compact, simple, and rigid, and
have featured two typical configurations. Most favor a serial configuration, where motors power
each joint with links separating them. The RiceWrist, however, introduced a unique parallel
configuration that allows for a more compact and robust structure [9]. Serial configurations are
advantageous as they can lead to simpler mechanical structures for manufacture and reduce the
number of potential failure points [7]. However, they often don’t produce the same structural
rigidity as parallel configurations [7]. The decision to use a parallel configuration in the
RiceWrist was motivated primarily by the compactness of the mechanism, with added benefits of
higher torque output, higher stiffness and decreased inertia relative to a comparable serial
mechanism [9]. The full exoskeleton uses a serial-in-parallel configuration to accomplish both
positioning and orienting the wrist, showing a potential benefit in integrating the two
configurations [9].
Actuation is typically accomplished by one of two means. Brushless DC rotary motors are commonly used with various coupling options to drive the joint of interest. The simplest involve direct drive systems, such as the ExoRob, where the motor shaft lines up with the joint axis of rotation [6]. One of the most popular choices has been to use cable and pulleys to actuate joint movement ([7], [8]). The RiceWrist orthosis uses a capstan drive system, similar to a rack and pinion configuration, using a cable and pulleys to move a prismatic slide with the rotation of the motor [9]. The second means of actuation comes from pneumatic cylinders powered with a motorized compressor or compressed air source, used as a linear actuator [10]. As described by Allington et.al, who created the Supinator Extender (SUE) robot, the use of electric motors requires the addition of gearheads to output the needed torque, which can negatively affect the backdriveability and bandwidth of the system due to the high impedance introduced [10].

Backdriveability is the ability for interactive transmission of force between input axis and output axis [29]. Achieving high backdriveability is only attained by reducing the friction of power transmission considerably [29]. Rehabilitation robotics applications require high torque and low velocity outputs, resulting in the employment of motor-gearbox combinations, which can possess large levels of friction due to the gearbox. In this case, backdriveability is defined by the torque the user needs to apply to the robotic joint in order to perform a user-driven movement. In a case where backdriveability is perfectly optimized, this torque is reduced to zero [30]. Pneumatic systems can typically promote backdriveability by reducing system bandwidth through low-friction seals and lubricating the piston walls [10]. Radial bearings have also been a common addition, independent of the actuators, to help minimize friction and improve backdriveability. Typically, the bearings are implemented at the anatomical joint where motion introduced by the user will not be impeded.

Most systems use position and force feedback to control the assistive actions of the device. The RiceWrist uses three separate controllers, one for joint space position control, one for
task-based position control, and one for task space impedance control [9]. The impedance controller creates “virtual walls”, or boundaries to keep the user within a close range to the desired trajectory [9]. Both the ExoRob and SUE employ a “sliding mode controller” which takes advantage of the nonlinear and discontinuous natures of the dynamic system to provide robust control and trajectory tracking performance ([6], [10]). The control scheme also fits well in actuation systems that use on/off switching such as the SUE, which uses the controller to command the opening and closing of the solenoid valves that regulate the pressure within the pneumatic cylinders [10].

Across the literature characterizing robotic systems for rehabilitation ([1], [4], [5], [6], [7], [8], [9], [10]), several common features were identified and taken into account. The structure should remain compact and lightweight while being robust and strong enough to support an individual limb. Training experiences should also be provided that are relevant to the functional improvement being sought, keep the patient motivated to maintain attention, and engage plasticity mechanisms in the brain [3]. These considerations led to the creation of a rehabilitation system for the wrist that incorporates performance of everyday tasks within a framework that promotes normal movement.
3 DEVICE DESIGN AND OPERATION

The developed rehabilitation system consists of a 3 DOF passive arm and 4 DOF wrist orthotic (Figure 2). The combined effect of both components allows the subject to position and orient the arm and wrist within the workspace while performing therapy. Unlike the robotic systems presented previously, the design interfaces with the user solely at the distal end of the limb, giving the user more movement freedom. The hand is also left free to grasp and manipulate objects.

3.1 Design Requirements

Device design requirements were compiled from therapist feedback, understanding of current therapy practices, and literature of previously developed robotic therapy systems (Table 2). The list included many of the characteristics seen in the devices reviewed in Section 2, and features that focus on the use of the device in a task-based environment. The list breaks down the requirements based on the key interactions that take place within a rehabilitation setting. In
addition to the device, interactions with the human subject, the tasks they perform, and the environment in which they are performing the tasks were considered. These interactions are based on the HAAT (Human Activity/Assistive Technology) model, as defined by Cook and Hussey which was created to show how assistive technology could be incorporated into the basic human performance model [31]. These conditions interact when the device is used (Figure 3), and share equal consideration during device design. The requirements highlight the performance, maintenance, and modularity necessary to accommodate a pediatric, physically impaired population. The target age range and the different growth rates of children make it necessary to accommodate a range of wrist, hand, and forearm sizes, while the presence of spasticity and irregular limb orientations requires a strong and robust structure. Since the intended use of the device is for task-based therapy, it was important that the hand be allowed to remain open in order to grasp and manipulate objects. It was also important that the device be able to support tasks that promote wrist motion (pronation/supination and flexion/extension), and were fun and engaging.

Figure 3: Illustration of therapy interactions driving device requirements
<table>
<thead>
<tr>
<th>Interaction</th>
<th>Requirement</th>
</tr>
</thead>
<tbody>
<tr>
<td>Human</td>
<td>Thumb up neutral position</td>
</tr>
<tr>
<td></td>
<td>Impairment is not severe enough to make the upper extremity nonfunctional</td>
</tr>
<tr>
<td></td>
<td>Hemiplegic</td>
</tr>
<tr>
<td></td>
<td>6-15 years old</td>
</tr>
<tr>
<td>Human/Device</td>
<td>Range of motion – 115° flexion, extension (70° extension, 45° flexion), 150°</td>
</tr>
<tr>
<td></td>
<td>pronation/supination</td>
</tr>
<tr>
<td></td>
<td>Generated torque &gt; user applied torque at the wrist</td>
</tr>
<tr>
<td></td>
<td>Lightweight</td>
</tr>
<tr>
<td></td>
<td>Comfortable</td>
</tr>
<tr>
<td></td>
<td>Durable</td>
</tr>
<tr>
<td></td>
<td>Form-fitting</td>
</tr>
<tr>
<td></td>
<td>Diameter based on wrist size</td>
</tr>
<tr>
<td></td>
<td>Aesthetically pleasing</td>
</tr>
<tr>
<td></td>
<td>Size adjustable</td>
</tr>
<tr>
<td>Device</td>
<td>Backdriveable</td>
</tr>
<tr>
<td></td>
<td>Minimal damping to achieve fastest response time</td>
</tr>
<tr>
<td></td>
<td>2 actuated degrees of freedom (F/E, P/S)</td>
</tr>
<tr>
<td></td>
<td>One actuator for each DOF</td>
</tr>
<tr>
<td></td>
<td>Easy to clean</td>
</tr>
<tr>
<td>Device/Task</td>
<td>Does not inhibit use of the hand</td>
</tr>
<tr>
<td></td>
<td>Only acts when the task cannot be completed naturally</td>
</tr>
<tr>
<td>Task</td>
<td>Promote F/E, P/S movements</td>
</tr>
<tr>
<td></td>
<td>Reflect everyday activities</td>
</tr>
<tr>
<td>Human/Task</td>
<td>Engaging/fun</td>
</tr>
<tr>
<td></td>
<td>Challenging yet not impossible</td>
</tr>
<tr>
<td></td>
<td>Feedback provided</td>
</tr>
<tr>
<td>Environment</td>
<td>Does not restrict movement</td>
</tr>
</tbody>
</table>
3.2 Passive Arm Structure

The 3-DOF passive arm provides support and allows for 3-dimensional positioning of the wrist within the workspace. It employs a RRP (revolute-revolute-prismatic) configuration, creating a spherical volume in which the user is able to operate. Also known as a spherical, or polar, manipulator, this configuration allows the user to encompass a large workspace, and promotes linear motions. This fits well within task-based therapy, as objects are approached and grabbed most efficiently by using straight-line movement. The majority of the structure was 3D printed with a lightweight ABS plastic, while the extending prismatic joint was created from a lightweight aluminum slide.

Figure 4: Passive arm structure. (a) Base joint and bearing, allowing rotational motion parallel to the workspace (b) elevation joint, allowing up and down movement in workspace (c) prismatic slide, allowing translational movement away from the arm (d) full arm.

Rotation is initiated using a flange bearing with cast iron base (AST Bearing™ UCF210-32E) mounted to the edge of the workspace (Figure 4a). A cylindrical extension protruding from the bottom of the arm is sized to fit the opening of the bearing, and set screws are used to secure it.
in place. Two uprights extend upward from the bearing, providing the axis of rotation for the elevation joint. Structural support is provided in the form of two lattice-inspired pieces oriented perpendicularly to the uprights and placed between them. This maximizes strength while minimizing the amount of material needed and final overall weight.

Elevation is created from two arrow shaped pieces coupled through a pivot axis featuring a pair of plastic rotational bearings (Figure 4b). The volume created from these two pieces provides a space to attach and operate the prismatic slide, and run cabling for power and signal transmission to the orthosis. The back of the elevation joint holds a 5 kilogram counterweight used to offset the weight created when the orthosis is connected. The goal of the counterweight is to minimize the gravitational effects acting on the device, and promote natural movement. It is secured between two aluminum bars mounted to the back of the elevation joint, creating a gap large enough for the weight to fit between when oriented perpendicular to the bars. An additional aluminum piece is attached between the bars, and provides a support for the weight to sit atop of. Since the weight is slotted, this arrangement locks the weight in place.

The extension joint is comprised of two linear slides with a plastic standoff secured at one end of each (Figure 4c). These standoffs provide the mounting site for a pair of Teflon-surface bushings manufactured to fit within the grooves of the slide. The slides are interlocked with one another by running the bushings within the track at opposite ends. One slide is mounted within the open volume of the elevation joint, and the second slide moves relative to the first. At full extension, the two bushing standoffs make contact with one another, establishing a physical stop for maximum extension. At maximum retraction, the stationary bushing block makes contact with a mounting block attached to the end of the moving slide. The purpose of this mounting block is to provide a site where the orthosis can be attached and interface with the arm. To further decrease the coefficient of friction between the bushings and slide, a combination of white lithium grease and WD-40 was added within the grooves of the slides.
3.3 **Orthosis Structure**

The orthosis is a 4-DOF wrist manipuladum that allows orientation of the limb, and specifically supports motions of the wrist. Using a pitch-yaw-roll configuration, the orienting capabilities of the orthosis closely resemble those of the human limb, promoting natural joint movements. Only two of the specific movements making up wrist orientation are actuated. Roll (pronation/supination) and wrist flexion/extension are actively monitored and assisted, while yaw and pitch (elbow flexion/extension, wrist radial/ulnar deviation) remain passive. Since yaw movements for daily activities typically use less than 15° of ulnar deviation and 10° of radial deviation [13], the need to assist such subtle changes was viewed a secondary in the overall design. The orthosis itself was constructed from both 3-D printed ABS plastic and machined aluminum parts. Both materials were used to reduce weight where possible without compromising the mechanical strength in the load bearing elements of the system. The use of 3D printing, as opposed to machining, enabled the design of more complex geometries and decreased the duration of the design/test cycle. The orthosis was positioned below the interface to the passive arm such that limb orienting motions of the orthosis would occur above the wrist movement. The spatial separation relative to the telescoping axis minimizes interference between the wrist orthosis and the limb orienting mechanisms. This enables access to the environment beneath the orthosis and facilitates the free and natural movement of the user in any direction and/or orientation.

3.3.1 **Passive Arm Orientation**

Movement corresponding to yaw and pitch at the wrist occurs within a series of links and blocks outfitted with rotational bearings. A rectangular extension sits parallel to the end of the prismatic slide and is attached to the mounting block (Figure 5). A vertical hole aligned perpendicular to the slide and located at the end of the extension contains press fit rotational bearings. A cylindrical shaft with head, similar to the shape of a bolt, passes through the bearings
such that the underside of the head rests above the top bearing. This setup follows the “yaw” motion of the wrist. The shaft extends into a second block (yaw initiating block, see Figure 5) where it is secured by bolts perpendicular to the shaft.

![Figure 5: Passive (yaw/pitch) wrist orientation interface. The subassembly facilitates radial/ulnar deviation and elbow/shoulder rotation.](image)

From the yaw initiating block, two side support braces are bolted and extend down to the “pitch initiating block” containing an arced top to facilitate rotation. An axle is press fit into a hole aligned parallel to the prismatic slide. Each support brace has a press-fit rotational bearing to allow the axle of the pitch initiating block to follow the “pitch” motion of the wrist. Two T-shaped brackets placed perpendicular to the side braces link the yaw/pitch assembly to the rest of the orthosis.

The elements of the passive extension and yaw/pitch assembly are machined from aluminum to provide the durability and strength needed to transmit the forces exerted by the user’s arm on the orthosis and passive arm. Since the connection between the passive arm and orthosis is not a single, rigid piece, there is the potential for a bending moment to be created, and over time, the coupling between the parts could be weakened. Choosing a material that was at least as strong as the prismatic slide helps to prevent this issue.
3.3.2 Horseshoe Raceway and Trough for Pronation and Supination

Pronation and supination of the forearm is provided by a horseshoe-shaped frame and internally mounted turntable bearing coupled to a shallow forearm support trough (Figure 6). The horseshoe shape was selected to minimize the interference between the orthosis and tabletop during task-based therapy. An arc-shaped cable raceway support is bolted to the inner ring of the circular bearing and forearm support trough, coupling these two parts so that the forearm aligns with the rotation axis of the bearing. The raceway support serves as the attachment site for the cable drive used to provide active joint assist. See Figure 6 for component placement.

The turntable bearing (American Precision Group™ AT04535030 Quickmount® turntable bearing) is oriented vertically, and has a radial static load rating of 1.59 lbf. Because of this orientation, upward and downward forces applied by the user occur radially across the bearing. The greatest load applied to the orthosis occurs when the prismatic slide is fully extended. At full extension the orthosis requires at least 1.317 kg (2.9 lb) of upward force from the user to overcome the weight of the orthosis, as the counterweight alone only reduces this force requirement and does not eliminate it. Though the upward force provided by the user exceeds the radial load rating of the bearing, the applied load is not generated as a point force, but

**Figure 6:** Structure to facilitate pronation and supination. (Left) Front of gimbal and raceways for cable attachment, (Right) Back of gimbal with forearm trough.
a distributed force over the arc length of the cable raceway above the forearm support. With an arc length of 113.8 mm, only 0.025 lbf is seen per mm, well below the specification.

With exception of the bearing, the structural elements of the raceway and trough were manufactured through 3D printing using ABS plastic. This provided the flexibility to create parts, like the cable raceway, that would have been difficult to machine, and helped to cut down on the final orthosis weight.

### 3.3.3 Spindle and Brace for Flexion and Extension

A large aluminum spindle aligned with the anatomical axis of the wrist enables flexion and extension movements (see Appendix I for an exploded view). The center of the spindle contains a rotational bearing. A spacer rests against the bearing wheel, and extends an eighth of an inch beyond the spindle. This gap provides a space so that the spindle does not rub against the motor mounting plate while rotating. The spacer is locked in place with 3D printed rings on either side so that all spindle rotation occurs through the bearing. A resizable aluminum link extends away from the spindle and attaches to a thermoplastic brace oriented perpendicular to the link (Figure 7). The brace encloses around the hand below the metacarpal joints, allowing the fingers

![Figure 7: Structure to Facilitate Flexion and Extension. (Left) Side profile featuring drive spindle and hand brace, (Right) Front view featuring hand brace with load cell.](image)
to remain free. It is size adjustable using a Velcro strap to tighten around the hand. Two set screws placed on the top of the spindle were used to provide a mechanical limit to the range of motion of the orthosis through contact with the motor plate mounted directly above. The arrangement of the spindle and brace places the hand in a “thumb up” neutral position, which causes the lateral side of the hand to make contact with the link bridging the brace and spindle. This neutral position was desired as it supported the functional ranges of motion for the wrist and forearm equally for flexion, extension, pronation, and supination.

3.4 Actuation

Active assist for pronation/supination and flexion/extension is provided by two high speed, low torque motors (Maxon™ EC-22 Brushless DC Motors with Hall sensors). The decision to use high speed, low torque motors was made due to their compact size, measuring only 80 mm in length and 22 mm in diameter. This made the motors easy to mount and integrate within the infrastructure of the orthosis. By themselves, the motors are capable of putting out 19.6 mNm of torque, well below what was needed to provide assistance. To provide active assistance, the motors needed to be able to overcome resistances within the physical system, and the isometric and isokinetic torques applied by the subject’s wrist. It was expected that frequency of movement would be around 1 Hz, based on a study by Mann where wrist and forearm motion were measured using a triaxial electrogoniometer while subjects performed 24 different activities of daily living (ADLs). It was found that the predominant frequency component for these ADLs was around 1 Hz, and ¾ of all the spectral energy occurred at <5 Hz. [32]. Low frequency movement minimizes the angular acceleration at the wrist and forearm, keeping the required torques low. The torque specification for the system was determined based on limb torques for ADL performance versus upper limb rehabilitation devices as described in [7]. Based on the highest spec’d comparative device, as described in [9], required system torques were set at 5 Nm.
for pronation/supination and 4 Nm for flexion/extension to overcome the inertia of the system and provide active assistance.

Several options were available for transmission of torque from the motors to the degrees of freedom of the wrist and forearm. A direct drive system could eliminate the need for a transmission, but would require positioning along the axis of rotation and have to be much greater in size to provide the needed torque. This would impede positioning of the limb, and was not feasible for this reason. Gearing or belts could be incorporated and easily synchronized with the confidence that they would not slip during use. However, their rigid nature limits the number of configurations in which they can be applied, and the added inertia and friction due to the size and surface contact with the motor and joint could decrease the torque bandwidth at the joint. A cable and pulley drive system had the advantage of being flexible, making it possible to customize the transmission path, and was thin, minimizing added inertia to the joint. This allowed the mechanics of device movement to be designed around the anthropometric movement of the wrist and forearm while also positioning the motors away from the actuated joints.

To actively assist pronation/supination movements, the motor was positioned horizontally on top of the horseshoe frame, offset approximately 30° from center. The motor was secured in place with an aluminum plate that was part of the passive orientation assembly described previously. The motor/gearhead couples to a high-strength mechanical cable that acts as the drive to move the device joint. One end of the cable was threaded through a 3D printed pulley, and wound around a second pulley, bridging the gap between the first pulley and the raceway (Figure 6). The raceway creates a 160° arc and places range of motion limits on the joint. At each end of the raceway, a small plastic pulley rotates around a screw and winds the cable until taught. The screw is tightened from a nut on the back that ensures tension is maintained. This simple drive concept has proven to be very effective in creating a reliable means for ensuring that all motion,
whether active or passive, passes through the motor combination and allows rotation measurements to be taken.

The motor was reduced with a planetary gearhead (Maxon™ GP-22) providing a 24:1 reduction. The motor and intermediate pulleys do not provide any mechanical advantage, but the transition from the pulleys to the raceway provides a 4.4:1 mechanical advantage. During the development and build of the device, it was mistakenly perceived that the winding of the cable around the motor pulley created an additional advantage of 3:1 due to the diameter increase from the shaft to the pulley (4mm to 12mm). If the pulley had been positioned in a serial configuration to the shaft and additionally wrapped around the shaft, this would have held true. However, since the pulley was positioned in a parallel configuration over the shaft, this did not result in any added mechanical advantage, limiting the total reduction to 105.6:1 and potential torque output to 2.07 Nm.

Flexion and extension is actively assisted by similar means, with the motor combination housed in a semicircular extension of the trough to protect the user from the motor operation and prevent the motor from coming into contact with external elements. A 3D printed spindle was set on the motor shaft, oriented parallel to the main drive spindle, and a pair of high-strength mechanical cables were wound around the motor spindle, anchored by a pair of screws to the opposite end of the drive spindle (Figure 8). The screws were threaded directly into the spindle to pinch off the cable and maintain tension. To obtain the required torque, a planetary gearhead (Maxon™ GP-22 Planetary Gearhead) was added to the motor assembly with 53:1 gear ratio. An additional mechanical advantage of 3.21:1 was formed from the differing diameters of the spindles employed at the motor and at the joint. When combined with the gearhead, this resulted in an overall reduction of 170.13:1. As was the case with the pronation/supination joint, it was wrongly perceived that an additional advantage was obtained between the motor shaft and motor
spindle to provide further torque to the joint. Instead, the resulting continuous torque output of the joint was limited to 3.335 Nm.

3.5 Power

The orthosis was wired to a power system, supplying DC voltage to the motors and all peripheral devices including potentiometers, force sensors, motor controllers, and signal conditioning circuitry. All motor and sensor power is transmitted over shielded cables terminating at a distribution box mounted beneath the workspace table. From here, power splits to each individual device over smaller, more flexible cables.

3.5.1 Control Box Layout

The orthosis control box is a lightweight metal enclosure that features an on/off switch on the front face with a reset button and status light. A lattice of holes was drilled into the sides of the box to allow cross ventilation and heat release. A series of connector ports are located on the back panel to interface the robot and control computers (Figure 9). Interior components provide power, motor control, signal conditioning, and signal routing. AC power enters and feeds into separate
Figure 9: Physical control box layout; (Top Left) Top down view of components, (Top Right) Front view, (Bottom) Back view.

Figure 10: Control box layout and system communication flow.
24V and 36V DC power supplies. The outputs from the power supplies run to a positive and negative bus, where power is distributed to all other system components. These components include a power contactor, safety relay, motor controllers, voltage regulator, and signal conditioning circuitry. A shunt regulator is included between the 36V power supply and motor controllers to block transient noise and maintain a consistent, well-controlled input voltage. Figure 10 shows a schematic description of the connections made within the control box.

3.5.2 Dual Power Supplies

Two DC power supplies work in tandem to provide power to all devices in the system. A 36V supply (Emerson™ LCM600U) serves as the primary DC power supply, acting as the main power source for the motor controllers. The supply was selected due to its robust structure and current supply capabilities, and to match the input voltage needs of the motor controllers. All other power needs are handled by a 24V DC supply (XD1202405000), which powers the safety relay, power contactor, voltage regulator, and logic supply for the motor controller. These supplies also work in tandem with the safety relay to allow power to be cut to the motors without losing power to the rest of the system.

3.5.3 Motor Controllers

Each motor is wired to a separate controller that serves as a power source and communication relay for commands and rotation data. The controllers (Maxon™ EPOS2 70/10 Digital Positioning Controller) were tuned specifically to their respective motors, using the “Autotune” feature within the EPOS Studio software package. After providing the program with general information about the motor connected to it, the internal controlling mechanism was tuned for current, position, and velocity commands used to generate motor movement. Transient response plots were used as verification that the controllers were tuned appropriately for optimal performance. The controllers interface to the host computer by two communication lines; a USB
connection that allows changes to be made within the EPOS software, including changing operating modes, enabling controllers, and mapping CAN addresses for commands, and a CAN connection that reads and transmits commands to the main control system. CAN stands for Controller Area Network, and provides a means for high speed communication between multiple devices. This communication standard was selected due to its robustness and ability to facilitate microsecond feedback.

3.5.4 **Voltage Regulator**

A voltage regulator was implemented to provide a constant voltage output to all connected components regardless of the performance of the power supply. In the current configuration, the voltage regulator (LM723CJ) acts as an intermediary to distribute power from the 24V supply to all analog sensors. Each of the analog sensors requires a stable voltage to provide a constant operating bandwidth. The voltage regulator prevents fluctuations in the power supply from exceeding the voltage specifications of the sensors, which can lead to irreparable damage over time.

3.6 **Failure Protection and Safety Systems**

The intended user demographic, physically impaired children, makes device safety crucial. The orthosis incorporates several safety features designed to provide redundant protection against undesired (and potentially harmful) movements. A manual hardware stop was incorporated with the capability of cutting all power to the motor controllers with the press of a button. This button is made available to the user, and when pressed, triggers a safety relay (Allen Bradley™ MSR127TP) that opens an electromechanical switch within the power contactor (Schneider Electric™ LC1D12BD). All power from the 36V supply flows through the power contactor, so opening the switch interrupts the main power source for the motor controllers. A successful trigger of the safety relay is shown via a status light mounted to the front of the control
box. The system can be reset by returning the emergency stop button to its original position, and pressing the reset button on the control box located next to the status light.

In addition to the emergency button, physical stops were integrated into the mechanical structure of the actuated joints to prevent over-rotation of the wrist and forearm due to actuator malfunction. Set screws were placed in the drive spindle for flexion/extension, making contact with the flexion/extension motor mount plate before the hand or hand brace. The gimbal frame was closed at both ends to prevent rotation of the gimbal bearing beyond the normal range of motion for the forearm. These stops work to prevent injury to the subject by only allowing movement in a functional range of motion (~80° for both pronation and supination; ~70° for flexion, ~58° for extension). Motors were also positioned or physically shielded from the user in order to prevent accidental contact with the moving components during actuation.

In the control system, communication between the motors and their respective controllers are configured using EPOS Studio™. This includes the ability to toggle between enabled and disabled modes for the controllers, and a “quickstop” button that cancels the current command to the motors. This provides the therapist the ability to stop and disable the system if necessary. The built-in firmware of the motor controllers also monitors motor performance to detect various device errors including over/under voltage errors, over/under current errors, excessive temperature, software parameter errors, communication errors, and sensor errors. When one of these errors is detected, the controller reacts and moves into a fault state. At this point the drive is disabled and no command can issued until the error has been addressed, and the drive reset.

3.7 Signal Processing

The operating abilities of the orthosis are heavily dependent on the ability of the system to transmit movement information accurately in real-time. This requires close cooperation between hardware (sensors, amplifiers, analog filters) and software (digital filters, processing
functions) to quickly and efficiently process all signals into a usable form. In real-time robotic therapy applications, filters and cabling can create delays that make it difficult to provide accurate compensation. Efforts have been made to minimize the delays originating from both these sources while maintaining a clean signal. While it is not possible to eliminate all delays in the system, a certain level of delay can be acceptable as long as frequency of movement remains low. The digital low pass filter responsible for minimizing signal noise from the analog position sensors produces a group delay of 352 samples (0.352 sec) due to the small passband needed to adequately attenuate the noise. Changes in movement trajectory occurring faster than this delay could result in system responses to movements that are too slow, leading to instability.

3.7.1 Hardware

Analog and digital sensor signals are processed and recorded for control purposes and as part of the data collection. Each of the five joints on the passive arm was outfitted with analog single turn potentiometers (ETI Systems SP12b) coupled to the joint using hard rubber o-rings that create a friction fit between the potentiometer shaft and joint. A simple voltage divider applied to the output of the voltage regulator provides the 12V supply that establishes the signal bandwidth for the sensors. Shielded cables carry the signals to the control box where they interface to the data acquisition unit (NI-6251). Force sensors (Omega LC201-50) provide bidirectional (tension/compression) readings using a +/- 5V differential supply voltage. These signals travel to the analog inputs of the DAQ after being amplified and filtered.

The motor units on the orthosis are outfitted with 512 count, 3 channel digital encoders (Maxon Encoder MR). The encoders return measurements of wrist orientation based on changes read at the motor shaft through the direct coupling of the wrist joint to the motors. The signals are passed back to the motor controllers, and through a line receiver circuit before interfacing with the digital inputs of the data acquisition unit. The line receiver is necessary, as the encoder has a
built-in line driver to more reliably facilitate high rate data transmission and limit the impact of noise over the length of transmission.

To power the load cells, a DC-DC converter (Texas Instruments™ DCP022405DP) was used to provide a +/- 5V output with current capabilities up to 400 mA. Decoupling capacitors were placed at the input and the outputs to limit transmission of transient voltages to connected loads during power up/down.

To provide adequate amplification of the load cell signal to overcome high frequency common mode noise, a low power, gain programmable precision instrumentation amplifier (Linear Technologies™ LT1168) was connected to the output of each load cell. Based on the observed signal to noise ratios, a 150Ω resistor (applied across the amplifier gain pins) was used to provide a gain of 330, increasing the signal bandwidth to +/- 3.3V. At this gain, a Common Mode Rejection Ratio of 135-140 was expected, creating a 2-orders-of-magnitude difference between the differential signal and noise. Due to the small signal output from the sensors and the large distance between the sensors and the conditioning circuitry, a pair of 1st order RC lowpass filters were added at the inputs to the amplifier. These filters work to attenuate both differential mode and common mode noise. The filters were configured to create a differential mode cutoff frequency of 5.3 Hz with a time constant an order of magnitude larger than the common mode time constant, as recommended by the amplifier specs. Common mode cutoff frequency was configured at 10.6 kHz. Whether shielded or unshielded cabling is used to transmit the input signal, the cable can often act as an antenna for high frequency interference that can enter the input stage of the amplifier and cause an unwanted DC shift in the input offset voltage (RFI rectification). Using this simple solution greatly reduces these effects.
3.7.2 Software

All data from the orthosis is collected using a data acquisition card (National Instruments™ NI-6251) installed in a target computer that interfaces with the host computer over an Ethernet connection. This setup allows for data collection within the MATLAB xPC Target framework (Mathworks™). Using Simulink, a data collection model was created, linking the DAQ within software and allowing real-time processing and routing of the data. All analog signals were subjected to digital low pass filtering (minimum order FIR, 2 Hz cutoff frequency, 60 dB stop band attenuation at 5 Hz), due to the small bandwidth of the signals of interest (0-2 Hz) and the high sampling rate (1000 Hz). All analog position data is passed through a unit conversion function, and digital encoder counts are converted to radian and degree measurements by relating counts per turn and gear reduction to physical angular displacement of the joints. Force and encoder data are transmitted to the control algorithm directly, while potentiometer data is applied to the forward kinematic equations that define the position of the mechanical system to provide wrist position information during movement through the workspace.

3.7.3 Position Tracking Using Forward Kinematics

As is common practice in robotic applications, a forward kinematic approach was used to locate the Cartesian space location of the end effector using the joint space position of each degree of freedom. Rotation about each degree of freedom was defined about the z-axis, and a coordinate frame was established around this. Translation between coordinate frames takes place about either the x-axis or z-axis, and is defined by the link length connecting the two joints. Frame assignments are displayed in Figure 11. This framework is defined by the Denavit-Hartenberg convention for kinematic analysis, which greatly simplifies the positioning process for a robot with several degrees of freedom. Transitions between frames of reference for each degree of freedom are represented by homogeneous transformations that are created from the product of four basic transformations characterized using four parameters; link length ($a_i$), link
twist ($\alpha_i$), link offset ($d_i$), and joint angle ($\theta_i$). Link length is the measured distance along the x-axis between two frames, link twist is the rotation about the x-axis between two frames, link offset is the translation along the z-axis between two frames, and joint angle is the rotation about the z-axis between two frames. Joint angle or link offset, depending on whether it’s a rotational or prismatic joint, acts as the joint variable. This is the variable that changes with joint movement while the other three parameters remain constant. The full derivation of the kinematic equations can be found in Appendix A, and are summarized in Equations 1-9 below. An “s” or “c” with a subscript number represents the sine or cosine of the specified joint variable.

**Table 3:** D-H parameters for determining robot forward kinematics. Joint transitions relate to the reference frames described in Figure 11.

<table>
<thead>
<tr>
<th>Joint Transitions</th>
<th>$\theta_i$</th>
<th>$d_i$</th>
<th>$a_i$</th>
<th>$a_i$</th>
</tr>
</thead>
<tbody>
<tr>
<td>1-2</td>
<td>$\theta_1$</td>
<td>$d_1$</td>
<td>-</td>
<td>-90°</td>
</tr>
<tr>
<td>2-3</td>
<td>$\theta_2 - 90^\circ$</td>
<td>-</td>
<td>$a_2$</td>
<td>-90°</td>
</tr>
<tr>
<td>3-4</td>
<td>-90°</td>
<td>$d_3$</td>
<td>-</td>
<td>+90°</td>
</tr>
<tr>
<td>4-5</td>
<td>$\theta_4 - 90^\circ$</td>
<td>$d_4$</td>
<td>-</td>
<td>-90°</td>
</tr>
<tr>
<td>5-6</td>
<td>$\theta_5$</td>
<td>-</td>
<td>$a_5$</td>
<td>+90°</td>
</tr>
<tr>
<td>6-7</td>
<td>$\theta_6 + 90^\circ$</td>
<td>$d_6$</td>
<td>$a_6$</td>
<td>-90°</td>
</tr>
</tbody>
</table>
Position Kinematics:

\[
x = d_3 c_1 c_2 + a_2 c_1 c_2 \\
y = d_3 s_1 c_2 + a_2 s_1 c_2 \\
z = -d_3 s_2 - a_2 s_2 + d_1
\]

Orientation Kinematics:

\[
x = a_6(c_4 s_5 c_6 - s_4 s_6) - d_6 c_4 c_5 + a_5 c_4 s_5 \\
y = a_6(s_4 s_5 c_6 + c_4 s_6) - d_6 s_4 c_5 + a_5 s_4 s_5 \\
z = a_6 c_5 c_6 + d_6 s_5 + a_5 c_5 + d_4
\]

System Kinematics:

\[
x = -a_6 c_6(c_1 s_2 c_5 + s_5(s_1 c_4 - c_1 c_2 s_4)) + a_6 s_6(s_1 s_4 + c_1 c_2 c_4) - d_6(c_1 s_2 s_5 - c_5(s_1 c_4 - c_1 c_2 s_4)) - a_5 c_1 s_2 c_5 - a_5 s_5(s_1 c_4 - c_1 c_2 s_4) - d_4 c_1 s_2 + c_1 c_2(a_2 + d_3) \\
y = -a_6 c_6(s_1 s_2 c_5 - s_5(c_1 c_4 + s_1 c_2 s_4)) - a_6 s_6(c_1 s_4 - s_1 c_2 c_4) - d_6(s_1 s_2 s_5 + c_5(c_1 c_4 + s_1 c_2 s_4)) - a_5 s_1 s_2 c_5 + a_5 s_5(c_1 c_4 + s_1 c_2 s_4) - d_4 s_1 s_2 + s_1 c_2(a_2 + d_3) \\
z = -a_6 c_6(c_2 c_5 + s_2 s_4 s_5) - a_6 s_2 c_4 s_6 - d_6(c_2 s_5 - s_2 s_4 c_5) - a_5(c_2 c_5 + s_2 s_4 s_5) - d_4 c_2 - s_2(a_2 + d_3) + d_1
\]

3.8 Force-based Compensation for Minimal Friction Movement

To achieve the torque specifications for actuation with motors that fit well within the orthosis design, the addition of planetary gearheads was required. One drawback to increasing torque through these means is that planetary gearheads are never 100% efficient, and often waste energy due to small misalignments between the gears. This setup created a system that provided limited backdriveability with noticeable resistance present when trying to move the actuated joints back and forth. This lack of backdriveability could cause the user to adopt an abnormal movement to overcome the resistance and complete the task. Since the goal was to create a system to facilitate the restoration of normative movement profiles, it was necessary to minimize the resistance so that an individual’s movement was not impacted.
3.8.1 Concept Rationale and Configuration

To improve backdriveability, a compensatory control scheme was developed that provides a motor torque to create a zero force reading at the load cells. The applied torque is related to the movement of the user, but does not use movement information to provide the compensation. Instead, forces applied by the user are used to initiate and maintain motion. This results in the orthosis joints moving with the user’s hand to give the feeling of frictionless movement. The assistive torque created by the orthosis is directly proportional to the forces measured by the load cells mounted to the backside of the user’s hand. The load cells are attached to an aluminum bracket mounted to the hand brace such that one load cell is aligned with the top of the hand while the other is aligned to the bottom (Figure 12). This configuration enables differentiation between movements of the forearm and wrist whereby forces measured in the same direction between load cells corresponded to flexion/extension movements, while readings in opposing directions corresponded to pronation/supination.

Figure 12: Force Vectors at the load cells for different types of wrist movement. (Left) Flexion results in a tension on both the top and bottom load cells; (Left-Middle) Extension results in a compression on both load cells; (Right-Middle) Supination applies compression to the top cell and tension to the bottom cell; (Right) Pronation applies tension to the top cell and compression to the bottom cell.
3.8.2 Force-Torque Relationship

*Isometric Characterization of the Force-Torque Relationship for the Orthosis.* To relate the forces measured at the load cells to the torque output from the motor, a jig was constructed with two holes allowing the threaded extensions of the load cells to pass through. A pair of nuts secured the load cells to the jig. This created an isometric condition for the orthosis where a

![Jig for load cell attachment](image1.png)

![Brace pieces for locking passive rotation](image2.png)

**Figure 13:** Isometric force-torque setup and results. (Top) Setup for measuring force-torque relationship. (Left) Force-torque relationship for flexion and extension. (Right) Force-torque relationship for pronation and supination. (Middle) Force-torque relationship of the load cell at the top of the hand. (Bottom) Force-torque relationship for the load cell at the bottom of the hand.
torque could be applied by the motor, and the relative force could be read from the load cells. Torques ranging from 0-5 Nm were commanded to the motors, and the resulting forces were measured along the four actuated degrees of freedom (pronation, supination, flexion, and extension) to characterize the force versus torque relationship. A proportional relationship was found between the torques applied by the motor and forces measured at the load cells (Figure 13). For each direction of movement, the response was well represented by a linear relationship ($R^2 > 0.98$), who’s slope was used to define the gain between the sensed force and commanded torque.

*Dynamic Characterization of the Force-Torque Relationship with Inclusion of Orthosis and Limb.* The gain relating commanded motor torque and force measured at the load cells characterized the robot system, but failed to include the influence of the user’s limb. While the jig in the isometric setup did not bias either load cell, it could not be presumed that the influence of a subject’s hand would also be distributed evenly against both load cells. To model the force-torque relationship of the system with the limb, data was collected from the robot with a user strapped into the orthosis. Motion was isolated to a single axis by locking out one actuated axis via

![Pure Pronation/Supination Load Cell Response](image)

*Figure 14:* Oscillatory movement through the pronation/supination range of motion.
software, while the user performed an oscillating movement using their full range of motion without assistance from the device. For pronation/supination this included back and forth rotation of the forearm, and for flexion/extension this included back and forth rotation of the wrist. The profiles were recorded for each load cell (Figure 14), and the ratio between the top and bottom load cells was calculated at each time step (Figure 15). Data collected while moving between fully pronated and fully supinated states were extracted to calculate the final relationship between the load cells, as these points showed the most consistent ratios between oscillations. Instances where the load cell readings approached zero were ignored, as unrealistically high ratios were created at these points due to a near-zero value in the denominator. Minimum and maximum peaks were identified for each load cell, and 100 data points occurring both before and after the peak were included. This array size was selected to provide a large sampling of data points while maintaining a buffer between ratios as the load cells approached zero. These 200-point arrays, depicting instantaneous load cell ratios, were then averaged to give a final ratio value. This final

**Figure 15:** Ratio of forces between top and bottom load cells during pronation/supination. Each line represents a single oscillation moving from max flexion to max extension, and vice versa. Ratios shown occurred within the 200 samples measured around the peak force reading for each load cell, corresponding to period of greatest movement. Each sample represents a 1 ms interval.
ratio was used to characterize the relationship between load cells with subject input. The
distribution of force ratios indicated a 1:1 relationship between the top and bottom load cells for
pronation/supination movements (Figure 15).

**Figure 16**: Oscillatory movement through the flexion/extension range of motion.

**Figure 17**: Ratio of forces between top and bottom load cells during flexion/extension movements. Each line represents a single oscillation moving from max flexion to max extension, and vice versa. The black box indicates the region used to establish the load cell relationship. Each sample represents a 1 ms interval.
A similar approach was used for flexion/extension movements using a 1000-sample (1 sec) window centered on the maximum extents of flexion and extension. Unlike the traces in Figure 14 where a peak force was easily discernable, the plots for flexion and extension movements did not yield a clear single peak that coincided between the load cells (Figure 16). The ratio plot, shown in Figure 17, revealed a high level of variability, with larger ratios tending to occur at the edges of the window, closer to where the load cell readings approached zero. A 200-sample window centered on the points of maximum torque was used to estimate the relationship between the load cells for flexion and extension with subject input. A relationship of 1:1.7 was obtained between load cells for pure flexion/extension movements.

### 3.8.3 Control Implementation

During a typical reach and grasp movement, a combination of wrist and forearm rotation is employed. Compensating for the resistance of the orthosis as subjects move their wrist required an active control strategy that separated the assistive torques into the orthogonal degrees of freedom (FE - Flexion/Extension; PS – Pronation/Supination) actuated by the motors. The force-torque relationships developed in section 3.8.2 were used to develop a mathematical expression for the commanded torques at the motor required to zero the forces experienced at the wrist in response to combinations of flexion/extension and pronation/supination movements about the wrist.

Equations 10 and 11 define the force-torque relationships identified in Section 3.8.2 for Flexion/Extension and Pronation/Supination movements respectively.

\[
F_{top, FE} = 1.7 F_{bottom, FE}; \quad F_{bottom, FE} = \frac{F_{top, FE}}{1.7}
\]

\[
F_{top, PS} = -F_{bottom, PS}; \quad F_{bottom, PS} = -F_{top, PS}
\]
The theoretical relationship between the measured force and flexion/extension and pronation/supination movements about an axis perpendicular to, and equidistant from the load cells is shown in Equations 12 and 13.

\[ F_{top} = F_{top,FE} + F_{top,PS} \]  \hspace{1cm} (12)
\[ F_{bottom} = F_{bottom,FE} + F_{bottom,PS} \]  \hspace{1cm} (13)

Equations 14-17 mathematically define component forces for each actuated degree of freedom in terms of the forces read at each load cell. These were derived by substituting in the relationships from Equations 1 and 2 into Equation 3 to solve for each component force (See Appendix B for full derivation).

\[ F_{top,PS} = \frac{F_{top}}{2.7} - \frac{F_{bottom}}{1.588} \]  \hspace{1cm} (14)
\[ F_{bottom,PS} = -\frac{F_{top}}{2.7} + \frac{F_{bottom}}{1.588} \]  \hspace{1cm} (15)
\[ F_{top,FE} = \frac{F_{top} + F_{bottom}}{1.588} \]  \hspace{1cm} (16)
\[ F_{bottom,FE} = \frac{F_{top} + F_{bottom}}{2.7} \]  \hspace{1cm} (17)

Control to reduce the perceived mechanical impedance, or the resistance to motion as the user applies a force, was achieved using a closed-loop PID controller for each actuated joint. Force values from both load cells were fed back and used to generate a command torque to the motors using equations 14-17, using these results to calculate a resultant force \( F_{FE} = \frac{10F_{top,FE} + 17F_{bottom,FE}}{27} \), and multiplying the resultant force by the radius of rotation to obtain the torque. The error between the resulting torques and the current torque state of the motors was passed to the PID controllers to generate a corrective current command.

The PID controller gains were subsequently tuned using the “autotune” feature in Simulink (controller 1: P=206.4, I=-412868.3, D=0; controller 2: P=235.6, I=-471180.3, D=0). Both quantitative and qualitative assessments were performed using oscillating wrist rotations and forearm rotations (~1 Hz) performed with and without compensation. The oscillations were
performed for three wrist orientations; resting on the workspace table (neutral), elevated pointing 45° upward, and elevated pointing 45° downward (Figure 18). The results are shown in Table 4 with torques calculated as the average of the absolute torques collected throughout execution of the motion. The reduction in resistance in pronation/supination during compensation was more than 30% for the upward and downward wrist position, but less for neutral positioning (<10%) except in trial 3. The reduction in resistance for flexion/extension during compensation was approximately 40% for all wrist positions. Although the active compensation was not 100%, it did result in a noticeable change in the perceived ease of movement and was deemed adequate for the purposes of normative testing.

Figure 18: Orthosis orientations during joint resistance reduction assessment. (Left) 45° upward angle, (Right) 45° downward angle.
Table 4: Percent reduction of resistance in rho and theta during force compensation. Torques are given in Nm. Fields denoted by a * indicate torques that were not calculated.

<table>
<thead>
<tr>
<th>Trial</th>
<th>Orientation</th>
<th>Movement</th>
<th>Torque (No Compensation)</th>
<th>Torque (With Compensation)</th>
<th>% Reduction (Rho)</th>
<th>% Reduction (Theta)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>Rho</td>
<td>Theta</td>
<td>Rho</td>
<td>Theta</td>
<td></td>
</tr>
<tr>
<td>1</td>
<td>45° Up</td>
<td>Pronation/Supination</td>
<td>0.1265 *</td>
<td>0.0853 *</td>
<td>32.57</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Flexion/Extension</td>
<td>* 0.1136</td>
<td>* 0.0582</td>
<td>-</td>
<td>48.77</td>
</tr>
<tr>
<td></td>
<td>Neutral</td>
<td>Pronation/Supination</td>
<td>0.0557 *</td>
<td>0.0555 *</td>
<td>0.36</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Flexion/Extension</td>
<td>* 0.0697</td>
<td>* 0.0455</td>
<td>-</td>
<td>34.72</td>
</tr>
<tr>
<td></td>
<td>45° Down</td>
<td>Pronation/Supination</td>
<td>0.118 *</td>
<td>0.0885 *</td>
<td>25.00</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Flexion/Extension</td>
<td>* 0.0921</td>
<td>* 0.061</td>
<td>-</td>
<td>33.77</td>
</tr>
<tr>
<td>2</td>
<td>45° Up</td>
<td>Pronation/Supination</td>
<td>0.1332 *</td>
<td>0.0859 *</td>
<td>35.51</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Flexion/Extension</td>
<td>* 0.0936</td>
<td>* 0.0572</td>
<td>-</td>
<td>38.89</td>
</tr>
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<td>Pronation/Supination</td>
<td>0.0725 *</td>
<td>0.0672 *</td>
<td>7.31</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Flexion/Extension</td>
<td>* 0.0731</td>
<td>* 0.0414</td>
<td>-</td>
<td>43.37</td>
</tr>
<tr>
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<td>Pronation/Supination</td>
<td>0.0751 *</td>
<td>0.0615 *</td>
<td>18.11</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Flexion/Extension</td>
<td>* 0.0711</td>
<td>* 0.0429</td>
<td>-</td>
<td>39.66</td>
</tr>
<tr>
<td>3</td>
<td>45° Up</td>
<td>Pronation/Supination</td>
<td>0.0952 0.0351</td>
<td>0.0587 0.0168</td>
<td>38.34</td>
<td>52.14</td>
</tr>
<tr>
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<td></td>
<td>Flexion/Extension</td>
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<td>0.0234 0.0398</td>
<td>32.56</td>
<td>54.36</td>
</tr>
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<td>Pronation/Supination</td>
<td>0.055 0.0547</td>
<td>0.0385 0.0237</td>
<td>30.00</td>
<td>56.67</td>
</tr>
<tr>
<td></td>
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<td>Flexion/Extension</td>
<td>0.0179 0.0767</td>
<td>0.0124 0.0322</td>
<td>30.73</td>
<td>58.02</td>
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<tr>
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<td>Pronation/Supination</td>
<td>0.0769 0.0418</td>
<td>0.0519 0.0172</td>
<td>32.51</td>
<td>58.85</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Flexion/Extension</td>
<td>0.0375 0.0577</td>
<td>0.019 0.0298</td>
<td>49.33</td>
<td>48.35</td>
</tr>
</tbody>
</table>

3.8.4 Calibration

Prior to data collection using the orthosis, all sensors were calibrated to ensure accurate position outputs. Since position in 3D space is determined from the combined measurements of all sensors, one incorrect sensor could introduce error to the measured 3D position. A custom T-shaped mount was 3D printed to align the device and provide stability during calibration (Figure 19). A notch was cut along the shorter axis to align the device to a line drawn down the midsection of the table. During calibration, the mount (w/ orthosis attached) was set on the table and aligned with the calibration line with the prismatic slide fully extended. This was defined as the “zero” point for the potentiometers, as it closely resembles the neutral orientation of a user.
prior to use. Since the elevation joint was tilted downward to allow the orthosis to reach the tabletop, the angle at “zero-position” was calculated using trigonometric identities together with the known height and length of the device relative to the table surface. The system was started, and the potentiometer voltage values were measured in this position. The mount (with orthosis) was then moved to a second line drawn at a 25° angle to the first line. The mount was again aligned to the reference line, and the potentiometer voltage values were measured. The process was repeated twice more for each reference line to characterize the uncertainty in the calibration, and generate an averaged value to define the “zero position”. This procedure collects values for the prismatic slide in a fully extended state. To obtain a fully calibrated range of motion, the

Figure 19: Setup for position sensor calibration. (Top) Orthosis in the calibration block, (Bottom) CAD rendering for device positioning during calibration along the center-line of the table and an orientation 25° relative to the center-line.
calibration process was repeated with the slide fully retracted. The difference in these positions, along with the known extension length was used to calibrate the sensor.

Within the data acquisition model, detailed in Appendix G, a function block handled the conversion of sensor values from volts to degrees and corresponding calibration. Correct operation was verified by building the model with the calibration values, and running it with the device fully extended along the zero reference line. The (x,y,z) position was compared with physical measurements obtained prior to calibration. Calibrated positions within +/- 5 mm of the actual position were considered acceptable.
4 NORMATIVE TRAJECTORY ASSESSMENT

Designing a system to provide position-based assistance to impaired persons required a thorough understanding of the ways in which normal functioning individuals perform tasks. Since the system is intended for therapy settings that use activities of daily living as the tasks of interest, it was necessary to collect and characterize the movement profiles of healthy human subjects to provide a normative baseline. Subjects were instructed to perform five tasks of interest while wearing the orthosis, keeping their movements as natural as possible. The changes in position and orientation of the wrist were recorded as subjects performed the tasks to provide a set of reference trajectories for use with the position control model of the orthosis.

4.1 Task-specific Movement Trajectories

4.1.1 Subjects

Over the course of four months, 21 normal functioning subjects were recruited to perform five ADLs while their position profiles were recorded (8 male, 13 female; 17 right handed, 4 left handed; age range 19-30 for 20 subjects, 1 subject age 75). Inclusion criteria for this normative cohort included no impairments or conditions that would inhibit the ability to grasp and manipulate objects and being able to physically fit within the device. All participants gave informed consent and test procedures were approved by the Institutional Review Board at Marquette University.

4.1.2 Setup

Tasks were set up on the tabletop workspace using a converted puzzle board fastened to the table (see Figures 21-25) using Velcro. Objects used within the tasks were glued to a puzzle piece so that each time the task was executed, the objects would start from and return to the same location. This ensured consistency in how the objects were moved and manipulated across subjects. A “home” position was marked on the table, and served as the point from which every
task started and ended. The data acquisition model, previously described in Section 3.7, was loaded onto the Target computer. A monitor connected to the target displayed scopes providing information on joint position, force sensor readings, and joint space changes for the positioning joints. Within the Simulink model, real-time displays provided instantaneous x, y, z positions and joint measures. The motor controller software (EPOS Studio by Maxon™) enabled the controllers in Current Mode, monitored current outputs, and provided a quickstop feature to cut motor operation if needed.

Each subject was asked to put on a custom glove with the fingers removed, and a series of Velcro straps protruding from it (Figure 20). The subject then placed their arm into the orthosis so that the part of the hand between the thumb and index finger butted up against the hand brace. The glove straps were then fastened around the brace to lock the hand in place to ensure accurate force measurements. Two additional Velcro straps on the trough were fastened around the subject’s forearm to couple the forearm to the device.

4.1.3 Testing

Three trials were performed for each task with 10 iterations of the task executed within each trial. Prior to testing, subjects was given an opportunity to move while wearing the device and to practice the tasks. This allowed each subject to acclimate to the device, increasing the
likelihood that natural movement would be used while performing the tasks. During this phase, subjects were provided with verbal instructions and demonstrations of the tasks.

Subjects performed 5 total ADL tasks; drinking from a cup, pouring from a pitcher, eating with a spoon, stacking cones, and placing pegs. During testing, subjects were asked to sit up straight with their feet flat on the floor. Following a countdown, the data acquisition model was started, and subjects were told to begin the task when they were ready. After completion of one iteration of a task, and each subject’s hand had returned to the home position, the operator used a series of verbal cues to have subjects alter their wrist orientation until it was in the “zero” position. This ensured that the same wrist orientation was used at the start of each task/iteration. Proper wrist position was verified using the scopes on the target computer before starting the next task/iteration. Upon completion of each trial, the recorded data array was renamed relative to the task and trial and the next trial was set up. After all tasks and trials were completed, the entire set of data arrays in the MATLAB workspace was saved as a .mat file.

**Spoon to Mouth:** Prior to the task a bowl was placed on the task board directly in front of the subject and the subject was given a spoon to hold (Figure 21). During the task the subject moved from the home position toward the bowl, mimicked a scooping motion, and then brought

![Figure 21: Spoon to mouth task setup.](image)
the spoon to their mouth (as if eating cereal from the bowl). When the end of the spoon reached the mouth, the subject returned their wrist to the home position in the neutral orientation.

**Cup to Mouth:** Prior to the task a large cup was placed on the task board, centered but away from the subject (Figure 22). During the task, the subject moved their hand to the cup, grasped the cup, and then brought the cup to their mouth. Once the rim of the cup was in close proximity to the mouth, subjects returned the cup to its place on the task board, released their grasp, and returned their wrist to the home position in the neutral orientation.

**Pouring from a Pitcher:** Prior to the task a small cup and measuring cup with handle (acting as the pitcher) were placed on the task board. The pitcher was placed directly in front of the subject, while the cup was placed to the left of the pitcher (Figure 23). During the task, the subject moved to the pitcher, grasped the handle, and pronated their forearm to imitate pouring from the pitcher into the small cup. Subjects were instructed to briefly hold the pitcher in the pouring position and then to return the pitcher to its starting place on the task board. The subject then released their grasp on the pitcher and returned their wrist to the home position in a neutral orientation.
**Stacking Cones**: This task used a stack of four plastic cones with one target cone glued to a task board puzzle piece. The target cone was placed in the lower left corner of the task board while the rest of the cones were placed in the upper right (Figure 24). During the task, the subject moved to the cones in the upper right of the task board, grasped the top cone and placed it on top of the glued cone. The process was repeated until all cones were moved from the upper right and placed on the target cone in the bottom left. Once all cones were moved, the subject returned to the home position with the wrist oriented in the neutral position.

**Figure 23**: Pouring from pitcher task setup.

**Figure 24**: Stacking cones task setup.
**Placing Pegs:** Prior to the task a foam board was placed on an easel and five plastic pegs were placed within openings on the task board (Figure 25). The order of placement occurred as follows; blue, purple, red, yellow, orange. During the task the subject grasped each peg, and placed it in its predetermined hole on the foam board. Once all pegs were inserted into their correct locations, the subject returned to the home position with the wrist oriented in the neutral position.

![Figure 25: Placing pegs task setup.](image)

**4.2 Trajectory Processing**

In total, 63 trials were compiled for each task with a total of 630 iterations of each task. Profiles for the cup task, averaged for each subject, are shown in Figure 26. Profiles for the other tasks are featured in Appendix C. Qualitative observations indicated that most tasks were performed in a similar manner across subjects. However, the variability in individual trajectories and the potential presence of multiple strategies for some tasks made it important to develop an automated approach to characterize normative task-specific movement profiles.
The primary means by which movement profiles were grouped within task was the creation of a custom artificial neural network that represented the combined effects of all iterations and trials into a single mathematical expression. With the intent of identifying the general movement profiles across subjects, the model design focused on correlating changes in orientation angle (henceforth referred to as rho for pronation/supination and theta for flexion/extension) and changes in position (x,y,z). This structure emphasized a model solution that could predict changes in position and orientation based on recent (x,y,z) position information and previous wrist orientation. This made it easier to compare movement strategies for each task across subjects, as the correlation between position and orientation was preserved irrespective of time course. The network was constructed using the Mathworks™ Neural Network Toolbox v.8.2, within the MATLAB (r2014a) environment.

**Figure 26:** Cup task average movement profiles. Each line represents the average path taken by each subject across all iterations.

### 4.2.1 Characterizing Approaches Within Tasks

The primary means by which movement profiles were grouped within task was the creation of a custom artificial neural network that represented the combined effects of all iterations and trials into a single mathematical expression. With the intent of identifying the general movement profiles across subjects, the model design focused on correlating changes in orientation angle (henceforth referred to as rho for pronation/supination and theta for flexion/extension) and changes in position (x,y,z). This structure emphasized a model solution that could predict changes in position and orientation based on recent (x,y,z) position information and previous wrist orientation. This made it easier to compare movement strategies for each task across subjects, as the correlation between position and orientation was preserved irrespective of time course. The network was constructed using the Mathworks™ Neural Network Toolbox v.8.2, within the MATLAB (r2014a) environment.
The network structure was based on a NAR (nonlinear autoregressive) model. NAR models are designed to predict the next value in a data series based on the current value and a specified number of points occurring previously. The NAR model requires that the number of previous samples used to predict the output, and number of hidden neurons be specified. As the number of samples used to predict the output are increased, older values become less correlated with the current sample and are less useful in predicting future values. However, the inclusion of older values make the model more robust and better at fitting all data sets, as the use of more samples creates a more generic model. The addition of more hidden neurons makes it possible to better track nonlinear interactions between inputs, but also complicates the expression and increases computational complexity. Using an iterative optimization process, the number of previous samples used and hidden neurons was minimized to reduce the likelihood of model overfitting and increase generalization across movement trajectories. Nine previous samples and twenty hidden neurons were used to predict the normative movement profiles.

Typical NAR models employ nonlinear transfer functions to better predict nonlinear data. However, this significantly increases the complexity of characterizing movement profiles using the structure of the network coefficients. For this reason, linear transfer functions were used in the model to approximate the nonlinear mapping between network inputs and outputs. For the relatively low frequencies associated with wrist motion (< 2 Hz), the linear approximation provided a good representation of the movement profiles (Figure 27).

Data was combined and re-sampled prior to training so that the created model used coefficients related to the kinematically relevant changes in the movement profile. Changes in position of the wrist along the x, y, and z-directions of the workspace, and changes in wrist orientation (pronation/supination, flexion/extension) were used as inputs to predict the next orientation of the wrist. For each subject, the three trials performed for each task were placed together in a single array, and periods of inactivity between iterations removed. Movement
sequences were then down sampled from a 1 ms to 100 ms resolution to enable model training at timescales relevant to kinematic changes in movement.

For each subject and task (63 total), NAR models were trained to predict the movement profile. Because of the custom applied network parameters, as described in Figure 27, the traditional toolbox was not adequate for training and validating the model. Instead, the built-in network training function was called manually to train the model using 90% of the collected iterations. 10% of the iterations were randomly parsed out for use in validating the model. Overall model fit was evaluated post hoc using a custom function to calculate the average movement error associated with the model prediction across the entire movement sequence (Figure 28). Errors were then averaged across subjects for each task (Table 5). Average error across subjects was low, within 1-2 mm for positional measures and .02-.03 rad for orientation measures.

**Table 5:** Average errors from predictive NAR model. Means and standard deviations for x, y, and z are given in mm while rho and theta are given in radians.

<table>
<thead>
<tr>
<th></th>
<th>X</th>
<th>Y</th>
<th>Z</th>
<th>Rho</th>
<th>Theta</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cup</td>
<td>1.32 ± .20</td>
<td>1.79 ± .28</td>
<td>.85 ± .10</td>
<td>.023 ± .005</td>
<td>.007 ± .002</td>
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<td>Pitcher</td>
<td>1.14 ± .31</td>
<td>1.47 ± .28</td>
<td>.89 ± .13</td>
<td>.023 ± .006</td>
<td>.008 ± .005</td>
</tr>
<tr>
<td>Spoon</td>
<td>1.11 ± .27</td>
<td>1.64 ± .56</td>
<td>.96 ± .16</td>
<td>.020 ± .007</td>
<td>.008 ± .004</td>
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<td>Cones</td>
<td>1.69 ± .50</td>
<td>2.00 ± .40</td>
<td>1.18 ± .40</td>
<td>.022 ± .005</td>
<td>.010 ± .004</td>
</tr>
<tr>
<td>Pegs</td>
<td>1.53 ± .21</td>
<td>2.37 ± .31</td>
<td>1.04 ± .16</td>
<td>.029 ± .006</td>
<td>.018 ± .007</td>
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4.2.2 Movement Profile Analysis using K-means Clustering

Each trained neural network model provided a single expression representing the changes in position and orientation occurring during each task for each subject. The recursive structure of the network provided a spatiotemporal model of the underlying movement profile determined by subject’s control strategy and limb kinematics. This allowed direct comparison of the model structure across subjects to determine whether significant inter-subject differences existed in the movement used to complete a task, and whether such differences could be parsed into distinct classes of movement. If multiple approaches (i.e., movement profiles) were found within a task,
each would have a characteristic set of 3D locations and orientations that would be difficult to account for within a single position-based assistive control scheme. In such cases a switching control scheme was developed based on the movement strategy being implemented by the subject. To make these determinations, model grouping was conducted using K-means clustering.

K-means clustering takes all observations submitted to it and partitions them into groups (or clusters) based on their proximity to the mean of the cluster. The neural network model output is based on the 9 previous samples used to predict the next value, the 5 input directions, and 5 output directions, resulting in 225 weight coefficients; each corresponding to a separate dimension in the K-means clustering. Principle Component Analysis (PCA) was performed prior to clustering to reduce the dimensionality of the cluster space from 225 down to 20. K-means clustering was then performed on the PCA components. Cluster results were confirmed visually along the first three PCA components which collectively accounted for 60-70% of the variance in the model coefficients for all tasks. In addition to creating 3-D cluster plots, silhouette plots were created giving insight into how well the data fit into the assigned clusters. Each observation was assigned a “silhouette index” value.

Complete results of the clustering analysis are provided in Appendix D and are illustrated in Figure 29 and Figure 30 for the cone stacking task. Clusters are shown graphically using combinations of the top three principle components. The silhouette plot (Figure 29, Bottom Right), characterizes the quality of the clustering, comparing the distance of the observation from the mean of the assigned cluster with the distance to the mean of the next closest cluster. Values near 1 indicate an excellent fit within the cluster while values less than 0 indicate a possible incorrect cluster assignment. Silhouette values greater than 0.5 indicate a good cluster fit. These values are governed by the equation

$$S_i = (b_i - a_i)/\max(a_i, b_i)$$

(18)
where $S_i$ is a point in a cluster, $b_i$ is the minimum average distance from $S_i$ to points in other clusters, and $a_i$ is the average distance from $S_i$ to other points in the same cluster. The number of distinct movement strategies was determined by adjusting the number of clusters to maximize the mean silhouette index while minimizing the number of clusters. The best group classification of the cup-to-mouth task occurred when movement profiles across subjects were divided into two groups, based on the silhouette plot. The clusters were represented by a low mean silhouette index value (0.3248), and a qualitative look at examination of the normative movement plots showed little difference between movement profiles. As a result, the task was characterized by a single movement strategy. Similar results were obtained following K-means analysis of the pouring-from-pitcher, spoon-to-mouth, and peg tasks. Low mean silhouette value (0.4017, 0.3892, and 0.3512 respectively) for two clusters and lack of support from the normative
movement plots led to these tasks being defined by a single movement strategy. This result is not surprising as the constraints imposed by the positions of the items (i.e., cup and pitcher, bowl and mouth, and pegs) limited the number/range of task-relevant movements.

The cone task was the only task that showed multiple (and distinct) movement profiles. In spite of the low silhouette index (0.2824) resulting from outlier samples, three clusters were identified based on the distribution of samples into the clusters. As shown in Figure 30, a separation could be seen between the position and orientation profiles used to complete the task. The greatest indication for multiple movement strategies came from the forearm rotation (Rho) vs. wrist rotation (Theta) subplot, which shows three distinct orientation profiles. Cluster 3 (green) shows much greater pronation and extension, while Clusters 1 (blue) and 2 (red) show

Figure 30: Mean normative movement profiles grouped by clustering assignment. (Top Left) XY Plane, (Top Right) XZ Plane, (Bottom Left) YZ Plane, (Bottom Right) Rho vs. Theta. Each color reflects the color groupings used in Figure 29.
similar levels of pronation. The primary difference between Clusters 1 and 2 was the increased level of extension used by Cluster 2. Examination of the average movement profiles indicated that members of Cluster 3 used an over-the-top approach to grasp the cone, as indicated by the greater pronation and increased wrist extension to remove or place the cone on the stack. This was supported by the lower heights recorded in the z-direction for this group. Cluster 1 subjects mostly used changes in shoulder and elbow position to transport the cones, as indicated by a consistent neutrality of the wrist and the greatest elevation changes observed in the z-direction. Cluster 2 subjects employed more wrist extension to create a straight-line movement between stacking locations.

**Figure 31:** Mean normative movement profiles grouped by clustering assignment for the cup task. (Top Left) XY Plane, (Top Right) XZ Plane, (Bottom Left) YZ Plane, (Bottom Right) Rho vs. Theta. Each color reflects each of the two clusters determined for this task, as shown in Appendix D.
The determination of individual approaches for the other tasks was also aided by graphically observing the position and orientation profiles for each task. Figure 31 shows the cup-to-mouth task, where distinct approaches were not clearly evident, even though two separate clusters were identified. The 3D trajectories do not show a clear distinction between the movements, either in position or orientation. This qualitative assessment, coupled with the low silhouette index, provided adequate evidence to justify the assimilation of all subject models into a single representative approach for use in active assistance. Similar results were observed for the other tasks, as multiple clusters could be formed, but no distinct differences were observed in the trajectories.

4.3 Summary

The neural network modeling of subjects’ movement profiles and subsequent K-means clustering revealed several important features of the task-specific movement profiles employed across subjects. In the cup-to-mouth and spoon-to-mouth tasks, movement profiles were highly variable along the z-direction due to differences in head height relative to the table. Taller subjects needed to lift their limb higher to reach their mouth than shorter subjects. By normalizing the peak height of movement profile to the subject’s head height, a single movement profile could be used to account for task-specific movement across subjects.

Across tasks, spatial constraints associated with object location and grip type resulted in less movement variability. Tasks like cup-to-mouth and spoon-to-mouth included a single object with a constrained start and end location, however subjects were free to move anywhere during the rest of the task. Tasks like the cone-stacking and peg-placement operated on a very specific order and location of where objects could start and end. These constraints greatly limit the degrees of freedom available to the subject to perform the task and was reflected in the consistency of the path taken across all subjects. The cone-stacking task was the only one to show...
evidence of multiple approaches. Even though the task provided less opportunity to alter the
beginning and ending positions of the objects, multiple types of grip could be used to perform the
task, resulting in different movement profiles/strategies. When using this approach on a greater
catalog on ADL tasks in the future, the availability of grip options should be taken into account
and used as a basis for classifying approaches within a task.
5 MAP-BASED ASSIST CREATION AND VALIDATION

Several approaches were considered when deciding on a method to provide active assistance during ADL tasks. Hard-line trajectories, or singular movement profiles (for both position and orientation) were considered, but were not practical in a system where location of the wrist in space could not be controlled. The artificial neural networks developed in Chapter 4 to characterize movement trajectories were examined but were unable to provide a task-specific path due to unconstrained prediction errors tied to the recursive feedback. Attempts to utilize this approach applied the centroid model of the approach cluster, representing the “ideal” case”, and examined active compensation using a normal approach, and an approach with an altered location in the 3D work space. Applying the altered location resulted in an oscillation of the commanded orientation that led to instability. This result was attributed to the unconstrained nature for the positioning degrees of freedom and the lack of an adaptive mechanism in the controller design. Further investigation of this method would have exceeded the scope of this thesis, so development was not continued.

The approach that showed the most promise was the creation of a means to directly map desired wrist orientation across the workspace. This allowed variation in movement profiles to be accounted for, providing a range of paths a subject could follow in order to complete a task. The method did not require control of the shoulder or elbow, as a subject making a reasonable effort to complete the task would cross through the mapped points and receive assistance as needed. Creating a spatial map independent of time also allowed subjects to complete the task at a speed comfortable to them without impacting the effectiveness of the map.

Validation of the robot performance during active control was performed in two phases. The first phase featured a robot-only condition where a mock position profile was submitted to the control system and the response of the robot was recorded. By performing this without the
limb, parameters such as response time and accuracy could be determined. The resulting profile also provided a baseline, enabling the influence of the limb on control system performance to be determined. The second phase of the validation used normal functioning subjects performing the five tasks (3 for cup and pitcher; 2 for spoon, cone, and peg) while allowing the robot to influence their wrist and forearm orientation. Each task was performed for up to four conditions (depending on the task) where the system’s ability to assist the subject in completing the task was evaluated. These included a wrist guidance condition, where the subject changed position of the wrist appropriately, but allowed the robot to fully influence wrist and forearm orientation, a co-contraction condition where the subject made an effort to keep the limb stiff while the robot applied assistance, and two alternate approach conditions that had the subject either take a wide approach or narrow approach to the task object. The purpose of these conditions was to simulate impaired conditions such as limb weakness (wrist guidance), spasticity (co-contraction), or abnormal movement profiles (alternative approaches) as part of a compensatory movement strategy. Performance during active control was evaluated by comparing the target trajectory profiles, obtained from the normative movement testing, to those created under active control. A performance acceptance criteria of 5° was set for the average error, based on the proprioceptive error in wrist motion for normally functioning adults [33], and evaluated across conditions.

5.1 Control Loop Structure

The control strategy for assistive movements uses close-loop PID control, commanding new wrist orientation values based on the current and desired orientations (Figure 32). The user’s wrist position in 3D space was used as an uncontrolled input to determine the desired task-specific wrist orientation based on a normative trajectory map created from the movement profiles of healthy neurologically intact subjects. In addition to the error signal generated between the actual and desired orientations, extraneous forces experienced at the wrist, including inertial effects, Coriolis and centrifugal forces, and gravity, were also taken into consideration to generate
the output torque sent to the motors to orient the wrist in space. The extraneous forces were included based on the Lagrangian equations of motion, commonly used in robotics applications, which are created from the dynamics of the system. The final form of the equations of motion are represented as a second order differential equation:

\[ \mathbf{\tau} = \mathbf{M}\ddot{\mathbf{\theta}} + \mathbf{C}\dot{\mathbf{\theta}} + \mathbf{G}\theta \]  

(19)

The mass matrix, \( \mathbf{M} \), provides inertial values based on the center of mass for each moving section of the device. Within the matrix \( \mathbf{C} \), terms related to Coriolis and centrifugal forces acting on the device while in motion, are contained and displayed as a function of the angular velocity at the joint. The final term, \( \mathbf{G} \), describes the gravitational effect on the device as a function of the height above the workspace surface. Together, these terms can be summed to determine the torque needed to drive the device based on the specified position, velocity, and acceleration for any

\[ \text{Figure 32: Closed-loop control of wrist orientation. Current wrist position and orientation (red circles) inputs into the control model, with the positions processed by the trajectory mapping (green rectangle) to determine the desired wrist orientation in space. Current orientation is routed through the modeled dynamics of the robot (blue rectangles). The PID controller that generates an error signal based on the measured differences between desired and current wrist orientation. The combined outputs of the dynamics and controller produce the needed torque at the motors.} \]
moment in time. The controller terms come into play when an error signal is generated due to differences between the actual measured position and the desired position of the actuated joints. This added term alters the torque used in order to compensate for the positional difference.

Rewriting the equation of motion with the controller included appears as:

\[
\tau = M\ddot{\theta} + C\dot{\theta} + G\theta + K_p e + K_d \dot{e} + K_i \int e
\] (20)

5.2 Trajectory Map Creation

Three dimensional trajectory maps were created for each task to link task-specific wrist orientations to 3D wrist locations in space. An advantage to using location based maps is the elimination of time-dependency; subjects are free to complete tasks at their own pace, and receive assistance from the robot as needed based on the wrist location in the task space. Each trajectory map covered the workspace of the manipulandum, encompassing a volume 600 mm x 600 mm x 300 mm.

Prior to analysis, limb position data was resampled at 10 Hz and task-specific movement sequences were parsed from the overall movement profile using the times at which the subject’s wrist left and returned to the 50mm x 50mm home region. Resampling was done to reduce the volume of data points represented in the trajectory map, while maintaining an accurate depiction for changes in position and orientation. Locations within the workspace were sampled at 1cm\(^3\) voxel resolution, after initial attempts to use a 1mm\(^3\) voxel resolution proved computationally infeasible. For 5 subjects, single iterations were omitted for a task due to a need to restart the iteration (object dropped, wrong order used), and in one case the subject began a task prior to the recording being started. All other recorded iterations were used.

Map volumes were created for each task using a custom MATLAB script (Appendix H) that assigned each wrist orientation measured at a 3D location to the equivalent voxel in the workspace throughout the subject’s movement sequence. This initial assignment created the
potential for multiple wrist orientation to be assigned to the same volume 3D location, due to the
number of subjects and the cyclic nature of the tasks. For each task and segment of the movement
profile, a trajectory map was created by averaging at each voxel in the workspace, the wrist
locations assigned to location across trials and subjects. This volumetric map would serve as the
basis for the trajectory mapping function implemented in the controller (Figure 32).

Most tasks utilized a cyclic path through the workspace, starting and ending at the same
point. As a result, many positions within the workspace were crossed twice; once at the beginning
of the task, and again at the conclusion of the task. This presented a problem for the use of a
spatially defined map since the spatial map did not allow more than one orientation to be mapped
to the same position. To overcome this issue, key features within each task were used to split the
full movement into one or more segments. The number of segments was task-specific and
corresponded to the fewest needed to ensure a unique mapping between wrist orientation and 3-D
position within the segment. The division of segments based on task are summarized in Table 6.
An illustration of this division for the cup task is depicted in Figure 33. Most tasks only required
two segments to ensure a unique mapping, and were easy to distinguish across all subjects.

Table 6: Transition Segments for Each Task

<table>
<thead>
<tr>
<th>TASK</th>
<th>NO. OF SEGMENTS</th>
<th>TRANSITION POINTS</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cup-to-Mouth</td>
<td>2</td>
<td>Cup at mouth (max pronation)</td>
</tr>
<tr>
<td>Pouring-from-Pitcher</td>
<td>2</td>
<td>Greatest pitcher tilt into cup (max pronation)</td>
</tr>
<tr>
<td>Spoon-to-Mouth</td>
<td>2</td>
<td>Spoon at mouth</td>
</tr>
<tr>
<td>Cone-Stacking</td>
<td>3</td>
<td>Cone picked up at starting position; Cone dropped on final position</td>
</tr>
<tr>
<td>Peg-Placement</td>
<td>2</td>
<td>Peg #1 placed in board</td>
</tr>
</tbody>
</table>
5.2.1 Map Smoothing Using Gaussian Kernel

Since each trajectory map characterized the task-specific wrist orientation across subjects, trajectory variability within and between subjects created a volume in which adjacent wrist orientations were not well correlated. This produced discontinuities in the trajectory assistance, particularly for location that were not tightly constrained by the task (e.g. between waypoints). To establish smooth transitions between adjacent points in space, a volume normalized 3-dimensional Gaussian kernel was applied with a spatial extent of 15 cm on each side (15cmx15cmx15cm) and standard deviation of 2.75 cm.

The spatial smoothing process reduced the desired wrist orientation at each location due to the redistribution of power associated with the smoothing kernel. This was countered by applying a linear scaling factor to restore the map values to the correct magnitudes. For maps where the maximum magnitude occurred at the transition between maps, data was mirrored about

Figure 33: Map Transitions within a task. For the cup task shown above, the task is split into two separate maps; the first maps the trajectory as cup is brought to the mouth, while the second maps the trajectory as the cup is returned to the table.
the discontinuity point prior to smoothing to maintain local power across the discontinuity, giving the appearance of spatially continuous map.

5.3 Trajectory Map Integration

During active control, a mapping subsystem was used to obtain desired wrist orientations from the trajectory maps. Cartesian wrist position and linear velocity, along with rotational velocity about the joint were provided as inputs to the subsystem, which then used the trajectory maps to determine the desired wrist orientation (rho and theta). Since it was unlikely that a subject’s wrist position fell precisely on a mapped value, trilinear interpolation was used to generate orientation values based on adjacent mapped values.

With each task, at least two maps were used because of their cyclic nature (Table 6). To make sure the transition between the maps was smooth, sigmoid equations were developed and integrated into the mapping function (see Appendix H). The applied sigmoid ramped up to the desired wrist orientation on the incoming map, while ramping down the desired wrist orientation on the outgoing map simultaneously. The two values were summed over an 800 ms temporal window around the map transition to provide temporal smoothing. All transitions were governed by the expression

$$\text{orientation} = \left(1 - \frac{1}{1 + e ^ {\frac{-\text{time} - t_i}{.058}} \times \text{map1}}\right) + \left(\frac{1}{1 + e ^ {\frac{-\text{time} - t_i}{.058}} \times \text{map2}}\right)$$

(21)

where time = current sample time, ti = initial time + 400ms. The initial time is established once a positional threshold is crossed near the end of the mapped volume, which initiates the switch to the next map. This starts the 800 ms sigmoid transition between maps. An initial weight of zero is applied to the orientation and smoothly ramped, reaching a weight of 0.5 after 400 ms and a weight of 1 by the time 800 ms has elapsed. By subtracting the first term from 1, the expression creates a ramp-down effect as opposed to a ramp-up. For cases where the subject is either leaving
Home or returning to Home, the map1 or map2 term is set to zero to facilitate a single ramp-up or ramp-down. The 800 ms temporal window was chosen to maximize the smoothness of the transition while minimizing interference with transitional movements below 1 Hz.

5.4 Active Control Validation: Robot-Only

The first phase of validation was performed using pre-generated position profiles, representing the average task trajectories across subjects used previously to define the trajectory maps. By creating the profiles based on the average trajectory, simulated wrist orientations were well aligned with the mapped orientations derived from the distribution of orientations in the mapped volume. The orthosis was mounted to an L-bracket fastened to the workspace tabletop, taking advantage of threaded holes present on the bottom of the turntable bearing. This kept the orthosis stationary, allowing only actuated joints to move during the tests (Figure 34). Movement profiles were generated at three different movement frequencies (slow, normal, and fast), to evaluate the system performance with respect to the bandwidth of movement frequencies. The normal speed condition corresponded to the average amount of time required for subjects to complete one iteration of the task. The speeds of the fast and slow conditions were defined to be 30±3% higher and lower than the normal condition respectively. For each task, five repetitions were performed for each movement profile with a two second stationary period at the home
location between repetitions. Times for completion of task iterations at each speed condition are provided in Table 7. Profiles for commanded and measured wrist orientation were directly compared for each task and speed condition, looking at response time, accuracy, and movement bandwidth as a function of behavioral task and speed (Table 8). Response time of the control system were consistent across trials. Accuracy was measured by calculating the average error between the measured orientation and commanded orientation. Across tasks and speed conditions, only one instance was observed where the average error exceeded 1° (1.11°). Error contributed due to the response delay can be determined by taking the difference between the average error with the delay and the error after aligning commanded orientation to measured orientation. Based on the low observed error contributed by the closed loop control system, further statistical analysis was not performed. When relating these results back to the application of the device, differences in error between tasks or speeds of less than a degree would be

Table 7: Time per iteration for each task during Robot-only evaluation.

<table>
<thead>
<tr>
<th>Task</th>
<th>Speed</th>
<th>Time per Iteration (sec)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cup to Mouth</td>
<td>Slow</td>
<td>8</td>
</tr>
<tr>
<td></td>
<td>Normal</td>
<td>6</td>
</tr>
<tr>
<td></td>
<td>Fast</td>
<td>4</td>
</tr>
<tr>
<td>Pouring from Pitcher</td>
<td>Slow</td>
<td>8</td>
</tr>
<tr>
<td></td>
<td>Normal</td>
<td>6</td>
</tr>
<tr>
<td></td>
<td>Fast</td>
<td>4</td>
</tr>
<tr>
<td>Eating with Spoon</td>
<td>Slow</td>
<td>6.5</td>
</tr>
<tr>
<td></td>
<td>Normal</td>
<td>5</td>
</tr>
<tr>
<td></td>
<td>Fast</td>
<td>3.5</td>
</tr>
<tr>
<td>Stacking Cones</td>
<td>Slow</td>
<td>14</td>
</tr>
<tr>
<td></td>
<td>Normal</td>
<td>11</td>
</tr>
<tr>
<td></td>
<td>Fast</td>
<td>8</td>
</tr>
<tr>
<td>Placing Pegs</td>
<td>Slow</td>
<td>21</td>
</tr>
<tr>
<td></td>
<td>Normal</td>
<td>16</td>
</tr>
<tr>
<td></td>
<td>Fast</td>
<td>11</td>
</tr>
</tbody>
</table>
Table 8: Robot-only validation results. Accuracy is shown as the average observed error in both an “aligned” (measured orientation shifted toward commanded orientation based on response delay) and “delay” (response delay included) conditions. Movement bandwidth is characterized in terms of the -3dB cutoff and 90% cumulative power frequencies.

<table>
<thead>
<tr>
<th>TASK</th>
<th>CONDITION</th>
<th>JOINT</th>
<th>RESPONSE/Delay (ms)</th>
<th>AVG ERROR (aligned) (deg)</th>
<th>AVG ERROR (delay) (deg)</th>
<th>BANDWIDTH (-3 dB) (Hz)</th>
<th>BANDWIDTH (90%) (Hz)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cup</td>
<td>Normal</td>
<td>Rho</td>
<td>10</td>
<td>0.321</td>
<td>0.474</td>
<td>0.325</td>
<td>1.05</td>
</tr>
<tr>
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<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>Theta</td>
<td>17</td>
<td>0.310</td>
<td>0.480</td>
<td>0.675</td>
<td>2.175</td>
</tr>
<tr>
<td>Fast</td>
<td>Rho</td>
<td></td>
<td>7</td>
<td>0.333</td>
<td>0.441</td>
<td>0.667</td>
<td>1.833</td>
</tr>
<tr>
<td></td>
<td>Theta</td>
<td></td>
<td>13</td>
<td>0.322</td>
<td>0.449</td>
<td>1.333</td>
<td>3.833</td>
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<tr>
<td>Slow</td>
<td>Rho</td>
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<td>12</td>
<td>0.313</td>
<td>0.475</td>
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<td></td>
<td>Theta</td>
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<td>18</td>
<td>0.351</td>
<td>0.514</td>
<td>0.61</td>
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<td>Pitcher</td>
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<td>Rho</td>
<td>11</td>
<td>0.243</td>
<td>0.411</td>
<td>0.36</td>
<td>0.987</td>
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<tr>
<td></td>
<td></td>
<td>Theta</td>
<td>15</td>
<td>0.308</td>
<td>0.341</td>
<td>0.62</td>
<td>1.625</td>
</tr>
<tr>
<td>Fast</td>
<td>Rho</td>
<td></td>
<td>6</td>
<td>0.251</td>
<td>0.362</td>
<td>0.491</td>
<td>1.15</td>
</tr>
<tr>
<td></td>
<td>Theta</td>
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<td>17</td>
<td>0.297</td>
<td>0.363</td>
<td>0.64</td>
<td>1.8</td>
</tr>
<tr>
<td>Slow</td>
<td>Rho</td>
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<td>13</td>
<td>0.235</td>
<td>0.415</td>
<td>0.283</td>
<td>0.9</td>
</tr>
<tr>
<td></td>
<td>Theta</td>
<td></td>
<td>17</td>
<td>0.255</td>
<td>0.290</td>
<td>0.588</td>
<td>1.494</td>
</tr>
<tr>
<td>Spoon</td>
<td>Normal</td>
<td>Rho</td>
<td>9</td>
<td>0.281</td>
<td>0.354</td>
<td>0.7</td>
<td>1.41</td>
</tr>
<tr>
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<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>Theta</td>
<td>25</td>
<td>0.229</td>
<td>0.448</td>
<td>0.54</td>
<td>1.137</td>
</tr>
<tr>
<td>Fast</td>
<td>Rho</td>
<td></td>
<td>7</td>
<td>0.249</td>
<td>0.301</td>
<td>0.536</td>
<td>1.966</td>
</tr>
<tr>
<td></td>
<td>Theta</td>
<td></td>
<td>17</td>
<td>0.236</td>
<td>0.377</td>
<td>0.875</td>
<td>1.62</td>
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<tr>
<td>Slow</td>
<td>Rho</td>
<td></td>
<td>10</td>
<td>0.264</td>
<td>0.342</td>
<td>0.576</td>
<td>1.41</td>
</tr>
<tr>
<td></td>
<td>Theta</td>
<td></td>
<td>26</td>
<td>0.240</td>
<td>0.439</td>
<td>0.347</td>
<td>1.045</td>
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<td>Cone</td>
<td>Normal</td>
<td>Rho</td>
<td>9</td>
<td>0.339</td>
<td>0.481</td>
<td>0.602</td>
<td>1.45</td>
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<td>0.473</td>
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<tr>
<td>Fast</td>
<td>Rho</td>
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<td>0.331</td>
<td>0.471</td>
<td>0.681</td>
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<tr>
<td></td>
<td>Theta</td>
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<td>0.278</td>
<td>0.469</td>
<td>1.185</td>
<td>2.2</td>
</tr>
<tr>
<td>Slow</td>
<td>Rho</td>
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<td>11</td>
<td>0.309</td>
<td>0.459</td>
<td>0.488</td>
<td>1.312</td>
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<tr>
<td></td>
<td>Theta</td>
<td></td>
<td>20</td>
<td>0.244</td>
<td>0.476</td>
<td>0.616</td>
<td>1.302</td>
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<tr>
<td>Peg</td>
<td>Normal</td>
<td>Rho</td>
<td>9</td>
<td>1.028</td>
<td>1.114</td>
<td>0.606</td>
<td>1.72</td>
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<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>Theta</td>
<td>18</td>
<td>0.355</td>
<td>0.669</td>
<td>0.442</td>
<td>1.38</td>
</tr>
<tr>
<td>Fast</td>
<td>Rho</td>
<td></td>
<td>6</td>
<td>0.455</td>
<td>0.569</td>
<td>0.9</td>
<td>3</td>
</tr>
<tr>
<td></td>
<td>Theta</td>
<td></td>
<td>14</td>
<td>0.342</td>
<td>0.625</td>
<td>0.825</td>
<td>1.929</td>
</tr>
<tr>
<td>Slow</td>
<td>Rho</td>
<td></td>
<td>8</td>
<td>0.494</td>
<td>0.572</td>
<td>0.515</td>
<td>1.61</td>
</tr>
<tr>
<td></td>
<td>Theta</td>
<td></td>
<td>18</td>
<td>0.359</td>
<td>0.619</td>
<td>0.47</td>
<td>1.3</td>
</tr>
</tbody>
</table>

imperceptible to a subject. Operational bandwidth of the orthosis was verified with respect to the movement tasks using two metrics, the -3 dB cutoff within the pass band and the frequency below which 90% of the cumulative spectral power was contained.
Figures 35-44 show the trajectory progression at key points through the mapped volume in each task. Each plot contains a 2D slice of the 3D trajectory map along the X-Y plane (cm) taken at the height in the work space associated with the task at the key point. Key points were selected within each task at locations that were considered an “end goal” of a movement segment; this included grasping propositioned objects, bringing objects to the mouth, or placing grasped objects at a predefined location. Figures 35 and 36 show the corresponding trajectory map for pronation/supination and flexion/extension during the cup-to-mouth task. Figure 35 shows maximum pronation occurred at the point when the cup reached the mouth. The extent of pronation ramps up and then down during the movement from cup grasp to mouth, with the forearm maintaining a neutral orientation throughout other areas of the task. A small wrist flexion occurs while grasping the cup, and transitions to extension when the cup reaches the mouth. The

![Figure 35: Movement profile (black line) and trajectory map of desired wrist pronation/supination (rho) at key points in Cup Task for the Robot-only validation. The color scale represents the angular deviation of the wrist (pronation (-) and supination (+)) in radians within the horizontal (x-y) plane at specific heights in the workspace (z). The white asterisk denotes the position of the wrist within the overall trajectory for the xy-plane of the trajectory map shown. The white box outlines the Home region.](image-url)
device provides primarily flexion as the user positions their wrist farther from the mouth (i.e., for smaller values along the y-dimension). Wrist extension is most prominent for larger x-y positions as the cup approaches the mouth.

In the pitcher task, four key points were identified: grasping the pitcher, reaching peak pour position, returning the pitcher to the table, and returning to the home position. As shown in Figure 37, wrist pronation was largest during peak pour. The wrist maintained a flexed orientation throughout the duration of the task, shown in Figure 38, and was largest during pouring. These profiles were consistent with the task; to complete the pour the pitcher needed to be tilted (provided by pronation) and the spout aligned with the vessel it was pouring into (provided by flexion). The task was dominated by a pronated and flexed orientation, only showing signs of extension when the wrist moved to higher values in the x-direction and y-direction. This would correspond to an altered approach where the subject held the pitcher much closer to their body,
Figure 37: Movement profile (black line) and trajectory map of desired wrist pronation/supination (rho) at key points in Pitcher Task for the Robot-only validation. Labeling conventions are the same as in Figure 35.

Figure 38: Movement profile (black line) and trajectory map of desired wrist flexion/extension (theta) at key points in Pitcher Task for the Robot-only validation. Labeling conventions are the same as in Figure 36.
and pour with the spout pointed more forward, as opposed to from the side. Due to the altered position of the pitcher relative to the receiving vessel, this required the change in wrist orientation.

The spoon task featured three key points; moving spoon to bowl, moving spoon to mouth, and returning home. Pronation was the favored rotation of the forearm, peaking as the spoon reached the mouth (Figure 39). The wrist flexion/extension (theta) changed from a flexed state, as the spoon was scooped from the bowl, to an extended state as it approached the mouth. The mapping also allow for theta to assume an extended state if the wrist is positioned closer to the subject while attempting to scoop from the bowl. This can be seen in Figure 40 as the blue region, representing flexion, exists farther away from the subject’s position, while the red region, representing extension, can be found much closer to the subject.

**Figure 39:** Movement profile (black line) and trajectory map of desired wrist pronation/supination (rho) at key points in Spoon Task for the Robot-only validation. Labeling conventions are the same as in Figure 35.
Figures 41 and 42 show the trajectory map of the first cone and beginning transport of the second cone in the Cone Task. Since this task is repetitive in nature, the movement profiles used to move each cone are very similar. Showing the movement of the first cone was a good representation for the movement of all the cones. Consistent with the task, the wrist maintained a pronated state throughout the movement while flexion/extension remained relatively neutral during the picking up and setting down of the cone. Travel between cones featured an extended state at the wrist that helped maintain neutral positioning of the hand as the elbow rotated between the cone locations.

Figures 43 and 44 show the trajectory map associated with the placement of the first three pegs in the Peg Task. Throughout the task, the wrist maintained a pronated state, only returning to a more neutral orientation as the wrist left or approached the home position (Figure 43). The wrist flexion/extension map showed a clear switch between a flexed state while picking up the peg to
**Figure 41:** Movement profile (black line) and trajectory map of desired wrist pronation/supination (rho) at key points in Cone Task for the Robot-only validation. Labeling conventions are the same as in Figure 35.

**Figure 42:** Movement profile (black line) and trajectory map of desired wrist flexion/extension (theta) at key points in Cone Task for the Robot-only validation. Labeling conventions are the same as in Figure 36.
Figure 43: Movement profile (black line) and trajectory map of desired wrist pronation/supination (rho) at key points in Peg Task for the Robot-only validation. Labeling conventions are the same as in Figure 35.

Figure 44: Movement profile (black line) and trajectory map of desired wrist flexion/extension (theta) at key points in Peg Task for the Robot-only validation. Labeling conventions are the same as in Figure 36.
an extended state when the peg was placed into the board. Regardless of height, the mapped wrist orientations remained similar.

5.4.1 Standalone Device Performance

Tests of active control using the orthosis by itself (without the influence of a subject’s limb) were performed to independently evaluate the baseline performance characteristics of the system. This allowed deficiencies in the device and closed-loop control to be parsed from deficiencies resulting from user interaction and adaptation during active compensation. During testing, three factors were examined to demonstrate fast and reliable assistance by the orthosis across a range of movements; time required for the device to respond to an error is wrist orientation, accuracy aligning the wrist to the desired orientation, and system bandwidth to accommodate various movement frequencies.

Results for the response time tests times ranged 7-26 ms across tasks. Because this delay spanned such a short time interval, its impact on providing accurate compensation was interpreted to be minimal. Interestingly, a lack of variability was observed between response times, both within and between tasks. Rho and theta were assessed separately, as they operated on separate control loops, resulting in separate delays. Performing a 95% confidence interval revealed that rho could be expected to fall between 9.07 ± 4.24 ms, and theta would follow suit at 17.53 ± 7.7 ms. Since subjects were not used for this part of the validation, delay contributors were narrowed to computation, signal transmission, and command execution. The fact that both actuated joints showed delays in the tens of milliseconds justifies the use of a CAN bus and cabling in signal transmission and the small magnitude for changes in orientation justifies the high sampling rate. CAN was selected for its high data rates, in order to minimize signal latency, while the large difference in magnitude between sampling frequency and low movement frequency for each task makes it very easy to command tiny positional changes for each sample.
Overall, response accuracy was defined by average movement errors of less than 1° (except for pronation/supination during the Peg task). Reported errors influenced by the loop delay, and “aligned” based on the movement response delay, revealed a greater impact on the error as the delay grew larger. In addition to the conditions responsible for the delay listed above, one consideration that was not mentioned was the interaction between the low torque, high speed nature of the motor, and large gearing that was applied to it in order to increase the desired torque output. While the rotor inertia of the motors (2.39 g*cm²) provides little resistance to movement, the gearing (24:1 for pronation/supination, 53:1 for flexion/extension) amplifies the resistance, requiring more revolutions from the motor to drive to the commanded orientation. Since this will always be present regardless of the movement speed or direction, the gearing will have a larger impact on delay and response accuracy.

Two methods were used to characterize the bandwidth of the movement, using the -3 dB cutoff frequency and 90% of the cumulative spectral power. The 90% method appeared much more sensitive to changes in movement frequency, but also gave magnitudes that were much higher. The -3 dB method showed more subtle changes, but was objectively found using a qualitative means to determine the -3 dB cutoff point of the signal power spectrum. There was an interesting result within the spoon task where the frequency bandwidth under the fast condition came out smaller than the bandwidth using “normal” speed. Since bandwidth determination using the -3 dB was an estimation made by objective means, it is expected results will vary. The slow and fast conditions were also created as a 30% increase or decrease in time to perform the task, based on the normal speed used. This change in speed was most likely not significant enough to create a noticeable change in movement frequency. The bandwidth frequencies were as expected though, as the cutoff frequency for the bandwidth never exceeded 1.5 Hz for any task. It was previously determined that the performance of ADLs should not require more than a 2 Hz movement [32].
5.5 Subject Validation

The performance of the orthosis when the limb was present was validated by having normal-functioning subjects perform each task under three different conditions; normal positioning by the subject and allowing robot guidance, normal positioning while co-contracting the forearm muscles, and using abnormal approaches to the initial task object (e.g., a narrow approach with the arm moving tight to the body, or a wide approach with arm extended further from the body). The experiment setup was the same as that used in the normative trajectory development, with the exception that the orthosis was operated in an active mode. Each subject perform up to 20 iterations for the robot guidance and co-contraction conditions, and 5 iterations for each alternate approach. Between each iteration, the subject was directed to pause for 2-3 seconds before proceeding to the next iteration. The acceptance criteria awarded a pass or fail grade for each task iteration, based on whether or not 95% of the error distribution fell below the 5° error criterion.

Each condition was evaluated for absolute error over the course of a run, the torque and current commanded to the motors, and the proportion of test iterations that were able to achieve the acceptance criteria. These characteristics served as the validation metrics to show that the device could accurately apply compensation when needed in response to the combined inertial dynamics of the device and the limb.

5.5.1 Condition #1: Robot Driving Wrist Orientation

In this condition, each subject kept their wrist loose while moving their limb normally for completion of the task, causing the orthosis to actively influence the necessary orientation of the wrist. This was done to confirm that the device could provide the proper compensation with the added inertial dynamics of the limb, and simulated a case where a user might lack the muscle tone or control necessary to complete these tasks. Table 9 shows the average wrist orientation error.
Table 9: Average absolute error for the Robot Guidance validation across tasks. In addition of average error, the absolute error at 2 standard deviations above the mean is also provided. The percent of iterations where 95% of the error distribution was below 5° is also specified. Errors in wrist orientation are expressed in radians for pronation/supination and flexion/extension.

<table>
<thead>
<tr>
<th>ROBOT GUIDANCE CONDITION</th>
<th>Pronation/Supination</th>
<th>Flexion/Extension</th>
</tr>
</thead>
<tbody>
<tr>
<td>Avg Error</td>
<td>+2σ</td>
<td>% &lt; 5°</td>
</tr>
<tr>
<td>Pouring from Cup</td>
<td>0.56</td>
<td>2.12</td>
</tr>
<tr>
<td>Pouring from Pitcher</td>
<td>0.67</td>
<td>1.93</td>
</tr>
<tr>
<td>Eating with Spoon</td>
<td>1.04</td>
<td>3.41</td>
</tr>
<tr>
<td>Stacking Cones (Approach 1)</td>
<td>0.29</td>
<td>0.45</td>
</tr>
<tr>
<td>Stacking Cones (Approach 2)</td>
<td>0.25</td>
<td>0.35</td>
</tr>
<tr>
<td>Stacking Cones (Approach 3)</td>
<td>0.80</td>
<td>2.23</td>
</tr>
<tr>
<td>Placing Pegs</td>
<td>0.88</td>
<td>2.74</td>
</tr>
</tbody>
</table>

Figure 45: Active compensation performance for wrist pronation/supination (Left) and flexion/extension (Right) during the Cone Task (Approach 1). (Top) Averaged absolute error across iterations. The dashed line (Top) represents the 95% confidence interval for the mean absolute error. Vertical lines indicate key instances within the task performance (blue=picking up cone, red=setting cone down). (Bottom) Averaged commanded torque vs. current across iterations.
across task cycle and iterations, and the corresponding standard deviation in the distribution of average errors across iterations. When evaluated with respect to the proportion of iterations that passed the acceptance criteria, the device performed very well during the cup, pitcher, and cone tasks with average wrist orientation errors all under 1 degree. Figures 45 and 46 illustrate the compensation performance during approach 1 for the Cone Task. The high degree of overlap between the torque and current traces, and limited motor saturation indicate minimal error. There are small differences in torque and current that do occur, but they do not necessarily reflect the ability of the device to provide compensation. Momentary spikes in error that are present are on the order of milliseconds, and with the inherently slow movements used with the therapy tasks, these instances will not be perceived by the subject. This was used as a primary consideration when the acceptance criteria was selected.

**Figure 46:** Percentage occurrence of motor saturation for wrist pronation-supination (Left) and flexion/extension (Right) during the Cone Task (Approach 1).

**Figure 47:** Active compensation performance for wrist pronation-supination during the Spoon Task. (Left) Average absolute error across iterations. The dashed line represents the 95% confidence interval for the mean absolute error. Vertical lines represent key instances within the task performance (blue=spoon at bowl, red=period of spoon at mouth). (Right) Average commanded torque vs. current across iterations.
A fall in accepted iterations was observed for the spoon and peg tasks, specifically for wrist pronation/supination, corresponding to a higher average and range of orientation errors for these tasks. Figure 47 shows the performance characteristics during the spoon task across subjects. A large increase in error occurred at 0.3 within the normalized movement profile and again between 0.5-0.6, which coincided with a disconnect between commanded torque and the current applied to the motor. Under ideal operating conditions, the commanded torque and applied current should match, indicating that the operation of the motor produced the commanded torque. The torque vs. current plot in Figure 47 shows a large dissociation between the traces, and an inability to meet the torque demand through most of the task. Throughout the movement cycle, the percentage of instances where the motor reached saturation aligned with the changes in the torque plot, and is featured in Figure 48. In this case, the error in wrist orientation was exacerbated by the inability of the motor to provide the required corrective torque. Similar deficiencies in active wrist compensation occurred during the Peg Task. Errors in wrist orientation were most prominent early on in the task cycle (0.1, 0.3) and again at the end (0.9-1) (Figure 49). The highest number of saturation instances occurred at the times where the disconnect between torque and current was greatest (Figure 50). Iterations that failed the acceptance criteria were observed in both subjects, and no pattern emerged between instances of higher error to specific positions in the task space.
Across subjects and tasks, periods of increased position error appear intermittently during active control. The periods coincided with changes in orientation direction and were characterized by a delayed change in the measured orientation following commanded orientation changes that lasted 1-2 seconds before achieving the commanded orientation. Points along the movement path where direction change occurs correspond to points where the greatest torque is needed. Initiating an opposing movement requires an initial jump in angular acceleration until the necessary movement velocity is achieved. The proportional relationship between torque and angular acceleration means that larger torques are needed initially. If the torque capabilities of the orthosis are not capable of meeting the torque required to change the movement direction, the applied torque and resultant angular velocity of the wrist saturate until the required torque falls

**Figure 49:** Active compensation performance for wrist pronation/supination during the Peg Task (Left) Average absolute error in wrist pronation/supination across iterations. The dashed line represents the 95% confidence interval for average absolute error. Vertical lines represent key instances within the task performance (blue=picking up cone, red=setting cone down). (Right) Average commanded torque versus current in wrist pronation/supination.

**Figure 50:** Percentage occurrence of motor saturation during wrist pronation/supination in the Peg Task.
below the maximum torque ability of the orthosis, and the system “catches up”. The intermittent nature of this effect indicates that the instantaneous torque requirement for movement can vary greatly based on the task (position of the limb, and the rate of change in orientation), and the subject demographics (i.e. gender).

5.5.2 Condition #2: Co-contraction of Limb During Task Performance

In the second condition, subjects were asked to contract the muscles in their forearm and wrist while moving their limb through the workspace to complete the task. Like the first condition, the orthosis needed to influence the orientation of the wrist and overcome the added resistance due to the co-contraction. This condition was intended to stress the orthosis’ active compensation and simulate an impaired subject with rigidity or spasticity in the limb to evaluate how well the orthosis would be able to assist these subjects. The orthosis performance during co-contraction is shown in Table 10. Performance varied across tasks with the best performance for pronation/supination (>80% of iterations below the error criterion) occurring for the cup, pitcher, and 2 of the 3 approaches for the cone task. The spoon, peg, and approach #3 of the cone task were less accurate with less than 56% of iterations achieving the 5° average error criterion. Active

Table 10: Average absolute error for the Co-contraction Condition across tasks. In addition of average error, the absolute error at 2 standard deviations above the mean is also provided. The percent of iterations where 95% of the error distribution was below 5° is also specified. Errors in wrist orientation are expressed in radians for pronation/supination and flexion/extension.

<table>
<thead>
<tr>
<th>CO-CONTRACTION CONDITION</th>
<th>Pronation/Supination</th>
<th>Flexion/Extension</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Avg Error</td>
<td>+2σ</td>
</tr>
<tr>
<td>Pouring from Cup</td>
<td>0.63</td>
<td>1.62</td>
</tr>
<tr>
<td>Pouring from Pitcher</td>
<td>0.73</td>
<td>2.13</td>
</tr>
<tr>
<td>Eating with Spoon</td>
<td>5.57</td>
<td>15.17</td>
</tr>
<tr>
<td>Stacking Cones (Approach 1)</td>
<td>0.35</td>
<td>0.62</td>
</tr>
<tr>
<td>Stacking Cones (Approach 2)</td>
<td>0.32</td>
<td>0.53</td>
</tr>
<tr>
<td>Stacking Cones (Approach 3)</td>
<td>1.66</td>
<td>4.88</td>
</tr>
<tr>
<td>Placing Pegs</td>
<td>1.40</td>
<td>3.39</td>
</tr>
</tbody>
</table>
compensation of flexion/extension remained high across tasks (>93%) with the exception of the peg task (78% criterion performance across iterations). Relative to condition #1, active compensation was less accurate during co-contraction. This is not necessarily surprising due to the increased resistance about the wrist, which would require larger torques to achieve the same error performance. Active compensation of wrist pronation/supination was worst for the spoon task. Analysis of the error time course showed that wrist position errors were most prominent from 0.3-0.9 of the task iteration and coincided with discrepancies between commanded torque and motor current. Figure 51 shows a strong similarity between increase in movement error, the torque demand at the motor, and a discrepancy between torque demand and current output. The

**Figure 51:** Active compensation performance for wrist pronation/supination during co-contraction in the Spoon Task. (Left) Average absolute error in wrist pronation/supination across iterations. The dashed line represents the 95% confidence interval for average absolute error. Vertical lines represent key instances within the task performance (blue=spoon at bowl, red=period of spoon at mouth). (Right) Average commanded torque versus current in wrist pronation/supination.

**Figure 52:** Percentage occurrence of motor saturation during wrist pronation/supination (with co-contraction) in the Spoon Task.
number of instances that the motor saturated increased as the difference between torque and current grew (Figure 52). In this case, the change in number of saturation instances did not align as closely with the torque-current difference. Figure 52 shows two distinct peaks where saturation was most prominent during task performance. However, when comparing against the torque-current plot in Figure 51, the corresponding torque peaks show clearly different relative magnitudes. When the motor reaches saturation, it is due to the commanded torque exceeding the torque capabilities of the motor. Once this threshold is exceeded, commanded torque can be any value that results in a saturation condition. This increases the variability in commanded torques observed at saturation. Commanded torques occurring around 0.6 of the cycle were higher than those around 0.3 in the cycle, and the errors showed this same relationship. Since the instances of saturation were similar, the greater discrepancy between torque and current at 0.6 was not due to more frequent error, but by a more severe error at that stretch in the iteration cycle.

The peg task was the only task in which more than 10% of iterations failed the acceptance criteria for both pronation/supination and flexion/extension. Orientation errors were most prominent in the beginning of the task (at 0.1) (Figure 53 and 55). There was also an

![Figure 53: Active compensation performance for wrist pronation/supination during co-contraction in the Peg Task. (Left) Average absolute error in wrist pronation/supination across iterations. The dashed line represents the 95% confidence interval for average absolute error. Vertical lines represent key instances within the task performance (blue=picking up peg, red=placing peg in board). (Right) Average commanded torque versus current in wrist pronation/supination.](image-url)
**Figure 54:** Percentage occurrence of motor saturation during wrist pronation/supination (with co-contraction) in the Peg Task.

**Figure 55:** Active compensation performance for wrist flexion/extension during co-contraction in the Peg Task. (Left) Average absolute error in wrist flexion/extension across iterations. The dashed line represents the 95% confidence interval for average absolute error. Vertical lines represent key instances within the task performance (blue=picking up peg, red=placing peg in board). (Right) Average commanded torque versus current in wrist flexion/extension.

**Figure 56:** Percentage occurrence of motor saturation during wrist flexion/extension (with co-contraction) in the Peg Task.
increase in error late in the task iteration (at 0.9) for pronation/supination. These error increases correspond to motor saturation. The number of saturation instances peaked at around 20% for Rho (Figure 54) and 8% for Theta (Figure 56) in proportion to the total number of instances performed, suggesting that saturation rates this low could still have a substantial effect on the acceptance rate of the performed iterations. Previous examples from the spoon and peg tasks showed saturation rates upwards of 50-60% resulting in lower acceptance rates. Performance of the peg task between the robot driving and co-contraction conditions showed a greater number of instances along the iteration cycle of occurring saturation for co-contraction, but the proportion of instances that saturated was very similar between the conditions.

The third approach for the cone task saw a much higher number of failed iterations in comparison to the first two with co-contraction and the robot driving condition. This approach was characterized by a consistent change in pronation/supination, alternating between a neutral orientation to 60°-90° pronation as each cone was grabbed and moved. Four distinct error peaks occurred during the movement corresponding to the instances of greatest pronation (Figure 57). These peaks in error were characterized by large commanded torques and a corresponding increase in the number of instances where the torque output of the motor saturated. The

Figure 57: Active compensation performance for wrist pronation/supination during co-contraction in the Cone Task (approach #3). (Left) Average absolute error in wrist pronation/supination across iterations. The dashed line represents the 95% confidence interval for average absolute error. Vertical lines represent key instances within the task performance (blue=picking up cone, red=setting cone down). (Right) Average commanded torque versus current in wrist pronation/supination.
combination of frequent changes in orientation and the increased joint resistance during co-contraction produced an environment that challenged the torque generating capacity of the motors.

5.5.3 Condition #3: Effect of the Use of Atypical Approaches on Device Performance

In the third and final condition, each subject was asked to use a narrow (tight to the body) and wide (stretched out away from the body) approach to reach the task objects. This was applied to the simpler tasks (drinking from cup, pouring from pitcher, eating with spoon) that featured a single cyclic motion. The objective was to stress the trajectory control features of the device by using atypical movement profiles that simulated the use of compensatory strategies by the user. By not moving the limb through the normally mapped space, this condition examined the ability of the orthosis to influence the wrist (and limb) to take a “normal” path. Wrist orientation errors were largest with both approaches in the spoon task, and in all tasks when using the wide approach (Table 11). The ability of the orthosis to compensate for deviations in wrist flexion/extension for atypical paths is notable. For the alternate approaches employed in the tasks, deviations in wrist flexion/extension were the most likely to occur as subjects adjusted to grasp the object based on the approach.

**Figure 58**: Percentage occurrence of motor saturation during wrist pronation/supination (with co-contraction) in the Cone Task (approach #3).
Table 11: Average absolute error for the Alternate Approach Conditions across tasks. In addition of average error, the absolute error at 2 standard deviations above the mean is also provided. The percent of iterations where 95% of the error distribution was below 5° is also specified. Errors in wrist orientation are expressed in radians for pronation/supination and flexion/extension.

<table>
<thead>
<tr>
<th></th>
<th></th>
<th></th>
<th>Rho</th>
<th></th>
<th></th>
<th>Theta</th>
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<tbody>
<tr>
<td></td>
<td>Avg Error</td>
<td>+2σ</td>
<td>% &lt; 5°</td>
<td>Avg Error</td>
<td>+2σ</td>
<td>% &lt; 5°</td>
<td></td>
<td></td>
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<tr>
<td></td>
<td>Narrow</td>
<td></td>
<td></td>
<td>Wide</td>
<td></td>
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</tr>
<tr>
<td>Pouring from Cup</td>
<td>1.11</td>
<td>4.96</td>
<td>87%</td>
<td>0.57</td>
<td>0.92</td>
<td>100%</td>
<td></td>
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<tr>
<td>Pouring from Pitcher</td>
<td>1.87</td>
<td>7.60</td>
<td>73%</td>
<td>0.53</td>
<td>0.97</td>
<td>100%</td>
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</tr>
<tr>
<td>Eating with Spoon</td>
<td>1.62</td>
<td>4.89</td>
<td>50%</td>
<td>0.76</td>
<td>1.97</td>
<td>90%</td>
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<tr>
<td></td>
<td>Wider</td>
<td></td>
<td></td>
<td></td>
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<td></td>
<td></td>
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</tr>
<tr>
<td>Pouring from Cup</td>
<td>1.07</td>
<td>2.44</td>
<td>47%</td>
<td>0.39</td>
<td>0.67</td>
<td>100%</td>
<td></td>
<td></td>
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</tr>
<tr>
<td>Pouring from Pitcher</td>
<td>4.16</td>
<td>17.85</td>
<td>47%</td>
<td>0.55</td>
<td>0.81</td>
<td>100%</td>
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<tr>
<td>Eating with Spoon</td>
<td>3.03</td>
<td>7.27</td>
<td>20%</td>
<td>0.72</td>
<td>1.46</td>
<td>100%</td>
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<td></td>
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</table>

Figure 59: Active compensation performance for wrist pronation/supination during alternate approaches in the Spoon Task. (Top) Average absolute error in wrist pronation/supination across iterations for the narrow approach (Left) and wide approach (Right). The dashed line represents the 95% confidence interval for average absolute error. Vertical lines represent key instances within the task performance (blue=spoon at bowl, red=period of spoon at mouth). (Right) Average commanded torque versus current in wrist pronation/supination for the narrow approach (Left) and wide approach (Right).
Figures 5 and 60 show the changes in absolute error, commanded torque vs. applied current, and instances of saturation of the motor for wrist pronation/supination during the spoon task with narrow and wide approaches. Periods of greatest error occurred later in the task cycle and were more prolonged with the wide approach path. Instances of saturation for motor torque matched periods in which the applied current did not meet the commanded torque. The likelihood

Figure 60: Percentage occurrence of motor saturation during wrist pronation/supination (with alternate approaches) in the Spoon Task.

Figures 59 and 60 show the changes in absolute error, commanded torque vs. applied current, and instances of saturation of the motor for wrist pronation/supination during the spoon task with narrow and wide approaches. Periods of greatest error occurred later in the task cycle and were more prolonged with the wide approach path. Instances of saturation for motor torque matched periods in which the applied current did not meet the commanded torque. The likelihood

Figure 61: Active compensation performance for wrist pronation/supination using wide approach for Cup and Pitcher tasks. (Top) Average absolute error in wrist pronation/supination across iterations for the Cup (Left) and Pitcher (Right) tasks. The dashed lines represent the 95% confidence interval for average absolute error. (Bottom) Average commanded torque versus current in wrist pronation/supination for Cup (Left) and Pitcher (Right) tasks. Vertical lines represent key instances within the task performance, (Left) grasp cup, cup at mouth, return cup, (Right) grasp pitcher, peak pour, return pitcher.
of saturation varied within the task cycle, exceeding 50% during the period of pronation/supination associated with orienting the spoon at the mouth.

Poor wrist compensation during pronation/supination was also observed in the cup and pitcher tasks when a wide approach path was used. The use of a wide approach was especially troublesome during the pitcher task, causing prolonged orientation errors throughout the task cycle. For both tasks, instances of saturation increased to nearly 40% (Figure 62), corresponding to mismatches between the commanded torque and current applied to the motor.

5.5.4 Subject-Influenced Device Performance

5.5.4.1 Task and Actuation-based Error

Across all conditions, there was one consistent contributor to the elevated levels of error observed. The reduced operational bandwidth of the motor-driven pronation/supination created an environment in which the torque demands of the system exceeded the capabilities of the motor, and prevented proper correction of error. This was seen in all the tasks and conditions, with instances of greatest error accompanied by the greatest proportion of saturation occurrences. Within tasks and conditions, this relationship varied due to decoupling when the motor output reached saturation.

Much of the association between commanded torque and orientation error can be explained by the squared relationship between them. Changes in torque are most directly

Figure 62: Percentage occurrence of motor saturation during wrist pronation/supination (with wide approach) in the Cup task (Left) and Pitcher task (Right).
influenced by changes in angular acceleration, or the second derivative of the limb orientation. When the torque capability of the motor is exceeded, the squaring causes a more pronounced error in wrist orientation. This results in an “uncoupling” between the wrist orientation error and the corrective torque applied by the system, and corresponds to periods where the torque and current traces did not overlap as shown in the torque-current plots (Section 5.5.3). During these periods, changes in error did not result in changes to the corrective response resulting in commanded torques that were often unrealistic. During validation, these conditions persisted until there was a change in orientation direction. As the joint moved in the opposite direction, control was restored, or “recoupled” to the rest of the system, and normal operation resumed. Thus the cyclic nature of the tasks provide opportunities for the control system to “catch-up”. These results indicate that a larger motor (with greater torque) would be helpful to eliminate control uncoupling during pronation/supination.

Comparisons of performance between conditions showed a clear degradation in performance between normal robot guidance and instances of co-contraction for both pronation/supination and wrist flexion/extension. Similar trends were observed across tasks; the spoon task showed the highest error followed by placing pegs, stacking cones (approach 3), and pouring from a pitcher. The similarities in task performance between the conditions suggests a secondary, task-based influence on the performance of the device. As shown in Figure 51, the largest error during the spoon task (co-contraction) occurred between 0.5-0.7 of the iteration cycle. This corresponded to a large difference in commanded pronation/supination orientation which occurred at the same point in the cycle (Figure 63). During the same period, differences were observed in wrist position along the x and z-axes. As a result of co-contraction, rigidity of the limb was increased and the movement profile was altered. Since position changed, an orientation was commanded that did not align well with the measured orientation of the forearm. The torque needed to overcome the rigidity of the limb and correct for the orientation mismatch exceeded the capability of the motor, and resulted in the large reported error. The differences in
movement profile can also be attributed to the less constrained nature of the spoon task. Outside of the bowl and the subject’s mouth, subjects were free to take any path, leading to greater variability in movement profiles. A similar case can be made for the worse performances seen for the cone and peg tasks, but due to its more constrained nature. Though the controlled nature of these tasks provide fewer opportunities for variability in the movement path, the increased complexity of the tasks induce errors through increased frequency of movements, and greater quantity of different movements.

The alternate approach condition showed an especially high proportion of failed iterations. This revealed a limitation to the control system, as movement profiles similar to the measured profiles on which the position-to-orientation mapping is based, receive an appropriate reaction from the device (robot guidance, co-contraction conditions) by interpolating between the mapped points. When subjecting the device to an unfamiliar profile, the control system applies an orientation of zero, which was set as a default placeholder for locations in the map volume not normally used while performing a task. Because of this assignment, the limb is forced to stay neutral, regardless of the movement path taken through this region. Any attempt by the subject to change their orientation would be met by resistance from the system and manifest as an increase in orientation error. In order for this increase in error to be avoided, the commanded orientation

**Figure 63:** (Left) Positional changes of the Wrist for the Spoon Task. Changes shown in x (blue), y (red), and z (green). Positions are compared between Robot Guidance (solid) and Co-contraction (dashed). (Right) Changes in Commanded and Measured Rho for the Spoon Task. Commanded (blue) and measured (red) values for Rho are shown for Robot Guidance (solid) and Co-contraction (dashed).
needs to create a situation where the subject is forced to alter their limb position to achieve the orientation needed to complete the task. One consideration for facilitating this would be replacing the “smoothing” of the map with extreme orientations at positions outside the normal movement path, so that the subject is forced to change their position in order to complete the task. This form of “error inducing” therapy applies greater error as the subject deviates further from the desired path. This forces the subject to search for the desired path to minimize the error in their movement. This approach could be easily implemented within this system. In the case of the cup task where a wide approach was used, the wrist could be subjected to a more extreme level of extension as the subject approaches the cup. This would force the subject to bring their limb closer to their body in order to obtain a more natural orientation at the wrist.

Based on the results, it was clear that the underpowered nature of the motor driving pronation/supination of the wrist was the primary catalyst for the number of failed iterations observed. As limb resistance was increased in the co-contraction and alternate approach conditions, performance of the device suffered as the required corrective torques exceeded the capabilities of the motor. The performance of a properly powered motor was evident in wrist flexion/extension, wherein 15 of the 20 possible task and condition combinations resulted in all iterations passing the acceptance criteria, and 19 of 20 combinations resulted in > 90% of iterations passing the acceptance criteria. This supports the notion that the addition of a more powerful motor for the control of pronation/supination should significantly improve overall device performance. Additional contributors to reduced performance included movement constraints and abnormal approaches; the transition to an “error inducing” map would help accommodate a variety of approaches and help further constrain each task, reducing the observed error.
5.5.4.2 Perceived Delay

The impact of system delay, characterized by the time required for computation and signal transmission between command and execution of orientations, on device performance was examined in Section 5.4.1. An additional delay that was excluded from this analysis resulted from the low-pass filter used during signal conditioning of the inputs from the analog potentiometers used to calculate wrist position. Electrical noise due to cross talk between the potentiometers, Hall sensors, and encoders, and the length of the wiring harness made it difficult to accurately calculate the wrist position. Despite efforts to isolate the signal paths through shielding, a digital low pass filter was required to attenuate the remaining noise and obtain an acceptable signal to noise ratio. The implemented filter used a narrow pass band ($f_c = 5$ Hz), and sharp stop band attenuation (-60 dB/dec) to obtain a clean signal. As a result, the group delay associated with this filter was found to be 353ms. At this length, it was expected that the delay would have an effect on the ability of the device to provide real-time compensation within the closed-loop control system, based in part on the frequency of movement. Since this effect could not be measured explicitly, the relationship between measured error and movement changes during task performance were examined.

The colored vertical lines in Figures 45-61 denote key instances of movement associated with the respective tasks. These instances also break the tasks into natural trajectory segments. If movement along a segment is rapid, or orientation changes rapidly over a short distance, the added delay in the control path will command past values that are potentially much different from the present command orientation required. The effect is exacerbated within movement segments that encounter a large change in the orientation gradient within the trajectory map. This was especially true with the spoon task (Figures 47, 51, and 59). In the respective plots, the blue line indicates the instance in the cycle when the spoon reached the bowl, and the red lines bookend the period during which the spoon is at the mouth. Interestingly, during both robot driving and co-
contraction conditions, error peaked during the movement from the bowl to the mouth, and again once the spoon reached the mouth. Error peaks during the alternate approach condition appear prominently only when the spoon reached the mouth. The corresponding trajectory map (Figure 39), shows a rapid increase in pronation as the wrist moves from the bowl to the mouth. The rapid change, coupled with the increased loop delay from signal conditioning, is a likely contributor to the errors created at these locations. This was also evident during the peg task across conditions, illustrated in Figures 49, 53, and 55. Blue lines indicate instances where a peg was picked up, while the red lines indicate instances when the pegs were placed into the board. Error appeared most prominently while picking up the first two pegs. In the corresponding trajectory map (Figure 43), a change in pronation occurs when the first two pegs are picked up. When picking up the third peg, pronation stays relatively consistent, and relates to negligible error seen during this period of the task. Again, it seemed the combinatorial effect of the gradient and the signal conditioning delay contributed to the increased control error observed during the task.

An example of the impact of the signal conditioning delay on system performance is illustrated in Figure 57, and Figures 64 and 65 below. During performance of the cone task with co-contraction, the use of approach #3 resulted in larger errors compared to the other approaches. All of these errors occurred during the period between when a cone was picked up (blue lines) and when it was set back down (red lines). To test whether the promotion of a different movement strategy could create the higher error during approach #3, the positional movements, broken down into x, y, and z components, were compared between approaches 1 and 3. Figure 64 shows that movements along x and y aligned well with one another, while in z, a clear difference exists. The range of z-values represented in Figure 65 are responsible for the difference in orientation gradients that exist for each approach. Based on this figure, it is clear a much larger gradient exists for approach #3, based on the greater change along the z-axis and within the x-y
Figure 64: Wrist position during the co-contraction condition for Approaches 1 and 3 of the Cone Task. (Top) Changes in X (Left) and Y (Right). (Bottom) Changes in Z. The blue line represents Approach 1, where 100% of iterations passed the acceptance criteria, and the red line represents Approach 3, where only 51% of iterations passed the acceptance criteria.

Figure 65: Trajectory maps for the Cone Task with Approach 1 (Top) and Approach 3 (Bottom). The color maps for each approach represent the range along the z-axis covered while transporting each cone. The average movement profile across iterations for each approach is represented by the black trace on the map.
plane. As with the spoon and peg tasks, this gradient was likely a contributor to the error spikes observed during transport of each cone.

The results suggest that the combination of the orientation gradients created within the mapped volume, coupled with the signal conditioning filter delay, contributed to the periods of large error observed during subject validation. While it is possible that error could have been subject-initiated, the effort to perform each task “normally” by each subject makes it unlikely that it is the primary source. While the underspec’d nature of the motor providing assistance for pronation/supination appears to be the primary contributor to performance errors in the system, system delays during signal conditioning provided a second compounding source of error. By retuning or removing the signal conditioning filter from the control loop, error in active compensation of task performance can be decreased.
6 CONCLUSIONS AND FUTURE DIRECTIONS

Presented was the development and validation of an active wrist orthotic capable of assisting pronation and supination of the forearm, and flexion and extension of the wrist. A mapping-based closed-loop control system was designed using data collected from normally functioning subjects performing a series of five ADL tasks. The maps established a set of “nominal” task-based wrist orientations that could be used to compensate, in real-time, the user’s wrist orientation throughout the workspace. The performance of the orthosis during active control was evaluated both with and without a user wearing the device, examining factors such as response time, accuracy, speed, position variability, and distal influence. Aside from the performance issues resulting from the signal conditioning delay and use of an underpowered motor for pronation/supination, the device was able to successfully provide a desired wrist orientation based on the subject’s limb position in space.

6.1 Device Evaluation

As previously outlined in Section 3.1, the overall design addressed requirements associated with interactions between the user, device, and task environment. With respect to user/device interactions, the system was required to be durable, form-fitting, size adjustable, backdriveable, and to provide ranges of motion adequate for completing ADL tasks. Implementing aluminum components at the orthotic-passive arm interface helped to maximize the durability of the device. Though never subjected to harsh, abusive movements, the orthosis withstood use from over 20 different subjects without issue. To accommodate a wide range of subject limb/hand sizes, three extension links of differing lengths were created and different sized glove attachments were developed. While the gimbal places an upper limit on hand size, only one individual was unable to fit their hand into the orthosis during evaluation. Since the intended demographic for use is pediatric subjects, who generally have smaller limbs, this restriction is not expected to limit usage within the target population. The trough and hand brace conformed well
to the shape of the user’s arm, and the orthosis was easily manipulated by users throughout the workspace. There was widespread acceptance of the movement abilities provided by the orthosis across all subjects once they had been given a period to acclimate to the device. Some subjects found that the Velcro straps anchoring the device to the forearm produced discomfort after continued use, but this was usually remedied by wearing a long sleeved shirt, or placing a piece of fabric between the arm and straps.

Two requirements in particular were paramount for addressing interactions between the device and the task environment. These included the ability to perform all tasks without interference from the device and the application of active compensation by the device only when an improper task-based movement occurred. The open nature of the hand brace and actuated linkage was designed to keep the hand free and open to facilitate grasp and manipulate the fingers. All subjects tested were able to grasp and manipulate objects and none reported difficulty performing the tasks. The trajectory mapping approach defined nominal, task-specific wrist orientations at each location in the workspace, which were used within the closed-loop control system to determine the amount of assistance to provide at the wrist. While motor actuation was always active in the active control configuration, it only became apparent to the user when the wrist trajectory deviated from the nominal range for the task. A user that performed a task normally triggered a small assistive torque, but was relatively undetectable due to the small error produced.

6.2 Advantages and Disadvantages of a Task-based, Position-controlled Assistive Therapy Device

The two primary aspects of the proposed device that set it apart from other assistive therapy devices were its usability in task-based therapy, and the position-based control algorithm that provides task-specific assistance within the workspace independent of the subject’s speed of movement. Currently, most devices used in robot-mediated therapy are geared to simple,
repetitive motions that do not readily facilitate physical therapy tasks requiring object manipulation. The actuated wrist orthosis developed here deviates from this paradigm with a design that enables full, unobstructed use of the hand and can be used to perform ADLs.

By emphasizing object manipulation in the orthosis design, the variety of therapy usage scenarios increases considerably. Incorporating a task-based focus in the design required minimization of the interface between the limb and the device. This increased the uncontrolled degrees of freedom within which the subject could move, which placed limitations on the subject groups for whom this system would be most effective. Currently, evaluation has been performed on normal-functioning adults capable of guiding the device along an ideal path. Subjects who lack movement control in the shoulder or elbow will not be well supported with the current system due to the uncontrolled degrees of freedom provided at those points. The same holds true for use of the hand; this function is essential for completion of the tasks, but active assistance is not provided in the current device design. The current design should be effective for higher-functioning impaired subjects, but to cater to more impaired groups, changes would need to be made to the overall design to further actively constrain the uncontrolled joints.

The use of a position-based control scheme minimizes constraints on a task by responding to the subject’s movement based on the task-specific path chosen. This approach complements the reduced degrees of active control, limited here to wrist and forearm, by emphasizing distal compensation that need not be tightly coupled to the uncontrolled degrees of freedom at the elbow and shoulder. By focusing assistance at the wrist and distal part of the limb, it may be possible to influence more proximal joints. The “error inducing” strategy introduced in Chapter 5 could play a key role in this process, by exposing the subject to greater orientation error in the wrist and forearm as they deviate from a “proper” path. This would force the subject to alter their movement path to achieve the elbow and shoulder orientations needed to produce a normative movement trajectory.
6.3 Device Role in Therapeutic Interventions

The analysis of therapy strategies described in Chapters 1 & 2 described several means by which CP and other movement disorders can be treated. Some subjects responded better to certain strategies, and the combination of multiple strategies typically resulted with the best functional outcome. The orthosis developed here is capable of leveraging (and enhancing) effective task-based physical therapies.

Since task-based therapy is frequently used with CP patients already, the device was designed to fit within this environment. Chapter 1 described the advantages to the integration of a robotic system with faster and more accurate feedback that can enable the therapist to fine tune clinical therapies. The proposed device accomplishes this through rapid response time (<20ms) and accurate assistance (<5° error). The system can provide the therapist with information regarding the subject’s positioning and orienting of the wrist, error in orientation, and the effectiveness of the system in response to commanded torques. With this information, the therapist will be able to pinpoint periods in the task where the subject struggles more, and supplement therapy with additional strategies to work on these deficient areas specifically. The therapist can also get a sense of the subject’s movement profile prior to therapy and judge whether the system is able to adequately support rehabilitation by initially operating the device in a passive mode. By visualizing the movement profile first, the therapist can choose the correct approach map, if multiple exist, or choose not to use the device if the subject’s impairment level prevents them from reasonably completing the task. The difficulties shown with assisting abnormal movement profiles in Chapter 5 shows that the use of this system may not be appropriate for all subjects, and it will be up to the therapist to make a decision whether or not this system should be used in each case.
The ability of the device to measure both force and position information, as demonstrated during the normative trajectory investigative study, enables the device to be used as an assessment tool outside of therapeutic use. There exist very few means by which a therapist can quantitatively track changes in movement profiles as a result of therapy participation. The orthosis provides a simple and direct means by which position and force data can be collected and analyzed quantitatively. Acquiring this data by other means, such as with motion capture systems, can require time-consuming setup and post-processing. Also, motion capture systems are unable to measure forces applied at the wrist. Other robotic therapy systems, which are capable of recording similar position and force information, can only help determine gains in functional outcome on targeted repetitive motion. Many of the systems described in Chapter 2 ([6], [7], [8], [9], [10]) focus on a single rotational movement of the forearm or wrist, and require a handle to be grasped, limiting their ability to measure movement in a task-based therapy environment. With a focus on task-based therapy, the orthosis developed here can provide quantitative insight on therapeutic progress within a variety of clinical tasks. These quantitative measures can also be used to help clarify the relative effectiveness of different therapy strategies.

Finally, the device can also be placed into an isometric mode where the actuated joints are locked, and user applied forces can be measured in all four actuated movement directions. This feature can be used to evaluate muscle strength and control, and provides the opportunity the relate force output at the forearm and wrist to the orientation profile within a task.

### 6.4 Future Directions

With a successful proof-of-concept, the next stage for the device will be to prepare it for introduction into the clinical environment and an impaired population. Several modifications should be made to the overall design (structural, electrical, and control) before moving to testing
within the clinical setting. Considerations should also be made to explore implementing elements of adaptive control or error inducing strategies to expand use across a broader population.

Most pressing of all changes to the system would be to replace the motors of the actuated joints with properly sized motors capable of providing the necessary torque bandwidth for active compensation. To meet the torque requirement of 5 Nm stated in section 3.4, the gearhead on the motor for the pronation/supination joint should be upsized to a ratio of at least 58:1 and at least 80:1 for the flexion/extension joint. Though little issue was seen in performance of the flexion/extension joint, the observed commanded torques did reach the upper limit of the current torque bandwidth. It would be ideal to maintain a buffer between the upper torque limit and the expected maximum torque demand, to ensure any scenario when greater torque is needed can be met. With the upsizing of the gearhead, it is likely that the closed loop delay due to the execution of a torque command will increase. Delays associated with the flexion/extension motor, geared down at a 53:1 ratio, produced delays up to 26 ms, so it would be expected that the new gearing of the pronation/supination motor would be similar. Delays at the flexion/extension joint would also be expected to increase, but remain low enough where an impact on the subject’s ability to complete a task would not be expected.

The primary structural change would be to replace the prismatic slide used in positioning with a low-friction alternative to ensure this joint does not negatively impact the user’s ability to perform the tasks. The cable drives used on the actuated joints were effective, but had the potential to slip if the tension was not maintained. A better approach would be to transition to a belt-driven system, which would be easily maintained at a constant tension and would ensure synchronization between the joint and motor. Future designs should also consider looking into alternate options for providing gravity compensation beside the passive weight used in the current design. The static nature of the weight limits its effectiveness to compensate for gravity as the
user extends out from the passive arm, or rapidly changes movement direction, as an increase in the gravitational effect on the limb cannot be countered dynamically.

A priority in future designs should include making the cabling between electronic systems more robust. As the first iteration in the design, many of the cables and wire connections were laid out to make the device functional without fully considering long-term robustness. A cable run should be incorporated to smoothly extend and retract the cable set as the prismatic joint moves, and an enclosure should be incorporated for the motor connections made at the orthosis. This would add an additional layer of safety, by providing an additional physical separation between the user and electronic components.

Finally, limitations identified in the current control scheme should be addressed. The most pressing need is the reduction or elimination of the filter delay responsible for much of the error seen during validation. Efforts should be made to implement a hardware filter or outright replace the analog sensors with digital equivalents to avoid filtering altogether. This will ensure real-time response of the system regardless of the present gradients within the orientation maps. The current trajectory mapping approach is limited to assisting around a single task-specific trajectory created from an average of sampled normative movement profiles. The incorporation of a smoothing kernel helps account for subject-specific variability around the primary trajectory, but still forces the user to conform to the mapped path. Currently, the system does not have the ability to incorporate alternate, yet valid, movement strategies. The system is also always in an active state, responding to error signals even when the calculated error is effectively zero. If the user is moving normally, the active assist is minimal, but it would be ideal to have a system that could phase assistance in and out based on need. Though this was not an issue during initial evaluation, the consistently active nature of the device could potentially hinder the progress of a patient by not allowing them to eventually complete the task fully by themselves.
To circumvent this, adaptive control could be implemented that incrementally adjusts the target movement to drive the user toward a more biomechanically normal movement profile. The autoregressive (artificial neural network) models developed during the project showed promise for providing this solution, and should be further explored. A key benefit of the autoregressive modeling approach lies in the use of an analytical solution to providing active assistance as opposed to an empirical one. With this approach, movement profiles could be parameterized, with the “ideal” movement profile acting as the therapeutic goal. A subject’s impaired profile could then be characterized within the model parameter space and a therapeutic target profile could be created by systematically adjusting the model parameters of the active assist to move the subject’s profile closer to the ideal case. This would allow therapy to be incremented in order to promote a progressive strategy customized to the subject’s deficiencies.

With the unique design and control of the orthosis focused only on the orientation of the wrist and forearm, there is a lack of understanding as to the influence that control of the distal limb will have on the remainder of the limb. It is possible that the constraint of the distal degrees of freedom from active compensation, along with task-based constraints, will naturally limit the orientation of the elbow and shoulder. Controlling just the distal degrees of freedom could provide a similar therapeutic impact to active control of the entire upper extremity and could be tested in future studies using the current system. If found to provide a similar benefit, this would help promote the use of robot-mediated therapy devices that are minimally coupled to the subject, allowing greater freedom of movement and flexibility of use.


APPENDIX A – FORWARD KINEMATICS DERIVATION

\[
\begin{align*}
A_1 &= \begin{bmatrix}
1 & 0 & -s_1 & 0 \\
0 & 1 & 0 & 0 \\
0 & 0 & 1 & 0 \\
0 & 0 & 0 & 1
\end{bmatrix} \\
A_2 &= \begin{bmatrix}
s_2 & 0 & 0 & a_2 \ c_2 \\
-1 & 0 & 0 & a_2 \ s_2 \\
0 & 0 & 1 & 0 \\
0 & 0 & 0 & 1
\end{bmatrix} \\
A_3 &= \begin{bmatrix}
0 & 0 & 0 & 0 \\
0 & 0 & 0 & 0 \\
0 & 0 & 0 & 1 \\
0 & 0 & 0 & 0
\end{bmatrix} \\
A_4 &= \begin{bmatrix}
s_4 & 0 & -s_4 & 0 \\
0 & 1 & 0 & 0 \\
0 & 0 & 0 & 0 \\
0 & 0 & 0 & 0
\end{bmatrix} \\
A_5 &= \begin{bmatrix}
0 & 0 & 0 & 0 \\
0 & 0 & 0 & 0 \\
0 & 0 & 0 & 0 \\
0 & 0 & 0 & 0
\end{bmatrix} \\
A_6 &= \begin{bmatrix}
s_6 & 0 & -s_6 & 0 \\
0 & 1 & 0 & 0 \\
0 & 0 & 0 & 0 \\
0 & 0 & 0 & 0
\end{bmatrix}
\end{align*}
\]
$$A_1 A_2 = \begin{bmatrix} C_1 & 0 & -S_1 & 0 \\ S_1 & 0 & C_1 & 0 \\ -S_2 & 0 & -S_2 + d_1 & -1 \\ 0 & 0 & 0 & 1 \end{bmatrix} \begin{bmatrix} S_2 & 0 & C_2 & a2C_2 \\ -C_2 & 0 & S_2 & a2S_2 \\ 0 & 0 & 1 & 0 \\ C_2 & 0 & -S_2 & -a2S_2 + d_1 \end{bmatrix} = \begin{bmatrix} C_1S_2 & S_1 & C_1C_2 & a2C_1C_2 \\ S_1S_2 & -C_1 & S_1C_2 & a2S_1C_2 \\ C_2 & 0 & -S_2 & -a2S_2 + d_1 \end{bmatrix}$$

$$A_1 A_2 A_3 = \begin{bmatrix} C_1S_2 & S_1 & C_1C_2 & a2C_1C_2 \\ S_1S_2 & -C_1 & S_1C_2 & a2S_1C_2 \\ C_2 & 0 & -S_2 & -a2S_2 + d_1 \end{bmatrix} \begin{bmatrix} 0 & 0 & -1 & 0 \\ 0 & 0 & 0 & 1 \\ 1 & 0 & d_3 & 0 \\ 0 & 0 & 0 & 1 \end{bmatrix} = \begin{bmatrix} -S_1 & C_1C_2 & -C_1S_2 & d3C_1C_2 + a2C_1C_2 \\ C_1 & S_1C_2 & -S_1S_2 & d3S_1C_2 + a2S_1C_2 \\ 0 & 1 & 0 & d_3 \end{bmatrix}$$

$$A_1 A_2 A_3 A_4 = \begin{bmatrix} -S_1 & C_1C_2 & -C_1S_2 & d3C_1C_2 + a2C_1C_2 \\ C_1 & S_1C_2 & -S_1S_2 & d3S_1C_2 + a2S_1C_2 \\ 0 & 0 & 0 & 1 \end{bmatrix} \begin{bmatrix} -C_1S_2 & -S_1C_4 + C_1C_2S_4 & S_1S_4 + C_1C_2C_4 & -d4C_1S_2 + d3C_1C_2 + a2C_1C_2 \\ -S_1S_2 & -C_1C_4 + S_1C_2S_4 & C_1S_4 + S_1C_2C_4 & -d4S_1S_2 + d3S_1C_2 + a2S_1C_2 \\ 0 & -S_2 & d3S_2 - a2S_2 + d_1 & -d4C_2 - d3S_2 - a2S_2 + d_1 \end{bmatrix}$$

$$A_1 A_2 A_3 A_4 A_5 = \begin{bmatrix} -C_1S_2 & -S_1C_4 + C_1C_2S_4 & S_1S_4 + C_1C_2C_4 & -d4C_1S_2 + d3C_1C_2 + a2C_1C_2 \\ -S_1S_2 & -C_1C_4 + S_1C_2S_4 & C_1S_4 + S_1C_2C_4 & -d4S_1S_2 + d3S_1C_2 + a2S_1C_2 \\ 0 & -S_2 & C_1C_2S_4 & S_1S_4 + C_1C_2C_4 \\ 0 & 0 & 0 & 1 \end{bmatrix}$$

$$= \begin{bmatrix} -C_1S_2C_5 - S_5(S_1C_4 - C_1C_2S_4) & S_1S_4 + C_1C_2S_4 & -C_1S_2C_5 + C_5(S_1C_4 - C_1C_2S_4) & -a5C_1S_2C_5 - a5S_5(S_1C_4 - C_1C_2S_4) - d4C_1S_2 + d3C_1C_2 + a2C_1C_2 \\ -S_1S_2C_5 + S_5(C_1C_4 + S_1C_2S_4) & -C_1S_2C_5 + C_5(S_1C_4 + S_1C_2S_4) & -a5S_1S_2C_5 + a5S_5(C_1C_4 + S_1C_2S_4) - d4S_1S_2 + d3S_1C_2 + a2S_1C_2 \\ 0 & S_2C_4 & C_1S_2S_5 + S_5S_4C_5 & -a5C_2C_5 - a5S_5S_4C_5 - d4C_2 - d3S_2 - a2S_2 + d_1 \end{bmatrix}$$
APPENDIX B - DERIVATION OF FORCE COMPENSATION EQUATIONS

\[ F_{top} = F_{top,FE} + F_{top,PS} \]
\[ F_{bottom} = F_{bottom,FE} + F_{bottom,PS} \]

\[ F_{top,FE} = 1.7F_{bottom,FE}; \quad F_{bottom,FE} = \frac{F_{top,FE}}{1.7}; \quad F_{top,PS} = -F_{bottom,PS}; \quad F_{bottom,PS} = -F_{top,PS} \]

Derivation 1:
\[ F_{top} = F_{top,FE} + F_{top,PS} \Rightarrow F_{top,FE} = F_{top} - F_{top,PS} \]
\[ F_{bottom} = F_{bottom,FE} + F_{bottom,PS} = \frac{F_{top,FE}}{1.7} - F_{top,PS} \Rightarrow F_{top,FE} = 1.7(F_{bottom} + F_{top,PS}) \]
\[ F_{top} - 1.7F_{bottom,PS} = 1.7F_{bottom} + 1.7F_{top,PS} \]
\[ F_{top} - 1.7F_{bottom,PS} = 2.7F_{top,PS} \]
\[ F_{top,PS} = \frac{F_{top} - F_{bottom}}{2.7} + \frac{F_{bottom}}{1.588} \]

Derivation 2:
\[ F_{top} = F_{top,FE} + F_{top,PS} \Rightarrow F_{top,FE} = F_{top} + F_{bottom,PS} \]
\[ F_{bottom} = F_{bottom,FE} + F_{bottom,PS} = \frac{F_{top,FE}}{1.7} + F_{bottom,PS} \Rightarrow F_{top,FE} = 1.7(F_{bottom} - F_{bottom,PS}) \]
\[ F_{top} + F_{bottom,PS} = 1.7F_{bottom} - 1.7F_{bottom,PS} \]
\[ F_{top} - 1.7F_{bottom,PS} = -2.7F_{bottom,PS} \]
\[ F_{bottom,PS} = -\frac{F_{top}}{2.7} + \frac{F_{bottom}}{1.588} \]

Derivation 3:
\[ F_{top} = F_{top,FE} + F_{top,PS} \Rightarrow F_{top,PS} = F_{top} - F_{top,FE} \]
\[ F_{bottom} = F_{bottom,FE} + F_{bottom,PS} = \frac{F_{top,FE}}{1.7} - F_{top,PS} \Rightarrow F_{top,PS} = \frac{F_{top,FE}}{1.7} - F_{bottom} \]
\[ F_{top} - F_{top,FE} = \frac{F_{top,FE}}{1.7} - F_{bottom} \]
\[ 1.7F_{top} - 1.7F_{top,FE} = F_{top,FE} - 1.7F_{bottom} \]
\[ 1.7F_{top} + 1.7F_{bottom,PS} = 2.7F_{top,FE} \]
\[ F_{top,FE} = -\frac{F_{top} + F_{bottom}}{1.588} \]

Derivation 4:
\[ F_{top} = F_{top,FE} + F_{top,PS} \Rightarrow F_{top,PS} = F_{top} - 1.7F_{bottom,FE} \]
\[ F_{bottom} = F_{bottom,FE} + F_{bottom,PS} \Rightarrow F_{top,PS} = F_{bottom,FE} - F_{bottom} \]
\[ F_{top} - 1.7F_{bottom,FE} = F_{bottom,FE} - F_{bottom} \]
\[ F_{top} + F_{bottom} = 2.7F_{bottom,FE} \]
\[ F_{bottom,FE} = \frac{F_{top} + F_{bottom}}{2.7} \]
APPENDIX C - AVERAGE TRAJECTORY PROFILES ACROSS SUBJECTS

**Figure 66:** Average trajectories across subjects for the pitcher task.

**Figure 67:** Average trajectories across subjects for the spoon task.
Figure 68: Average trajectories across subjects for the cone task.

Figure 69: Average trajectories across subjects for the peg task.
APPENDIX D - K-MEANS CLUSTERING RESULTS:

Figure 70: Cluster and silhouette plots for Cup-to-Mouth task. (Top Left) Cluster plot PC1-PC2 plane. (Top Right) Cluster plot PC1-PC3 plane. (Bottom Left) Cluster plot PC2-PC3 plane. (Bottom Right) Silhouette plot. * = Cluster 1, * = Cluster 2

Figure 71: Cluster and silhouette plots for Pouring-from-Pitcher task. (Top Left) Cluster plot PC1-PC2 plane. (Top Right) Cluster plot PC1-PC3 plane. (Bottom Left) Cluster plot PC2-PC3 plane. (Bottom Right) Silhouette plot. * = Cluster 1, * = Cluster 2
Figure 72: Cluster and silhouette plots for Spoon-to-Mouth task. (a) Cluster plot PC1-PC2 plane. (b) Cluster plot PC1-PC3 plane. (c) Cluster plot PC2-PC3 plane. (d) Silhouette plot. * = Cluster 1, * = Cluster 2

Figure 73: Cluster and silhouette plots for Peg task. (a) Cluster plot PC1-PC2 plane. (b) Cluster plot PC1-PC3 plane. (c) Cluster plot PC2-PC3 plane. (d) Silhouette plot. * = Cluster 1, * = Cluster 2
## APPENDIX E - DEVICE VALIDATION WITH SUBJECTS – RAW DATA

NOTE: Instances highlighted in red indicate iterations failing the acceptance criteria.

### Robot Driving Condition

#### Rho:

<table>
<thead>
<tr>
<th>Iteration</th>
<th>Cup to Mouth Mean</th>
<th>Pouring from Pitcher Mean</th>
<th>Eating with Spoon Mean</th>
<th>Stacking Cones 1 Mean</th>
<th>Stacking Cones 2 Mean</th>
<th>Stacking Cones 3 Mean</th>
<th>Placing Pegs Mean</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.745</td>
<td>2.149</td>
<td>0.281</td>
<td>0.752</td>
<td>0.338</td>
<td>0.866</td>
<td>0.160</td>
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<td></td>
<td></td>
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<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>2</td>
<td>1.003</td>
<td>3.32</td>
<td>3.667</td>
<td>0.395</td>
<td>0.991</td>
<td>0.235</td>
<td>5.901</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
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- **Cup to Mouth**
- **Pouring from Pitcher**
- **Eating with Spoon**
- **Stacking Cones 1**
- **Stacking Cones 2**
- **Stacking Cones 3**
- **Placing Pegs**
Co-Contraction Condition

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APPENDIX F – TESTING PROTOCOL

1. Test Setup

1.1. Prescreen

1.1.1. Subject fills out IRB-approved prescreen form as first eligibility check. See Appendix 1 for proper form.

1.1.2. Subject is asked to try on orthosis to ensure that their limb is appropriately sized for the evaluation. They should be able to slide their arm in without difficulty and be able to extend their hand within the hand brace with the fingers unimpeded.

1.1.3. If subject passes items 1 and 2, measurements are made from the 3rd metacarpal to the wrist on each hand to allow the orthosis to be adjusted prior to testing.

1.2. Data Acquisition Setup

1.2.1. Plug in the crossover cable between the host and target computers, and turn on the Target PC, allowing the connection between the two computers to be established.

1.2.2. Start MATLAB and open the appropriate test model.

1.2.2.1. For a passive run allowing joint measurement, but not providing active assistance, load data_acq_passiveV2.slx.

1.2.2.2. For an active run providing active assistance using the task maps, load file ortho_CAN_control.slx.

1.2.3. On the main command line type offset=[0 0]; This preallocates values for the force sensor calibration. The model will not build correctly without it.

1.2.4. Build the model. This initial build should at least include all five potentiometer scopes. This can be adjusted later within the Simulink model.

1.2.5. Turn on the control box and start EPOS Studio. Load the file 2ch_config. Connect to the motor controllers. STOP at this point until Physical Setup protocol is complete. See Section 1.1.3 for this procedure.

1.2.6. With the subject oriented in the neutral position, reset both nodes within EPOS Studio to zero the encoders and ensure proper orthosis resistance compensation.

1.2.7. Rebuild the model. Active assistance testing should include at least the two encoder output scopes. Passive testing should have at least the encoder scopes and two force sensor scopes.

1.3. Physical Setup

1.3.1. Adjust the hand link per the measurements taken during the prescreen process.

1.3.2. Potentiometer Calibration:

1.3.2.1. Place the orthosis within the calibration block and align along the joint 1 zeroing line, with the prismatic slide fully extended.

1.3.2.2. Start the model and record up to 10 seconds of data.

1.3.2.3. Record the mean voltage values for joints 1,2,4, and 5. Omit the first second of data due to the start up transient.

1.3.2.4. Rotate joint 1 so the calibration block aligns with the 25 degree line. Only joint 1 should change position.

1.3.2.5. Start the model and record up to 10 seconds of data.
1.3.2.6. Record the mean voltage values for joints 1, 2, 4, and 5. Omit the first second of data due to the startup transient.

1.3.2.7. Repeat steps 1.3.2.1-1.3.2.6. Collect at least 3 sets of data. If the observed values are not consistent with one another, collect additional sets of values until consistency is reached.

1.3.2.8. Return to the joint 1 zeroing position. Start the model and after a few seconds, pick up the orthosis and push the prismatic slide to the fully retracted position. After a few seconds, pull the slide back out to its fully extended position. Finish by pushing the slide back to the retracted position.

1.3.2.9. Look at the plot for joint 3 and identify the periods where fully extended and retracted. Generate average values for each position.

1.3.2.10. Take the average of the values for both joint 1 positions, joint 2, joint 4, and joint 5.

1.3.2.11. Write all values into the function volt2deg within the Simulink model.

1.3.2.12. Rebuild the model and return the orthosis to the original calibration position. Run the model and look at the forward kinematic outputs. Ideal results should be x=855, y=-67, z=52. Results within 10 mm of these are considered acceptable.

2. Testing Procedure

2.1. Test Preparation

2.1.1. Putting on the Orthosis:

2.1.1.1. Have the user put on the glove, tightening the wrist strap to give a snug fit.

2.1.1.2. Have the user carefully slide their arm into the orthosis until their wrist aligns with the flexion/extension joint of the orthosis. There are nuts and bolts that slightly protrude along this path, which can become snagged on the glove, causing it to tear.

2.1.1.3. Tighten the hand brace Velcro strap first, creating a snug fit around the hand.

2.1.1.4. Tighten the medial side glove strap around the thin extension of the hand brace.

2.1.1.5. Tighten the lateral side glove strap around the large body of the hand brace. At this point the hand should not be able to easily slip longitudinally out of the hand brace.

2.1.1.6. Tighten the two straps attached around the trough to secure the forearm in place.

2.1.2. Training:

2.1.2.1. Explain the Home position to the subject and its importance. Be sure to describe how the wrist should be aligned within the Home position.

2.1.2.2. Set up the cup task and describe how it is to be performed. Allow the subject to practice the task a couple of times to get used to the feeling of the orthosis on their wrist.
2.1.2.3. Do the same with the other four tasks. *For descriptions of the tasks, refer to the Testing protocol, step #4.*

2.2. Running the Test

2.2.1. Have the subject sit up straight in the chair with feet flat on the floor. They should be centered in front of the task board and easel.

2.2.2. Have them align their wrist with the Home position, leaving the arm and hand in a neutral position. The neutral position has the hand oriented vertically, with thumb pointing up, and directed straight ahead.

2.2.3. *(FOR PASSIVE TESTING ONLY)* With the motors disabled, run the model to get an initial force sensor reading. Make sure the subject is as relaxed as possible while doing this. Collect up to 10 seconds of data, and run the script *bias.m.* A box will be displayed confirming the offset be adjusted. Click yes. Run the model again to confirm that the baseline force sensor reading is nearly zero. If needed, the *bias.m* script can be run again until a desirable result is seen.

2.2.4. Tasks: These are performed over 3 trials with 10 iterations per trial. Each iteration starts and end at the Home position.

2.2.4.1. **Cup to Mouth:** This task uses the large cup, placed in the appropriate position on the task board. The subject will need to move their hand to the cup, grasp the cup, and bring the cup to their mouth. Once the rim of the cup is in close proximity to the mouth, it can be returned to its place on the task board, grasp can be released, and the subject can return their wrist to the Home position in the neutral orientation.

2.2.4.2. **Pouring from a Pitcher:** This task uses the small cup and measuring cup with handle acting as the pitcher. Place both on the task board in their corresponding positions. The subject will move to the pitcher, grasping the handle, and pronating their forearm to imitate pouring from the pitcher into the small cup. The pitcher oriented in the pouring position should be held briefly, and then returned to its original place on the task board. The subject can release grasp on the pitcher, and then return their wrist to the Home position in a neutral orientation.

2.2.4.3. **Spoon to Mouth:** This task includes the spoon and bowl. Place the bowl on the appropriate space on the task board. The spoon is held by the subject prior to testing. It is important to conduct step #3 of the Testing protocol while they are holding the spoon. This task requires the subject to move from the Home position toward the bowl, mimicking a scooping motion, and bringing the spoon to the mouth. Once the end of the spoon is in close proximity to the mouth, the user can return their wrist to the Home position in the neutral orientation.

2.2.4.4. **Stacking Cones:** This task uses a stack of four plastic cones. There is also a cone glued to a task board base piece. This should be placed in the appropriate location while the rest of the cones are placed in the upper right of the task board within the yellow square. The subject moves to the cones in the upper right of the task board, grasping the top cone and moving it on top of the glued cone in the bottom left of the task board.
This process is repeated until all cones have been transported from the upper right to the bottom left. Once all cones are moved, the subject returns to the Home position with the wrist oriented in the neutral position. This task also requires the operator to move the stacked cones back to the starting position once an iteration has been performed.

2.2.4.5. **Placing Pegs:** This task uses the foam board placed on the easel and five plastic pegs. Each peg is placed within an opening on the task board to start. The subject moves and grasps the first peg, moving and placing the peg within a predetermined hole on the foam board. The subject then moves to the second peg, grasping and placing it within a predetermined hole on the foam board. This is repeated until all pegs have been placed on the foam board. Once this is accomplished, the subject returns to the Home position with the wrist oriented in the neutral position. This task also requires the operator to remove the pegs from the foam board and place them back on the task board once an iteration has been completed.

2.2.4.6. **ADDITIONAL NOTE:** The model is started prior to the first iteration and continues to run until the ten iterations for the trial have been completed. When the subject completes an iteration and returns to the Home position, it’s important to make sure the wrist joints are zeroed prior to starting the next iteration. This can be done with the scopes displaying the joint position information of the wrist. If the traces are off from zero, the subject should be asked to adjust their wrist until a desirable trace is seen.

2.2.5. Observe the approach used by the subject to complete the task. Make qualitative notes regarding hand position, region of object grasped, and any other movement that may be relevant.

2.2.6. Once a trial has been completed, the data should be saved to a .mat file unique to the subject, task, and trial run. Information to be included in the file should include the x, y, and z Cartesian position for the wrist, encoder outputs for flexion/extension and pronation/supination, and the raw output from the two force sensors. This time is also allotted as a rest period for the subject, and to set up for the next trial or task.
APPENDIX G – CONTROL SYSTEM WALKTHROUGH AND DESCRIPTION

1. High Level Control

![Figure 74: Top Level View of Active Control]

1.1. Data Acquisition Block (Blue Box):
This block handles incoming position and orientation signals coming from the passive arm and the orthosis. This includes 3 analog potentiometers on the arm, 2 analog potentiometers on the orthosis, and 2 digital encoders mounted to the two motor assemblies. Each signal is filtered and processed, outputting an (x,y,z) Cartesian position and an angular (rad) orientation for the wrist.

1.2. Map Block (Red Box):
This is the most important subsystem, as it takes the current Cartesian position of the wrist and generates an ideal orientation based on a direct mapping between position and orientation, created based on data collected off normal functioning adults while performing tasks with the device. This block needs to be adjusted every time a new task is used.

1.3. Orthosis Model Blocks (Green Box):
The Orthosis Model blocks are made up of four blocks that mathematically model dynamic characteristics of the orthosis, including mass, Coriolis forces, gravitational forces, and a PID controller that generates an error signal based on differences between the projected orientation and actual orientation. The basis for this modeling comes from the Euler-Lagrangian equations of motion that, when derived, represent these discrete forces.
1.4. CAN Current Send (Purple Box):
This block is the final transmission of the commanded current to the motor controllers that execute the command at the motors. This current is created using the compensating torque, calculated from the dynamic model and error compensation, and multiplying it by a gain value. The CAN protocol was selected based on its ability to transmit data quickly, as the device is required to respond to the user in real time.

2. Low Level Control – Data Acquisition Subsystem

**Figure 75: Top Level View of Data Acquisition Subsystem**

2.1. DAQ Signal In:
This lumps both the analog and digital inputs from the DAQ card, and the channel selection to ensure that each signal is differentiated correctly. Matlab provides an entire library of blocks for common DAQ cards, so it is important to make sure the right blocks are used. The system currently uses the NI PCI-6251 for data collection.
Figure 76: Parameters for Inputting Analog Signal to DAQ from Potentiometers.

The analog input block will provide a dialog box like the one shown in Figure 76 that allows the appropriate channels, voltage range, and type of signal (differential, referenced to ground) to be set. In this case, voltage range was set to max at 10V as each potentiometer is powered by a 12V source, and potentiometers were all referenced to a common ground separate from the internal ground provided in the DAQ card. The last two channels correspond to analog signals from the button load cells used for passive evaluation, and were set as a differential signal. Separate blocks were given for each digital input, where input channel and sample time were specified.

2.2. Low Pass Filter:
This block handles filtering of the analog signals by applying a minimum-order FIR filter with a cutoff frequency of 2 Hz. Normal limb movements should not exceed 2 Hz, so applying the cutoff this low is reasonable in order to attenuate as much noise as possible. Stopband frequency was set at 5 Hz, with an attenuation of -60 dB. Though this may seem excessive, it was experimentally found that these settings provided the best signal to noise ratio without degrading the signal of interest.

2.3. Volt to Degree Conversion:
This block houses a function script that performs a mathematical unit conversion from the raw signal input to a usable angular measurement. This relationship was found experimentally, and was integrated as part of the calibration of sensors prior to testing. During calibration, the robot is placed in an orientation that has been predetermined as the “zero” position for each potentiometer, and the corresponding voltages are noted in the script as the relative zero for each joint. The relationship between voltage and angle is created by the voltage difference observed along two lines creating a 25° angle. A volts/degree factor is created and is used as the divisor for the difference between the
read voltage and relative zero for each joint. The calculation for the elevation joint requires an extra step, as “zero” position places this joint as a slight angle. This angle was calculated mathematically as 5.6932°, and the relative zero for this joint is set relative to this. Performing the conversion for the linear string potentiometer used a simpler process, as the fully retracted position is the relative zero, while a volts/distance factor is determined based on the voltage difference over the known distance between the fully extended and fully retracted states.

2.4. Forward Kinematics:
This block provides the conversion that takes the angular position of each potentiometer and mathematically calculates the Cartesian position using the Denavit-Hartenberg convention. This convention represents each joint using a set of four basic transformations that feature a pair of rotations and a pair of translations in order to move to the next joint. Each homogenous transformation is multiplied with the other transformations in order to express the end effector position. Refer to section 3.6.3 to see the generation of expressions used for the orthosis.

2.5. Filtered Derivatives / Velocity Out:
The filtered derivatives take the angular orientation determined by the motor encoders for the actuated joints of the orthosis, and extract the instantaneous velocity for these joints. This information is utilized within the mapping subsystem. A discrete IIR filter is applied first, configured using the \texttt{fdatool} command in Matlab, with a cutoff frequency of 10 Hz. The discrete derivative blocks apply the derivative following the filter to create the final velocity output.

2.6. Real Time Displays:
This subsystem simply acts to display the real-time position and orientation values while running the model. This is most useful during the calibration steps to make sure the displayed position is in line with the position of the orthosis. These displays can also be useful in diagnosing faulty sensor readings, or monitoring whether or not proper compensation is being applied at the actuated joints.
3. Low Level Control – Map Subsystem

3.1. Inputs and Outputs:
The mapping function takes in both Cartesian position and velocity to generate the ideal orientation of the wrist for that position. As mentioned previously, orientation velocity is also made available, and can be used in cases where this velocity is better suited to define transitions in the map (this will be better explained later). A switch function delays the x-value for the first second the signal comes in to avoid the startup transient from the analog sensors. If left unaccounted for, this voltage spike could be read and transmitted as a rapid change in position, causing the system to falsely compensate.

3.2. Mapping Function:
The mapping function contains all the instructions necessary to project a wrist orientation from the read Cartesian position. Since each map has a resolution of 1 cubic centimeter, positions occurring between these units are not explicitly written in the maps. Instead, trilinear interpolation is used to interpolate the orientation based on the nearest mapped points within the 3-dimensional volume. This interpolated value is used as the ideal orientation within the control loop.

Since the programmed tasks are cyclic in nature, it is possible that the user will cross a position more than once. To handle this, each task trajectory was split into multiple
phases so that a position was crossed only once in the map. The remainder of the function handles the transitions between the maps for each task.

To create a smooth transition between maps, transitions are placed at instances where movement suddenly changes direction. Position thresholds were set to narrow the area where the transition takes place, and velocity thresholds were set near zero. When position was not enough to constrain the transition area, velocity was used to mark the direction change, which appeared as zero.

3.2.1. Code:

Each task has its own set of code that handles applying the maps and transitioning between maps. Each map is split into components of Rho (pronation/supination) and Theta (flexion/extension) per map and loaded into the Matlab workspace where they can be accessed by the function. Another block, denoted “segin”, is an array of numbers that identify the order that the maps should be accessed per cycle of the task. There are two recursive variables responsible for indexing the map transitions and noting the run time at which the transitions take place.

Each block of code is made up of a set of nested if-else statements that handle each map case. Maps are denoted by a number, so the first map used during the task is labeled “1”, the second map labeled “2”, and so on. The nested if-else statements work to determine if the user has entered the transition zone, and maps need to be changed based on position and velocity. Rotational velocity of the wrist and forearm is inputted to the function, in addition to translational velocity within the workspace. This has proven to be more effective in tasks, like the cone task, where changes in movement occur more frequently. This can be changed by connecting the “rho_dot” clock to the velocity input and using the selector to choose either the forearm or wrist. The code will continue to loop without any change until the position and velocity criteria are met, at which point the map selection index will increment, and a time window of 0.4 seconds will be created based on the current run time. At this point, the final two lines of code are implemented, which consists of sigmoid functions acting to smooth the transition between maps. This prevents movement during the transition from feeling sudden and jerky. These functions ramp from the first map to the second map over that 0.4 sec window so that the transition isn’t felt at all.
4. Low Level Control – Mathematical System Models

4.1. Purpose:
The purpose of these four blocks is to maintain control of the torque commands to the motors through a PID control loop and accurate modeling of the equations of motion that govern the robot and the user’s limb. These can be combined to form a modified version of the PD control law,

\[ \tau = M(\theta)\ddot{\theta} + C(\theta, \dot{\theta})\dot{\theta} + N(\theta, \dot{\theta}) - \left( K_p e(t) + K_i \int_0^t e(t) dt + K_d \dot{e}(t) \right). \]

The first term, referred to as the “mass matrix” is derived based on Lagrangian expressions for kinetic energy. Final expression of this term can be found using the equation,

\[ M(\theta) = \sum_{i=1}^{n} m_i J_{v_i}(\theta)^T J_{v_i}(\theta) + J_{w_i}(\theta)^T R_i(\theta) I_i R_i(\theta)^T J_{w_i}(\theta) \]

Where \( J_{v_i} \) is the Jacobian for linear velocity, \( J_{w_i} \) is the Jacobian for angular velocity, \( R_i \) is the rotation matrix between degrees of freedom, and \( I_i \) is the inertia matrix for the rotating body. Values for moments of inertia and mass were taken from the Solidworks model for the robot. Matrix \( C(\theta, \dot{\theta}) \) is derived from the Euler-Lagrange equations of motion formed into a set of second order ordinary differential equations. This term can be defined as

\[ C_{k,j} = \frac{1}{2} \sum_{i=1}^{n} \left\{ \frac{\partial M_{k,j}}{\partial \theta_i} + \frac{\partial M_{k,j}}{\partial \dot{\theta}_j} - \frac{\partial M_{k,j}}{\partial \theta_k} \right\} \]

where \( k \) and \( j \) are elements of the matrix \( C \), and \( i \) refers to each degree of freedom of the system. Matrix \( N \), also known as the “gravity term”, is calculated as the partial derivative of the potential energy for the system and can...
be written as \( N_i = \sum_{l=1}^{n} \frac{\partial m_i g^T r_c}{\partial \theta_i} \) where \( m \) is the mass of the link, \( g \) is the gravity vector, and \( r_c \) is the coordinates for the center of mass of link \( i \). Together, these three terms can model the dynamic movement of the system. The final three terms form the PID controller responsible for compensating the robot as needed to stay on the projected path.

4.2. Mass:

4.3. Centrifugal/Coriolis:
Centrifugal and Coriolis forces pertain to those forces that act on the body while moving over a rotational path. Calculating matrix \( C(\theta, \theta_{dot}) \) requires inertial inputs from the forearm and limb, including mass and location of the center of mass.

4.4. Gravity:
Gravitational forces are taken into consideration based on the distance that the robot sits above the tabletop. Inertial inputs from the forearm and wrist are again included in the final calculation.

4.5. Controller:
The PID controller set up to minimize the errors between input signal and set point was tuned using a computational model of the robot dynamics.

\[
N_i = \sum_{l=1}^{n} \frac{\partial m_i g^T r_c}{\partial \theta_i}
\]

*Figure 79: Model Flow Mass Effect on System Dynamics.*
Figure 80: Computational Model of System Dynamics for Controller Tuning.

The computational model, shown in Figure 80, uses the same mathematical models used to represent the robot dynamics, but includes an additional block that generates the inverse dynamics used to translate from torque back to position, velocity, and acceleration (Figure 81).

Figure 81: View of Inverse Dynamics Subsystem for Computational Model in Figure 80.

With this arrangement, the “autotune” feature can be used within the PID blocks, which will linearize the model and allow the controller gains to be tuned to provide adequate
response time and accuracy. Once found, gains were tested experimentally on the actuated system, and evaluated based on performance. If the response felt too jerky or slow, the gains were adjusted and tested until responsive and smooth result was found.

5. Low Level Control – CAN Current Send

![Diagram of Low Level Control](image)

**Figure 82:** Pack and Send of Current Commands. (Top) Final stage of control where torques are converted to current values and sent. (Bottom) View of CAN Current Send subsystem.

5.1. Purpose:
The purpose for these blocks was to establish a communication line between the motor controllers and the host computer where current commands could be sent and executed by the motors. This was accomplished by using the CAN communication protocol, which was chosen for its ability to handle large packets of data very quickly. Commands were sent to preallocated “addresses” that were responsible for handling a certain command within the controller. These addresses could be found within the EPOS Studio software interface, and adjusted to send or receive commands.

5.2. Preparing the command:
Commands were first converted from a torque to a current value by applying a gain term based on the torque constant of the motors and the mechanical advantage provided by the system. This then passed through a saturation block, setting the upper and lower limits for the commanded current. This was done to ensure an adequate buffer was established, and the commanded current never exceeded the maximum acceptable current to the motors.
Finally, the command was converted to 16-bit format, which was needed to package and send it to the motor controllers.

5.3. CAN Pack/Send:

Here, current commands are transmitted to the motor controllers. The address where the command is to be sent is provided as the “identifier”, converted from decimal to hexadecimal format. Commands were required to be sent as 64-bit numbers, and since most commands were only 16 or 32-bit in length, the rest of the bits were filled with zeros as placeholders. Once packed, the command was sent through the Send block. For a successful send to take place, it was important to make sure that the identifier listed matched the identifier listed in the Pack block.

Figure 83: Parameters for transmitting current commands using CAN protocol.
1. Function “volt2deg” – Converted raw potentiometer voltages to angular joint measurements in degrees.

```matlab
function [y1,y2,y3,y4,y5,y6,y7] = volt2deg(v1,v2,v3,v4,v5,e6,e7)
%#codegen
%Enter Calibration Voltages
joint1_zero=3.715;
joint1_25=4.6682;
joint2=7.068;
joint3_extended=9.4258;
joint3_retracted=1.9599;
joint4=6.6983;
joint5=7.289;

slope1245=(joint1_25 - joint1_zero)/25;
slope3=(joint3_extended - joint3_retracted)/438;

joint1_os=joint1_zero;
joint2_os=joint2 + (slope1245 * 5.6932);
joint4_os=joint4;
joint5_os=joint5;

y1=(joint1_os - v1)./slope1245;
y2=(joint2_os - v2)./slope1245;
y3=(v3 - 1.9599)./slope3;
y4=(v4 - joint4_os)./slope1245;
y5=(v5 - joint5_os)./slope1245;
y6=(pi/180)*(30/18750).*e6;
y7=(pi/180)*(30/33920).*e7;
```

2. Function for Forward Kinematics – Processed angular measurements at each joint to produce x,y,z position of the wrist.

```matlab
function [x,y,z] = fcn(j1,j2,d3,j4,j5,j6)
%#codegen
d1=336.55;
da2=434.46;
da4=45.86;
da5=120.51;
da6=67.08;
da6=39.52;
```
s1 = sind(j1);
c1 = cosd(j1);
s2 = sind(j2);
c2 = cosd(j2);
s4 = sind(j4);
c4 = cosd(j4);
s5 = sind(j5);
c5 = cosd(j5);
s6 = sind(j6);
c6 = cosd(j6);

x = -a6*c6*((c1*s2*c5)+s5*((s1*c4)-(c1*c2*s4)))+a6*s6*((s1*s4)+(c1*c2*c4))-
    d6*((c1*s2*s5)-c5*((s1*c4)-(c1*c2*s4)))-(a5*c1*s2*c5)-a5*s5*((s1*c4)-(c1*c2*s4))-
    (d4*c1*s2)+((a2+d3)*c1*c2);
y = -a6*c6*((s1*s2*c5)-s5*((c1*c4)+(s1*c2*s4)))-a6*s6*((c1*s4)-(s1*c2*c4))-
    d6*((s1*s2*s5)+c5*((c1*c4)+(s1*c2*s4)))-(a5*s1*s2*c5)+a5*s5*((c1*c4)+(s1*c2*s4))-
    (d4*s1*s2)+((a2+d3)*s1*c2);
z = -a6*c6*((c2*c5)+(s2*s4*s5))-(a6*s2*c4*s6)-d6*((c2*s5)-(s2*s4*c5))-(a5*c2*c5)-(a5*s2*s4*s5)-(d4*c2)-
    ((a2+d3)*s2)+d1;

3. Mapping Function – This is responsible for applying the assigned map, generating
   orientations based on position, and making transitions between maps.

function [rho, theta, iout, to] = fcn(pos, vel, ~, ~, r1, t1, r2, t2, r3, t3, r4, t4, segin, i, ti, time)
 %codegen

p = pos./10;
v = vel;

%TRILINEAR INTERPOLATION
%set upper and lower map points
up = ceil(p);
low = floor(p);
%determine distance from current point to lower map point
d = p - low;

rho001 = (r1(low(1), low(2), low(3))*(1-d(1)))+r1(up(1), low(2), low(3))*d(1);
theta001 = (t1(low(1), low(2), low(3))*(1-d(1)))+(t1(up(1), low(2), low(3))*d(1));
\[ \rho_{101} = (r_1(\text{low}(1), \text{up}(2), \text{low}(3)) \times (1-d(1))) + (r_1(\text{up}(1), \text{up}(2), \text{low}(3)) \times d(1)); \]
\[ \theta_{101} = (t_1(\text{low}(1), \text{up}(2), \text{low}(3)) \times (1-d(1))) + (t_1(\text{up}(1), \text{up}(2), \text{low}(3)) \times d(1)); \]
\[ \rho_{011} = (r_1(\text{low}(1), \text{low}(2), \text{up}(3)) \times (1-d(1))) + (r_1(\text{up}(1), \text{low}(2), \text{up}(3)) \times d(1)); \]
\[ \theta_{011} = (t_1(\text{low}(1), \text{low}(2), \text{up}(3)) \times (1-d(1))) + (t_1(\text{up}(1), \text{low}(2), \text{up}(3)) \times d(1)); \]
\[ \rho_{111} = (r_1(\text{low}(1), \text{up}(2), \text{up}(3)) \times (1-d(1))) + (r_1(\text{up}(1), \text{up}(2), \text{up}(3)) \times d(1)); \]
\[ \theta_{111} = (t_1(\text{low}(1), \text{up}(2), \text{up}(3)) \times (1-d(1))) + (t_1(\text{up}(1), \text{up}(2), \text{up}(3)) \times d(1)); \]
\[ \rho_{1} = \rho_{001} \times (1-d(2)) + (\rho_{101} \times d(2)); \]
\[ \theta_{1} = \theta_{001} \times (1-d(2)) + (\theta_{101} \times d(2)); \]
\[ \rho_{11} = \rho_{011} \times (1-d(2)) + (\rho_{111} \times d(2)); \]
\[ \theta_{11} = \theta_{011} \times (1-d(2)) + (\theta_{111} \times d(2)); \]
\[ \rho_{02} = \rho_{002} \times (1-d(2)) + (\rho_{102} \times d(2)); \]
\[ \theta_{02} = \theta_{002} \times (1-d(2)) + (\theta_{102} \times d(2)); \]
\[ \rho_{12} = \rho_{012} \times (1-d(2)) + (\rho_{112} \times d(2)); \]
\[ \theta_{12} = \theta_{012} \times (1-d(2)) + (\theta_{112} \times d(2)); \]
\[ \rho_{2} = \rho_{02} \times (1-d(3)) + (\rho_{12} \times d(3)); \]
\[ \theta_{2} = \theta_{02} \times (1-d(3)) + (\theta_{12} \times d(3)); \]
\[ \rho_{003} = (r_3(\text{low}(1), \text{low}(2), \text{low}(3)) \times (1-d(1))) + (r_3(\text{up}(1), \text{low}(2), \text{low}(3)) \times d(1)); \]
\[ \theta_{003} = (t_3(\text{low}(1), \text{low}(2), \text{low}(3)) \times (1-d(1))) + (t_3(\text{up}(1), \text{low}(2), \text{low}(3)) \times d(1)); \]
\[
\begin{align*}
\rho_{103} &= (r_3(\text{low}(1), \text{up}(2), \text{low}(3))*(1-d(1))) + (r_3(\text{up}(1), \text{up}(2), \text{low}(3))*d(1)) \\
\theta_{103} &= (t_3(\text{low}(1), \text{up}(2), \text{low}(3))*(1-d(1))) + (t_3(\text{up}(1), \text{up}(2), \text{low}(3))*d(1)) \\
\rho_{013} &= (r_3(\text{low}(1), \text{low}(2), \text{up}(3))*(1-d(1))) + (r_3(\text{up}(1), \text{low}(2), \text{up}(3))*d(1)) \\
\theta_{013} &= (t_3(\text{low}(1), \text{low}(2), \text{up}(3))*(1-d(1))) + (t_3(\text{up}(1), \text{low}(2), \text{up}(3))*d(1)) \\
\rho_{113} &= (r_3(\text{low}(1), \text{up}(2), \text{up}(3))*(1-d(1))) + (r_3(\text{up}(1), \text{up}(2), \text{up}(3))*d(1)) \\
\theta_{113} &= (t_3(\text{low}(1), \text{up}(2), \text{up}(3))*(1-d(1))) + (t_3(\text{up}(1), \text{up}(2), \text{up}(3))*d(1)) \\
\rho_{03} &= (\rho_{003}*(1-d(2)) + (\rho_{103}*d(2)) \\
\theta_{03} &= (\theta_{003}*(1-d(2)) + (\theta_{103}*d(2)) \\
\rho_{13} &= (\rho_{013}*(1-d(2)) + (\rho_{113}*d(2)) \\
\theta_{13} &= (\theta_{013}*(1-d(2)) + (\theta_{113}*d(2)) \\
\rho_{3} &= (\rho_{03}*(1-d(3)) + (\rho_{13}*d(3)) \\
\theta_{3} &= (\theta_{03}*(1-d(3)) + (\theta_{13}*d(3)) \\
\end{align*}
\]

%%% Map Transitions for Each Task
% Code written and separated by task for moving between maps
% within a task. Only the task that is currently being used is
% uncommented during this portion of the function.

%%% CUP TASK
%  if segin(i) == 1
%      if p(2) >= 35 && p(3) >= 15
%          if v > -1e-2 && abs(v(2)) <= 5
%              iout = i+1;
%              to = time+.4;
%          else
%              iout = i;
%              to = ti;
%          end
%      else
%          iout = i;
%          to = ti;
%      end
%  else
%      iout = i;
%      to = ti;
%  end
%  rho = (1./(1+exp((time-ti)/.058)))*rho1;
%  theta = (1./(1+exp((time-ti)/.058)))*theta1;

elseif segin(i) == 2
  if p(1) <= 34 && p(2) <= 27.5
    iout = 1;
    to = time+.4;
}
else
    iout=i;
    to=ti;
end
rho=((1-(1./(1+exp(-(time-ti)./0.058))))*rho1+((1./(1+exp(-(time-ti)./0.058)))*rho2);
theta=((1-(1./(1+exp(-(time-ti)./0.058))))*theta1+((1./(1+exp(-(time-ti)./0.058)))*theta2);

else
    rho=((1-(1./(1+exp(-(time-ti)./0.058))))*rho2);
    theta=((1-(1./(1+exp(-(time-ti)./0.058))))*theta2);
    theta=0;
    if p(1) >= 34 && p(2) <= 22
        iout=i+1;
        to=time+.4;
    else
        iout=i;
        to=ti;
    end
end

elseif segin(i) == 2
    if p(1) <= 34 && p(2) <= 27.5
        iout=1;
        to=time+.4;
    else
        iout=i;
        to=ti;
    end
else
    iout=i;
    to=ti;
end
rho=(1./(1+exp(-(time-ti)./0.058)))*rho1;
theta=(1./(1+exp(-(time-ti)./0.058)))*theta1;
elseif segin(i) == 2
    if p(1) <= 34 && p(2) <= 27.5
        iout=1;
        to=time+.4;
    else
        iout=i;
        to=ti;
    end
rho=(1./(1+exp(-(time-ti)./0.058)))*rho1;
theta=(1./(1+exp(-(time-ti)./0.058)))*theta1;
elseif segin(i) == 2
    if p(1) <= 34 && p(2) <= 27.5
        iout=1;
        to=time+.4;
    else
        iout=i;
        to=ti;
    end
rho=(1./(1+exp(-(time-ti)./0.058)))*rho1;
theta=(1./(1+exp(-(time-ti)./0.058)))*theta1;
else
  iout=i;
  to=ti;
end
rho=((1-(1./(1+exp(-(time-ti)./0.058)))))*rho1+((1./(1+exp(-(time-ti)./0.058)))*rho2);
theta=((1-(1./(1+exp(-(time-ti)./0.058)))))*theta1+((1./(1+exp(-(time-ti)./0.058)))*theta2);
else
  rho=((1-(1./(1+exp(-(time-ti)./0.058)))))*rho2;
  theta=((1-(1./(1+exp(-(time-ti)./0.058)))))*theta2);
  if p(1) >= 34 && p(2) <= 22
    iout=i+1;
    to=time+.4;
  else
    iout=i;
    to=ti;
  end
end

elseif segin(i) == 1
  if p(2) >= 35 && p(3) >= 18
    if v > -1e-2 && abs(v(2)) <= 5
      iout=i+1;
      to=time+.4;
    else
      iout=i;
      to=ti;
    end
  else
    iout=i;
    to=ti;
  end
else
  iout=i;
  to=ti;
end
rho=(1./(1+exp(-(time-ti)./0.058)))*rho1;
theta=(1./(1+exp(-(time-ti)./0.058)))*theta1;
elseif segin(i) == 2
  if p(1) <= 34 && p(2) <= 27.5
    iout=1;
    to=time+.4;
  else
    iout=i;
    to=ti;
  end
rho=(1./(1+exp(-(time-ti)./0.058)))*rho1;
theta=(1./(1+exp(-(time-ti)./0.058)))*theta1;
elseif segin(i) == 2
  if p(1) <= 34 && p(2) <= 27.5
    iout=1;
    to=time+.4;
  else
    iout=i;
    to=ti;
  end
rho=(1./(1+exp(-(time-ti)./0.058)))*rho1;
theta=(1./(1+exp(-(time-ti)./0.058)))*theta1;
rho=((1-(1./(1+exp(-((time-ti)./0.058)))))*rho1)+(1./(1+exp(-((time-ti)./0.058)))))*rho2;%
theta=((1-(1./(1+exp(-((time-ti)./0.058)))))*theta1)+(1./(1+exp(-((time-ti)./0.058)))))*theta2);

else
    rho=((1-(1./(1+exp(-((time-ti)./0.058)))))*rho2);
    theta=((1-(1./(1+exp(-((time-ti)./0.058)))))*theta2);
    if p(1) >= 34 && p(2) <= 22
        iout=i+1;
        to=time+.4;
    else
        iout=i;
        to=ti;
    end
end

%% CONE TASK

if segin(i) == 1
    if p(1) >= 50
        if abs(v(1)) < 20
            iout=i+1;
            to=time+.4;
        else
            iout=i;
            to=ti;
        end
    else
        iout=i;
        to=ti;
    end
endif segin(i-1) == 9
    rho=(1./(1+exp(-((time-ti)./0.058))))*rho1;
    theta=(1./(1+exp(-((time-ti)./0.058))))*theta1;
else
    rho=((1-(1./(1+exp(-((time-ti)./0.058)))))*rho2)+(1./(1+exp(-((time-ti)./0.058)))))*rho1;
    theta=((1-(1./(1+exp(-((time-ti)./0.058)))))*theta2)+(1./(1+exp(-((time-ti)./0.058)))))*theta1;
endif segin(i) == 2
    if p(2) <= 17.5
if abs(v(1)) < 20
    iout=i+1;
    to=time+.4;
else
    iout=i;
    to=ti;
end
else
    iout=i;
    to=ti;
end
rho=((1-(1./(1+exp(-(time-ti)./0.058))))*rho1)+((1./(1+exp(-(time-ti)./0.058)))*rho2);
theta=((1-(1./(1+exp(-(time-ti)./0.058))))*theta1)+((1./(1+exp(-(time-ti)./0.058)))*theta2);
elseif segin(i) == 3
    if p(1) <= 34 && p(2) <= 27.5
        iout=1;
        to=time+.4;
    else
        iout=i;
        to=ti;
    end
    rho=((1-(1./(1+exp(-(time-ti)./0.058))))*rhol)+((1./(1+exp(-(time-ti)./0.058)))*rho2);
    theta=((1-(1./(1+exp(-(time-ti)./0.058))))*thetal)+((1./(1+exp(-(time-ti)/.058)))*theta2);
else
    rho=((1-(1./(1+exp(-(time-ti)./0.058))))*rho1)+((1./(1+exp(-(time-ti)./0.058)))*rho2);
    theta=((1-(1./(1+exp(-(time-ti)./0.058))))*theta1)+((1./(1+exp(-(time-ti)./0.058)))*theta2);
end
%        rho=0;
%        theta=0;
end
%        PEG TASK

% if segin(i) == 1
%    if p(1) >= 47.5 && p(2) <= 10 && p(3) >= 10
%        if v < 1e-2
% iout=i+1;
% to=time+.4;
else
% iout=i;
% to=ti;
end
else
% iout=i;
% to=ti;
end
rho=(1./(1+exp(-(time-ti)./0.058)))*rho1;
theta=(1./(1+exp(-(time-ti)./0.058)))*theta1;
elseif segin(i) == 2
  if p(1) <= 34 && p(2) <= 27.5
    iout=1;
    to=time+.4;
  else
    iout=i;
    to=ti;
  end
  rho=(((1-(1./(1+exp(-(time-ti)./0.058)))))*rho1)+((1./(1+exp(-(time-ti)./0.058)))*rho2);
  theta=(((1-(1./(1+exp(-(time-ti)./0.058)))))*theta1)+((1./(1+exp(-(time-ti)./0.058)))*theta2);
else
  if p(1) >= 34 && p(2) <= 22
    iout=i+1;
    to=time+.4;
  else
    iout=i;
    to=ti;
  end
  rho=(((1-(1./(1+exp(-(time-ti)./0.058)))))*rho2);
  theta=(((1-(1./(1+exp(-(time-ti)./0.058)))))*theta2);
end

4. Function “segmap” – This was the function used to create the raw compensation maps based on the collected subject data.

%% Function SEGMAP takes collected trajectory data and maps wrist orientation angles to the x,y,z position the wrist was at.
% An arbitrarily sized 3-d volume is created first before values are set.
% Output MAPCOUNT extracts the locations that are used as part of
% the path. Output MVOLOUT rewrites VOLUMEOUT using the
mean orientation at
% each position. Output MVOLOUTSTAT provides info including
the number of
% orientations at each position and the standard deviation
between the
% orientations.

function
[volumeout,mapcount,mvolout,mvoloutsum,mvoloutstat]=segmap(
data,t,ymin,xmax,volumein,or)

%Mapping orientations from raw data
for i=1:length(xmax)
    for j=xmax(i,4):t(2*i)
        % [xmax(i,1):ymin(i,2),xmax(i,2):ymin(i,3)
        %,xmax(i,3):ymin(i,4)]
        % [t(2*i-1):xmax(i,1),ymin(i,2):xmax(i,2),ymin(i,3):xmax(i,3),ymin(i,4):xmax(i,4)]
        n=1;
        x=round(data(1,j)/10);
        y=round(data(2,j)/10);
        z=round(data(3,j)/10);
        while volumein(x,y,z,n) ~= 0;
            n=n+1;
            if size(volumein,4) < n
                volumein(:,:,n)=zeros(60,60,30);
            else
                end
            end
            volumein(x,y,z,n)=data(or,j);
        end
    end
end

volumeout=volumein;

%Extracting all positions on the used paths.
q=1;
for l=21:56
    for m=1:60
        for p=1:30
            end
        end
    end
end
if volumeout(l,m,p,1) ~= 0
    mapcount(q,:)=[l,m,p];
    q=q+1;
else
    end
end
end

end

%Creating mapped volume with mean orientations.
mvolout=zeros(60,60,30);
mvoloutsum=zeros(60,60,30);

for i=1:length(mapcount)
    j=1;
    while volumeout(mapcount(i,1),mapcount(i,2),mapcount(i,3),j) ~= 0
        && j < size(volumeout,4)
            j=j+1;
    end

    mvolout(mapcount(i,1),mapcount(i,2),mapcount(i,3))=sum(volumeout(mapcount(i,1),mapcount(i,2),mapcount(i,3),:))/(j-1);
    mvoloutsum(mapcount(i,1),mapcount(i,2),mapcount(i,3))=sum(volumeout(mapcount(i,1),mapcount(i,2),mapcount(i,3),:));
    mvoloutstat(i,:)=[mapcount(i,:),j-1,std(volumeout(mapcount(i,1),mapcount(i,2),mapcount(i,3),1:j-1)),mvolout(mapcount(i,1),mapcount(i,2),mapcount(i,3))]
end

5. Function massmatrix – Calculated inertia of forearm and hand based on anthropometric proportionalities fit to each subject.

function Mq = massmatrix(q)

%%Input Metrics
height=(1.6764); %m
weight=(72.57); %kg
hand_length=(.175); %m
hand_width=(.082); %m
hand_thick=(.028); %m

% s1=sind(q(1));
% c1=cosd(q(1));
\% s2=sind(q(2));
\% c2=cosd(q(2));
\% s4=sind(q(4));
\% c4=cosd(q(4));
\% s5=sind(q(5));
\% c5=cosd(q(5));
s6=sin(q(1));
c6=cos(q(1));
s7=sin(q(2));
c7=cos(q(2));

m6=.4242; \% kg
m7=.2256; \% kg

\% in \ m
a6cm=.02382;
d6cm=.04397;
a7cm=-.00458;
d7cm=-.00285;
d6=.06708;
a6=.03952;
a5=.12051;
d4=.04586;
a2=.37546;
d1=.33655;

\% Forearm Inertia
forearm_length=height*.1585;
forearm_mass=weight*.016;
forearm_volume=forearm_mass/1062;
forearm_radius=sqrt(forearm_volume/(pi*forearm_length));

arm_zcom=d6-(forearm_length*.568);
arm_xcom=a6-forearm_radius;
arm_ycom=0;

kx_forearm=forearm_radius*.276;
ky_forearm=forearm_radius*.265;
kz_forearm=forearm_length*.121;
Ixx_forearm=forearm_mass*kx_forearm^2 +
forearm_mass*arm_zcom^2;
Iyy_forearm=forearm_mass*ky_forearm^2 +
forearm_mass*(arm_xcom^2 + arm_zcom^2);
Izz_forearm=forearm_mass*kz_forearm^2 +
forearm_mass*arm_xcom^2;
Ixy_forearm=0;
Ixz_forearm=forearm_mass*arm_xcom*arm_zcom;
Iyz_forearm=0;
I_forearm=[Ixx_forearm, -Ixy_forearm, -Ixz_forearm;
-Ixy_forearm, Iyy_forearm, -Iyz_forearm; -Ixz_forearm, -
Iyz_forearm, Izz_forearm];

%Hand Inertia
hand_xcom=0;
hand_ycom=-hand_length*.468;
hand_zcom=hand_width/2;
hand_mass=weight*.00575;

kx_hand=hand_thick*.235;
ky_hand=hand_length*.184;
kJz_hand=hand_width*.288;
Ixx_hand=hand_mass*kx_hand^2 + hand_mass*(hand_ycom^2 +
hand_zcom^2);
Iyy_hand=hand_mass*ky_hand^2 + hand_mass*hand_zcom^2;
Izz_hand=hand_mass*kz_hand^2 + hand_mass*hand_ycom^2;
Ixy_hand=0;
Ixz_hand=0;
Iyz_hand=hand_mass*hand_ycom*hand_zcom;

I_hand=[Ixx_hand, -Ixy_hand, -Ixz_hand; -Ixy_hand, Iyy_hand, -
Iyz_hand; -Ixz_hand, -Iyz_hand, Izz_hand];

I6=[2.259e-3 -7.049e-5 -9.252e-5; -7.049e-5 2.091e-3 -
4.603e-4; -9.252e-5 -4.603e-4 8.911e-4];
I7=[7.022e-4 8.207e-5 2.535e-5; 8.207e-5 1.675e-4 -1.046e-
4; -2.535e-5 -1.046e-4 6.273e-4];

%Forward Kinematics for Positioning
% x=-a5*c1*s2*c5 - a5*s5*(s1*c4 - c1*c2*s4) - d4*c1*s2 +
d3*c1*c2 + a2*c1*c2;
% y=-a5*s1*s2*c5 + a5*s5*(c1*c4 + s1*c2*c4) - d4*s1*s2 +
d3*s1*c2 + a2*s1*c2;
% z=-a5*c2*c5 - a5*s2*s4*s5 - d4*c2 - d3*s2 - a2*s2 + d1;

%Modified Center of Mass Location
com6=[a6cm; -.01547; d6cm]*m6 +
[arm_xcom; arm_ycom; arm_zcom]*forearm_mass/(m6 +
forearm_mass);
com7=[a7cm; -.03363; d7cm]*m7 +
[hand_xcom; hand_ycom; hand_zcom]*hand_mass/(m7 + hand_mass);
%Jacobian
Jv1=[-com6(1)*s6,0;com6(1)*c6,0;0,0];
Jv2=[-com7(1)*c6*c7 + com7(3)*s6 - a6*c6,com7(1)*s6*s7 - com7(1)*s6*c7 - com7(3)*c6 - a6*s6,-com7(1)*c6*s7;0,-com7(1)*c7];
Jw1=[0,0;0,0;1,0];
Jw2=[0,-c6;0,-s6;1,0];

%Rotation Matrix
R6=[-s6,0,-c6;c6,0,-s6;0,1,0];

D6=m6*(Jv1'*Jv1)+Jw1'*(I6 + I_forearm)*Jw1;
D7=m7*(Jv2'*Jv2)+Jw2'*R6*(I7 + I_hand)*R6'*Jw2;
D=D6+D7; %Mass Matrix
Mq=D;

6. data_trigger.m – This script was used to separate task iterations from a test run by identifying the instances where the device left and returned to the Home position.

% Post-Processing Data Trigger
% Devon Holley
% Created 10/27/14

% Home Dimensions (x=290-340, y=225-275)
x1=290;
x2=340;
y1=160;
y2=275;

data=input('Enter Vector Name: ');
n=1;

clear trigger
for i=2:length(data)
    if data(i,1) < x2 && data(i-1,1) >= x2
        if data(i,2) < y2 && data(i,2) > y1
            trigger(n)=i;
            n=n+1;
        end
    elseif data(i,1) > x2 && data(i-1,1) <= x2
        if data(i,2) < y2 && data(i,2) > y1

    end
end
trigger(n)=i;
n=n+1;

end

elseif data(i,1) > x1 && data(i-1,1) <= x1
    if data(i,2) < y2 && data(i,2) > y1
        trigger(n)=i;
        n=n+1;
    end
end

elseif data(i,1) < x1 && data(i-1,1) >= x1
    if data(i,2) < y2 && data(i,2) > y1
        trigger(n)=i;
        n=n+1;
    end
end

elseif data(i,2) < y1 && data(i-1,2) >= y1
    if data(i,1) < x2 && data(i,1) > x1
        trigger(n)=i;
        n=n+1;
    end
end

elseif data(i,2) > y1 && data(i-1,2) <= y1
    if data(i,1) < x2 && data(i,1) > x1
        trigger(n)=i;
        n=n+1;
    end
end

elseif data(i,2) < y2 && data(i-1,2) >= y2
    if data(i,1) < x2 && data(i,1) > x1
        trigger(n)=i;
        n=n+1;
    end
end

elseif data(i,2) > y2 && data(i-1,2) <= y2
    if data(i,1) < x2 && data(i,1) > x1
        trigger(n)=i;
        n=n+1;
    end
end

end

end

disp(trigger')
APPENDIX I – ENGINEERING DRAWINGS

Orthosis-Arm Interface Subassembly:

<table>
<thead>
<tr>
<th>ITEM NO.</th>
<th>PART NUMBER</th>
<th>TITLE</th>
<th>QTY</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>CUSTOM MACHINED</td>
<td>YAW EXTENSION PLATE</td>
<td>1</td>
</tr>
<tr>
<td>2</td>
<td>CUSTOM MACHINED</td>
<td>YAW EXTENSION</td>
<td>1</td>
</tr>
<tr>
<td>3</td>
<td>MACMASTER CARTRIDGE</td>
<td>DOUBLE SHIELDED ROLLER BEARING WITH EXTENDED INNER RING</td>
<td>4</td>
</tr>
<tr>
<td>4</td>
<td>CUSTOM MACHINED</td>
<td>YAW INITIATOR</td>
<td>1</td>
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<tr>
<td>5</td>
<td>CUSTOM MACHINED</td>
<td>LEFT INTERFACE CONNECTOR PLATE</td>
<td>1</td>
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<tr>
<td>6</td>
<td>CUSTOM MACHINED</td>
<td>INTERFACE PLATE</td>
<td>1</td>
</tr>
<tr>
<td>7</td>
<td>CUSTOM MACHINED</td>
<td>PITCH INITIATOR</td>
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<tr>
<td>8</td>
<td>CUSTOM MACHINED</td>
<td>RIGHT INTERFACE CONNECTOR PLATE</td>
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<tr>
<td>9</td>
<td>CUSTOM MACHINED</td>
<td>PITCH AXLE</td>
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<td>10</td>
<td>MACMASTER CARTRIDGE</td>
<td>RUBBER O-RING</td>
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<tr>
<td>11</td>
<td>MITUTOYO POTentiOMETER</td>
<td>SNAP</td>
<td>2</td>
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<tr>
<td>12</td>
<td>3D PRINT</td>
<td>POCKET HOUSING</td>
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</tr>
<tr>
<td>13</td>
<td>3D PRINT</td>
<td>POCKET HOUSING (LONG)</td>
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<td>14</td>
<td>MACMASTER CARTRIDGE</td>
<td>6-32 X .25 SOKET HEAD CAP SCREW</td>
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<td>15</td>
<td>MACMASTER CARTRIDGE</td>
<td>6-32 X .25 SOKET HEAD CAP SCREW</td>
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<td>16</td>
<td>MACMASTER CARTRIDGE</td>
<td>6-32 X .25 SOKET HEAD CAP SCREW</td>
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<tr>
<td>17</td>
<td>MACMASTER CARTRIDGE</td>
<td>8-32 X .25 FLAT HEAD PHILLIPS MACHINE SCREW</td>
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<td>18</td>
<td>MACMASTER CARTRIDGE</td>
<td>1-032 X 26 SET SCREW</td>
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SCALE: 10:1

Orthosis-Arm Interface Subassembly
Orthosis Full Assembly:

<table>
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<th>ITEM</th>
<th>QTY</th>
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<td>1</td>
<td>SUBASSEMBLY</td>
<td>ORTHOSIS ARM INTERFACE SUBASSEMBLY</td>
<td>1</td>
</tr>
<tr>
<td>2</td>
<td>SUBASSEMBLY</td>
<td>ORTHOSIS OMNI SUBASSEMBLY</td>
<td>1</td>
</tr>
<tr>
<td>3</td>
<td>SUBASSEMBLY</td>
<td>ORTHOSIS HAND SUBASSEMBLY</td>
<td>1</td>
</tr>
<tr>
<td>4</td>
<td>MCMASTIC CAPB VHN 9 1234 5678</td>
<td>6-32 X 0.25&quot; SOCKET HEAD CAP SCREW</td>
<td>4</td>
</tr>
<tr>
<td>5</td>
<td>MCMASTIC CAPB VHN 9 1234 5678</td>
<td>6-32 HEX NUT</td>
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</tr>
<tr>
<td>6</td>
<td>MCMASTIC CAPB VHN 9 1234 5678</td>
<td>6-32 H. 5&quot; BUTTON HEAD SOCKET CAP SCREW</td>
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SCALE: 1:2.5