REDUCING DELAY TIME IN HAND MENTOR PRO TO IMPROVE CLINICAL FUNCTIONALITY

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REDUCING DELAY TIME IN HAND MENTOR PRO TO IMPROVE CLINICAL FUNCTIONALITY

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ABSTRACT

Stroke is third leading cause of disability worldwide, and the burden of stroke across the world is steadily growing. However, it is the leading cause of disability in the United States. Each year, 800,000 people in the United States alone have strokes. This has necessitated the development of more effective methods of stroke rehabilitation.

The Hand Mentor Pro is an assistive robotic device used stroke rehabilitation. Through the use of interactive goal-oriented games, it can deliver high intensity repetitive treatment. Currently, the Hand Mentor is actuated by a McKibben air muscle. This actuator has a significant lag of 1.85 milliseconds, presenting a serious obstacle to various clinical outcomes of the device. The aim of this study was to develop and test a novel actuation system that combines the Hand Mentor's air muscle with a DC motor in order to reduce actuation delay time.

Two experimental actuator treatments, DC motor and the combination of DC motor and air muscle showed to be statistically different from the control with p-values of 1.0320×10^{-4} and 8.7949×10^{-5}, respectively. Despite a significant decrease in delay time, the redesigned actuator system still has a delay greater than the normal range of stroke patients' affected limb wrist flexion. This indicates that further design and exploration is required for a clinically ideal solution. Nevertheless, the delay reduction achieved in this project can improve many of the clinical functionalities of the Hand Mentor by better synchronizing patients' effort with the active assistance of the Hand Mentor, as well as its extrinsic feedback.

CHAPTER 1

INTRODUCTION

Overview

Stroke is third leading cause of disability worldwide (Feigin, Forouzanfar et al.) However, it is the leading cause of disability in the United States (Krebs, Palazzolo et al. 2003). Each year, 800,000 people in the United States alone have strokes (Stinear, Byblow et al. 2014). The burden of stroke in across the world is steadily growing(Feigin, Forouzanfar et al.). There is a rising need for effective and affordable treatment. Great strides have been taken in learning how to rehabilitate patients after stroke, but there is still much to be learned.

Stroke is a disorder caused by damage of neurons in the brain due to poor or absent oxygen flow. There are two primary kinds of stroke, ischemic and hemorrhagic. Ischemic is caused by an obstruction of blood flow to the brain, while hemorrhagic stroke is caused by bleeding within the brain. In both cases, cells in the under-oxygenated regions die, causing further inflammation and edema. This ultimately leads to increased damage of the cells in the affected region (Caleo 2015). Neuronal loss results in neuromuscular dysfunction in the form of loss of motor control on one side of the body. This is a condition known as hemiparesis.

The aim of neural rehabilitation is recovery lost function via the same effectors that were formerly lost (Stinear, Byblow et al. 2014) **.**The period within three months after stroke is the period in which patients experience the most spontaneous recovery (Caleo 2015). During this time, the contralesional hemisphere of the patient's brain undergoes a period of heightened excitability to compensate for loss of excitability on one side of the brain. In fact, an important part of the stroke therapy consists of constraining the healthy arm in order to promote the activity in the lesioned hemisphere (Langhorne, Coupar et al. 2009). This is to reduce the chance of maladaptive compensation strategies or learned non-use that ultimately interfere with the recovery of lost function on the affected side of a patient's body (Oujamaa, Relave et al. 2009).

There are several key components of successful neural rehabilitation regimes. First, an intervention must utilize for high-volume repetition. Increased intensity and repetition of treatment or exercise is strongly correlated with improved patient outcomes during the motor rehabilitation process and greater generalizability of relearned motor functions (Arya, Pandian et al. 2011). A meta-analysis conducted by Langhorne and colleagues shows that this is a unifying feature among many effective stroke rehabilitation interventions (Langhorne, Coupar et al. 2009). Secondly, motion must be active not passive, meaning the patient must attempt to move on his or her own, regardless he or she is receiving external assistance. This is because a patient's engagement and effort is improved by being able to influence the way an ongoing movement occurs (Rosati, Bobrow et al. 2008). Thirdly, there must be extrinsic feedback to motivate the patient's movements. Extrinsic feedback consists of two primary components— knowledge of results, and knowledge of performance. Knowledge of results indicates the outcome of a task, or whether a goal was reached. Knowledge of performance addresses the characteristics of an action that led to the result. There is no single rehabilitation regime that is ideal for all patients. For this reason, many successful stroke rehabilitation robotic devices employ goal-directed games that target specific motor impairments (Krebs, Palazzolo et al. 2003). Research has shown that attention to the results of one's movements is more beneficial to motor learning than direct attention paid to the movement itself, making the use games the ideal platform for implementing high-intensity repetitive interventions (Wulf and Prinz 2001)**.**

Problem and Rationale

The Hand Mentor Pro is a robotic device designed to improve upper-extremity rehabilitation in stroke patients. The motivation for this kind of device is that robots allow greater levels of training intensity and repetition that may be strenuous or unfeasible for physical therapists to conduct. Apart from being an assistive robotic device, The Hand Mentor Pro also utilizes interactive games that serve as motivators for patient engagement, but also facilitate both knowledge of results and knowledge of performance.

Currently, the Hand Mentor experiences an average delay time of 1.85 seconds due to the time it takes to pressurize its McKibben air muscle system. When a new patient is learning to use the Hand Mentor he/she must understand various parameters, such as wrist position, speed, and timing, in order to successfully play the games. One of the most critical factors is coupling the temporal sequence of wrist motions with meaningful performance in the game. However, knowledge of performance and results might be underutilized if they do not match with the expectations of the patient. In other words, lag in the Hand Mentor's actuation causes lag in the game play which adds an additional parameter for the participant to learn. This detracts from the use of the device's intended extrinsic feedback. Such a long delay time also makes it impossible to synchronize patients' efforts with robotic assistance, resulting in a mainly passive relationship with the device's assistive force.

The goal of this design project was to design a modification to the actuator system of the current the Hand Mentor Pro to make a clinically significant reduction in delay time. In order to be a clinically meaningful improvement, two conditions must be met. First, the latency time must be within the normal range of stroke patient affected limb reaction times. This is to ensure that the actuator assists patients at the time assistance is required. Secondly, the new actuator system must retain compliance in the direction of actuation. Compliance is required to retain the causal relationship between patient effort and movement, even when active assistance from the robot is provided (Rosati, Bobrow et al. 2008).

A study comparing wrist flexion and extension reaction times on the ipsilesional and contralesional limbs of stroke patients found that wrist extension on patients' affected side exhibits a reaction time significantly greater than that on their non-affected sides. The average wrist extension reaction time for stroke patients' affected limb is 310.8 milliseconds, with a standard deviation of 164.6 milliseconds (Bi and Wan 2013).

CHAPTER 2

ENGINEERING MODIFICATIONS OF THE HAND MENTOR PRO

Mechanical Modifications

The aim of the design phase of this project was to design an actuator system that can significantly reduce the delay time of actuation of the Hand Mentor Pro. This had to be done in a way that could be controlled, and retains the movement and function of the current Hand Mentor design. In order to fulfill these requirements, two key modifications to the original structure of the Hand Mentor were required. First, a motor needed to be mounted onto the Hand Mentor. Second, a way to incorporate the force of that motor into the motion of the device was developed.

A modified t-bar was attached to the Hand Mentor's air muscle, to carrying the motor (Figure 1a). A pulley wheel was then attached to the shaft of the motor. A cable attached to the wheel was attached to the same anchor that was attached to the air muscle in the original Hand Mentor (Figure 1b). The wheel anchor allowed the motor to assist or supplant the air muscle during the actuation of the Hand Mentor's wrist joint.

Figure 1. Changes to Mechanical Design of Hand Mentor Pro

Figure 1a.

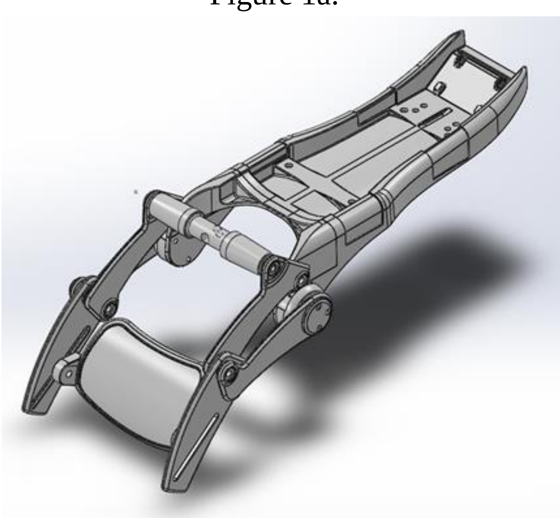


Figure 1b

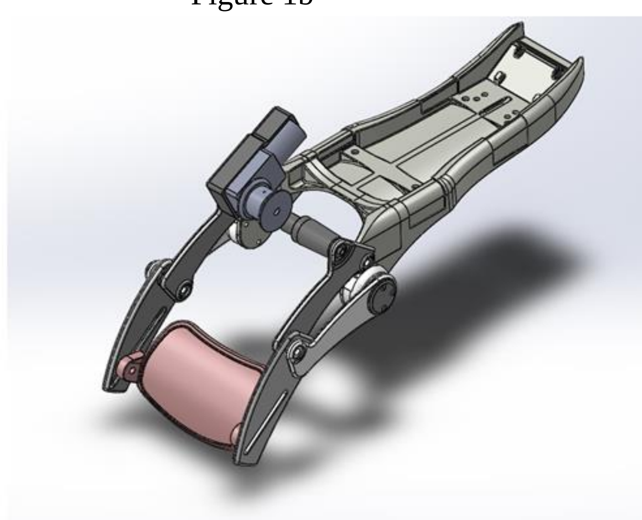


Figure 2. Steady State Analysis of Inner Metacarpal

Figure 2a.

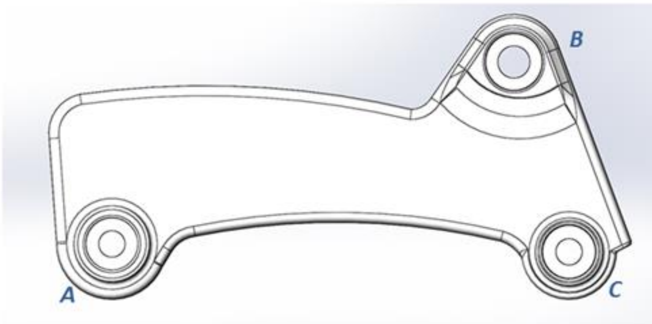
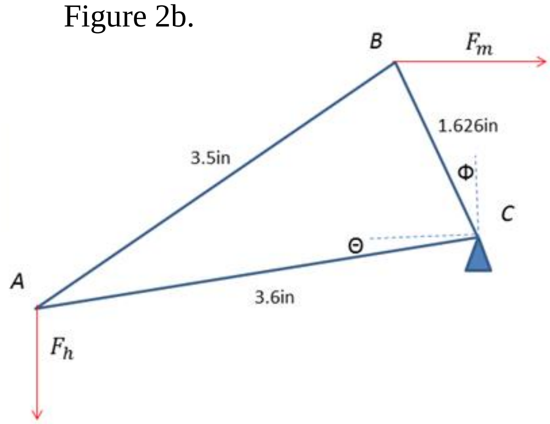


Figure 2b.



Selecting an Appropriate Motor

To select an appropriate motor for the design, an inspection of one of the Hand Mentor's key load-bearing components, the inner metacarpal, was required (Figure 2a). The inner metacarpal of the Hand Mentor can be modeled as a triangle with side lengths of 3.5, 3.6, and 1.626 inches (Figure 2b). The force from the motor, applied at point B, counteracts the downward force of the hand, at point A. In order to cause angular acceleration in the Hand Mentor's wrist joint, the torque caused by the motor must be greater than the torque caused by the hand about point C.

Equation 1. Calculating Motor Torque Requirements

$$\boldsymbol{\sum M = M_m + M_h}$$

$$\boldsymbol{M_m = (1.626)F_m cos(\varphi)}$$

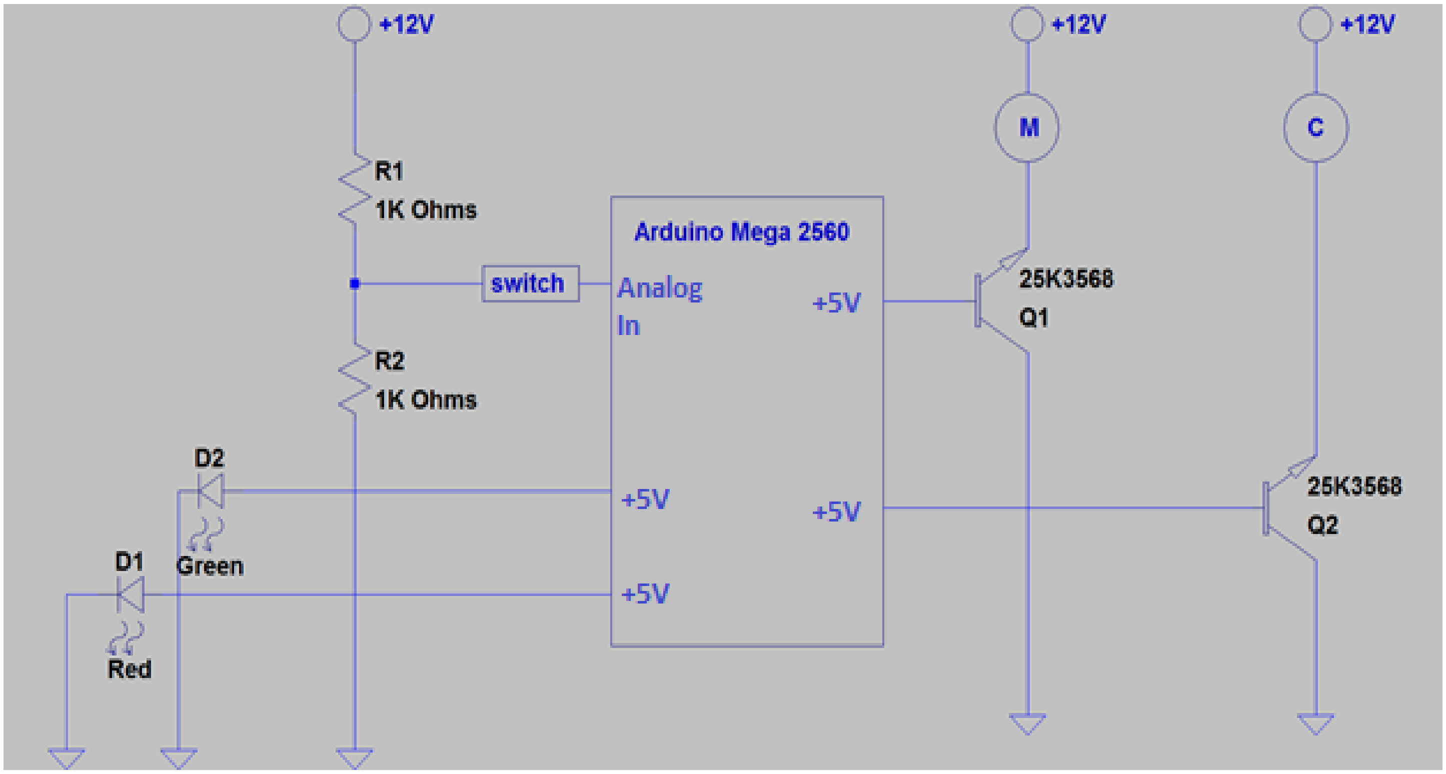
$$\boldsymbol{M_h = (3.6)F_h cos(\theta)}$$

To find the torque, a steady-state analysis was done (Equation 1). The maximum value of the torque resulting from F_h occurs when the angles Φ and θ are both equal to 0°. Assuming the magnitude of F_h is 2.0 lb_f, F_m must be greater than 4.43 lb-in. Since the radius of the pulley wheel mounted on the modified t-bar is 1.0 inch, then the torque delivered by the motor must be 4.43 pound-inches. The motor selected was a 12V-DC motor capable of 5.21lb-in (6kg-cm) and 24 rpm. The motor with these parameters is ideal for this application because it operates in the range of torques required by the device and also has low revolutions per minute, allowing movement of the Hand Mentors wrist joint, without the need for gears to step down the revolutions per minute.

Electronics Design of Test Bench

In order to compare actuator treatments for this project, the test bench for this device must be able to activate the air muscle and the motor, simultaneously and separately. The beginning of each trial must be triggered by the experimenter. Figure 3 shows the circuit schematic for test bench. When the switch is pressed, HIGH signals are sent from the Arduino Mega 2560 to the power transistors gating the motor and the compressor, labeled M and C on the schematic, respectively. The high signal causes the motor and compressor to receive +12V. The motor receives a 2.0 second pulse, while the air-muscle receives a 4.0 second pulse of +12V. The result is actuation of the Hand Mentor Pro. The green and red LEDs received the same pulse lengths as the motor and compressor, respectively. The synchronized output of the actuators and LEDs facilitated debugging during the prototyping and experimentation phase.

Figure 3 – Circuit schematic of the test bench:



CHAPTER 3

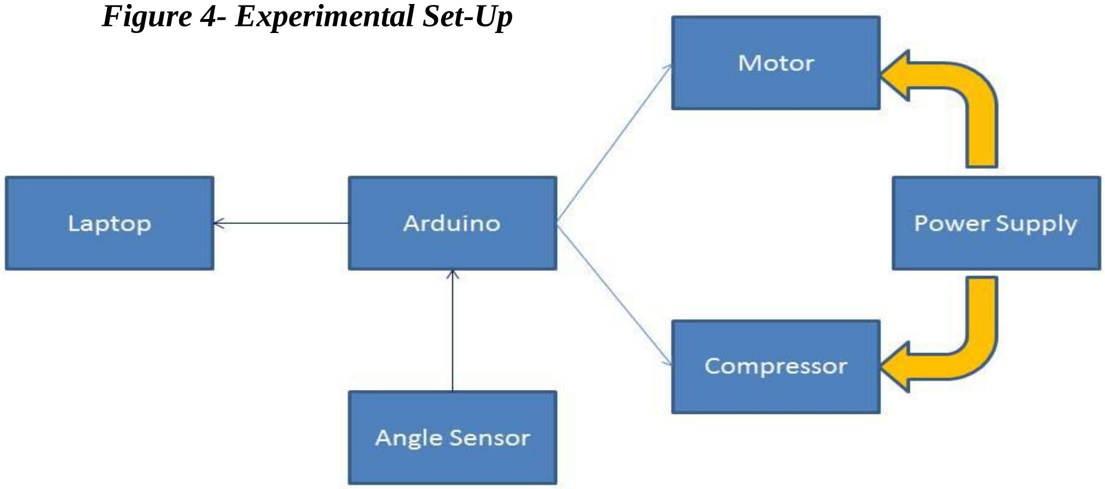
EXPERIMENTAL DESIGN

Experimental Method

The hypothesis tested in this project was that an actuator system could significantly reduce the delay time of the Hand Mentor, and that this reduction in delay time was enough to not interfere with rehabilitation outcomes.

The experiment contained two independent variables, actuator treatment and load. There were three actuator treatments. The first treatment used a combination of air muscle and motor to actuate. The second treatment used just the motor, and the third used just the air muscle. The third group functioned as the control treatment. There were 4 different weight treatments, 0 lb, 1.0lb, 0.5lb, and 0.75lb. These weights were placed in the hand-piece of the Hand Mentor to model the weight of a patient's hand during use. These treatment pairings resulted in 60 different trials. Because not all trials were done with the same air muscle, treatment trials were paired to corresponding control trials in order to control for the variability that could result from using different air-muscles. During each experimental trial, the Hand Mentor's built-in angle sensor was used to collect angular displacement data during each trial. Angle measurements from the potentiometer taken once every 10 milliseconds underwent a 10-bit analog-to-digital conversion in the Arduino Mega 2560. Data was then sent to a laptop where it was plotted and stored in MATLAB. Figure 4 shows how these components were connected during data collection. If the air muscle was not needed during a particular trial, then the compressor was disconnected from the air-muscle, but not from the power-supply. This allowed both actuator treatments to be tested using the same Arduino program, and electrical circuit.

Figure 4- Experimental Set-Up



Data Analysis

Prior to comparison among the actuator treatment groups, the data was tested for normality using the Kolmogorov-Smirnov test for normality. The null hypothesis of this test is that these samples come from a normally distributed population. After verifying the non-normality of all the collected data, the non-parametric Wilcoxon Signed-Rank Test for paired trials was used to test for statistically significant differences between motor groups. This tests the distribution of differences between the treatment and control under the null hypothesis that its mean is zero.

In order to test if the distribution of delay times was within the appropriate range for clinically significant improvement on the Hand Mentor's delay time a 95% confidence interval of the mean delay time for each treatment group was calculated, and analyzed. The selected target range was within one standard deviation from the mean reaction time for wrist extension in stroke patients' ipsilesional limbs, (310.8+/-164 milliseconds).

CHAPTER 4

RESULTS AND ANALYSES

Figure 5. Average Trajectory of each treatment

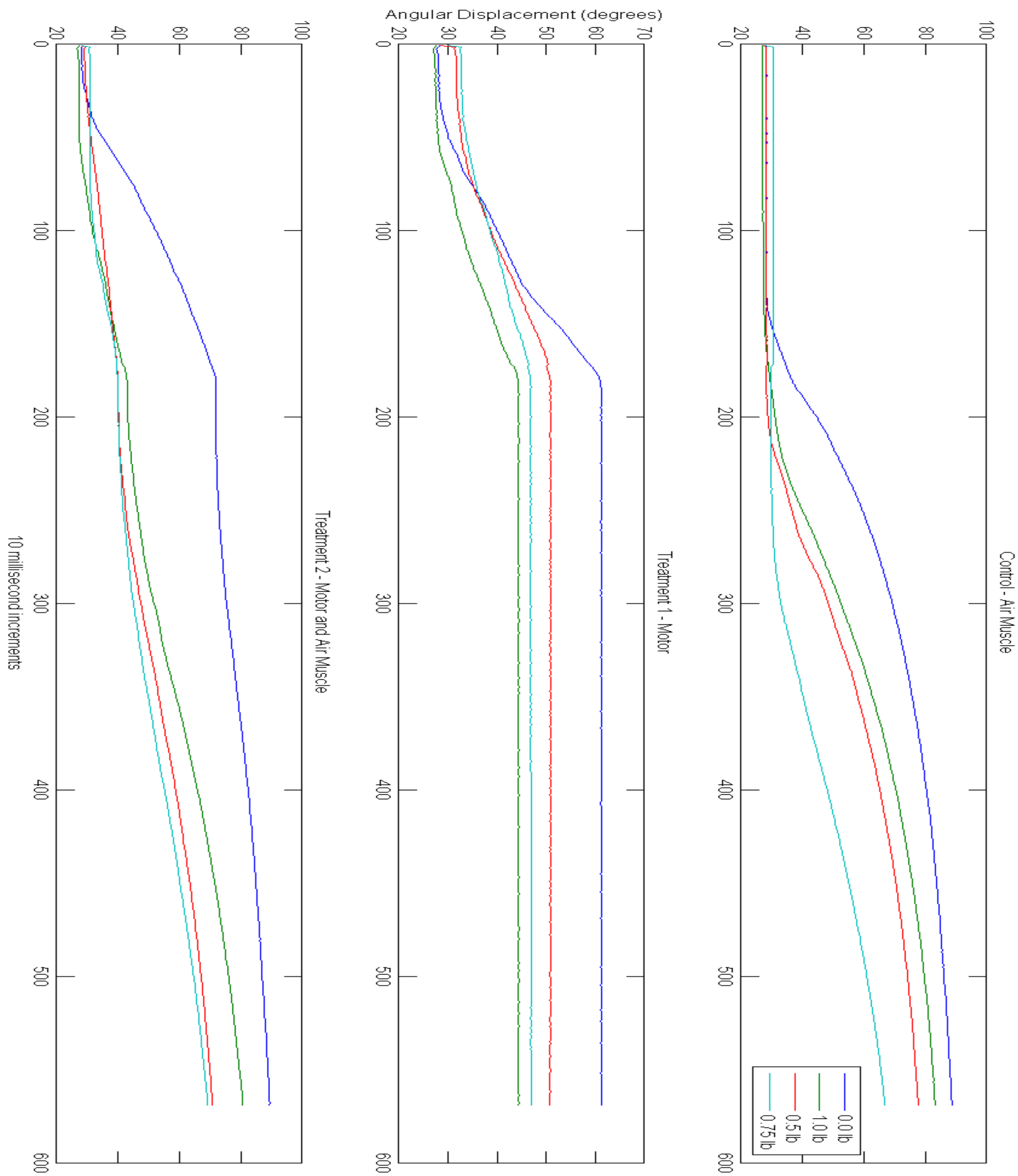


Figure 6a. Histograms of Delay Times for Each Actuator Treatment

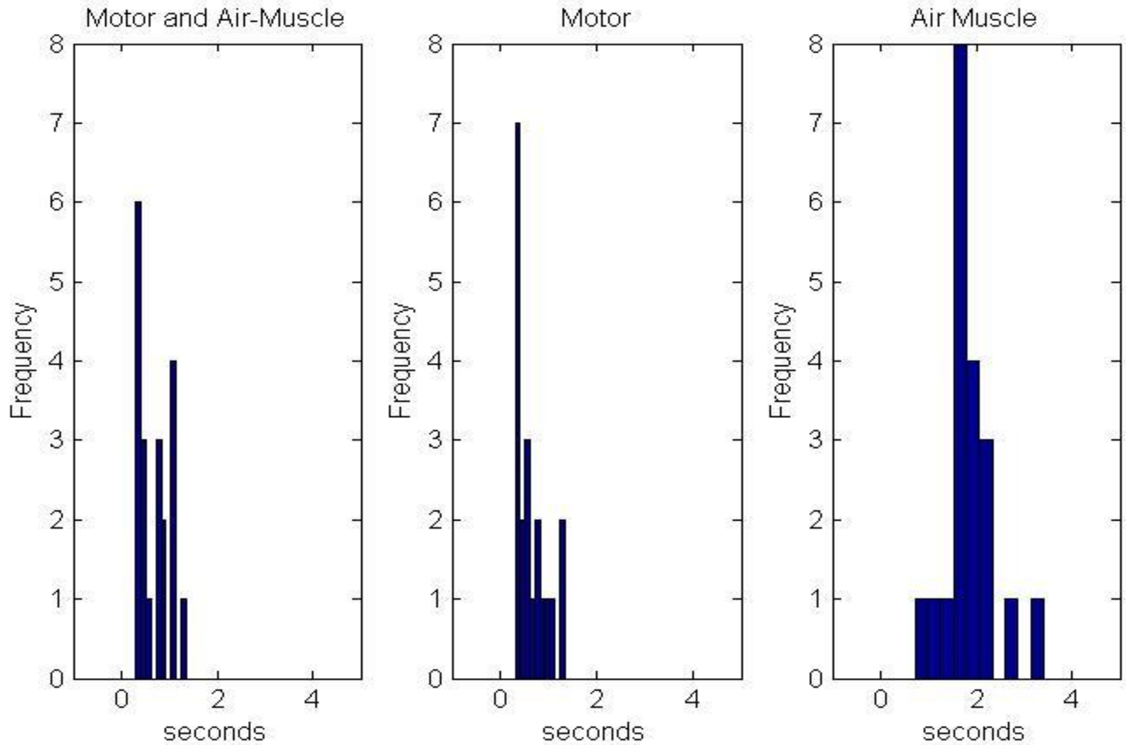


Figure 6b. Histograms of Differences between actuator treatments and control

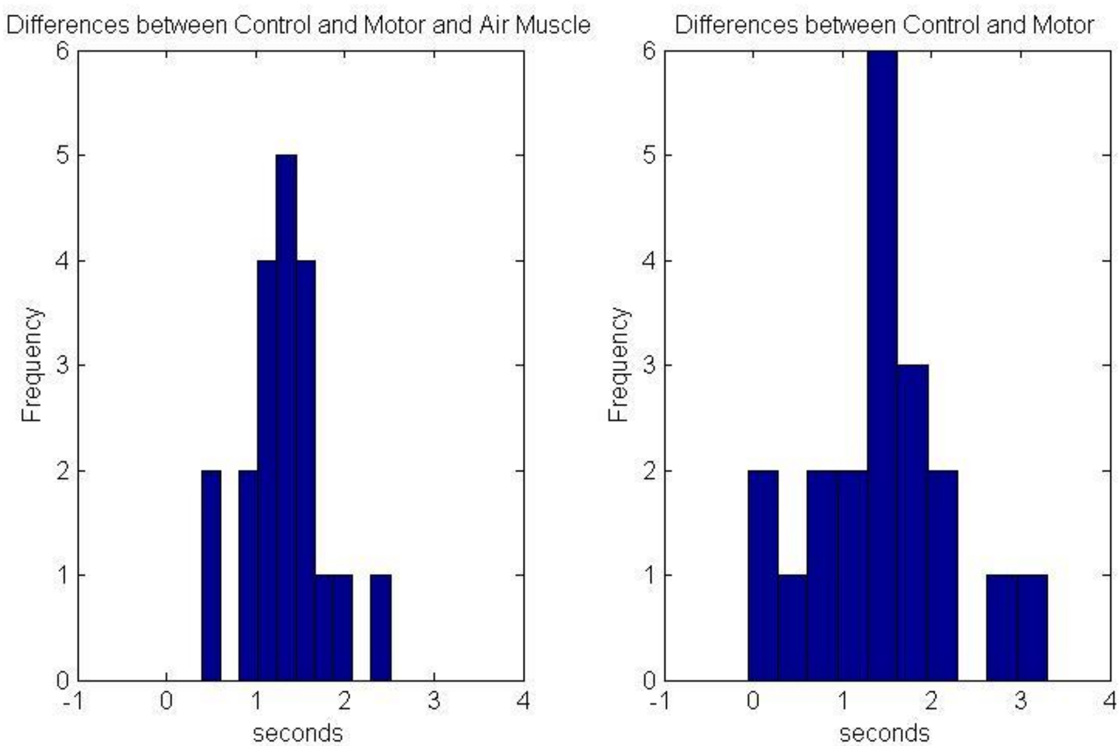


Figure 5 shows the apparent differences in latency among the different actuator treatments. Across all weight treatments, the actuator treatments using the motor are always quicker to initiate than the control treatment. The distributions of delay times for Motor and Air Muscle, the Air Muscle, and Motor groups trials were confirmed to be non-normal with Kolmogorov-Smirnov p-values of 1.262×10^{-7}, 1.05×10^{-13}, and 1.26×10^{-7}, respectively (Figure 6a and 6b). Both treatment groups, motor, and motor and Air Muscle were found to have statistically significant differences in delay time with the control. The Motor and Motor and Air Muscle groups had p-values of 1.0320×10^{-4} and 8.7949×10^{-5}, respectively. This shows that the addition of the motor to the Hand Mentor Pro produced a significant reduction in delay time.

Table 1. The 95% confidence Intervals for Means Actuator Treatment Groups

	Lower Bound (milliseconds)	Upper Bound (milliseconds)
Control	1578	2123
Motor and Air Muscle	522	841
Motor	473	780

Table 1 shows the 95% confidence intervals for the means of each of the actuator treatment groups. The target range for optimal delay times was between 146 and 474 milliseconds. These are the bounds within one standard deviation of the average wrist extension reaction time in stroke patients' affected limbs (Bi and Wan 2013). As can be seen from the table, the confidence intervals of the experimental actuator treatments are significantly outside of the bounds of the control treatment. However, neither of the experimental treatments had any overlap with the target interval of 310+/-164 milliseconds. This demonstrates that even though the new actuator design creates a notable reduction in actuation latency, the change was not great enough to fall within the range of ideal values for maximum clinical benefit.

CHAPTER 5

DISCUSSION AND CONCLUSION

Discussion

This project successfully developed and verified that the use of a short burst of torque from a DC motor can significantly reduce the delay in the actuation time of the Hand Mentor Pro. However, the resulting reduction in latency time was not less than the average wrist extension response time of stroke patients, 310 milliseconds (Bi and Wan 2013). This shows that in spite of having reduced the Hand Mentor's actuator latency time, this solution is still not within the bounds of an ideal latency time for clinical applications, and further investigation and design iteration is warranted. Nevertheless, the improvement in delay is easily perceptible, which prevents discrepancies between patient's knowledge of results and performance, and the extrinsic feedback provided by the Hand Mentor itself. Another possibility that this design enables is the opportunity for coincidence of patient effort and robotic assistance. This will facilitate games such as the sine-wave tracking task. This game is often frustrating for patients because assistance is often required to fulfill the task, but the current Hand Mentor is too slow to actively assist the task of wrist flexion. The reduced actuation delay time will allow for real-time assistance and improved performance in this game. Because the updated design retains compliance in the direction of upwards actuation, there can be an active cooperation between the patient's wrist extension and the Hand Mentor's actuation.

The clearest limitation of this study was the lack of human testing. Although a significant improvement in time delay has been accomplished, it is still uncertain whether the solution in its current stage is sufficient to fully solve the problems associated with the delay of actuation in the Hand Mentor. Proprioception and other modes of sensory feedback play an important role in the recovery of motor function after stroke (Massie, Kantak et al. 2015). Because of this, the appropriate speed and force used to move a patient's arm is not arbitrary, and should be adaptive to the patient's own motion and patterns. Currently, this device design merely applies a force at the time of actuation, but would not have any real-time feedback. This could result in issues such as a mistiming of actuation with patient efforts. This possibility is corroborated by the results of this study, which showed that the 95% confidence interval for the motor and motor and muscle actuator treatments are completely beyond the likely of most stroke patient's wrist extension reaction time. This could result in a mismatch of knowledge of performance and knowledge of results due to the device moving independent of patients' intentions.

Future exploration of this design modification could include the development of closed-loop feedback from the Hand Mentor's angle sensor, and games that take advantage of the new design's decreased delay time. Additionally, the possibility of simultaneous patient effort and robotic assistance necessitates a means of measuring the amount of patients' contribution in instances where both the patient and the Hand Mentor are jointly engaged in wrist extension. This would offer a new mode of feedback for patient and clinicians to better understand patient outcomes.

Conclusion

This study successfully demonstrated that the use of a DC motor mounted on the Hand Mentor Pro significantly reduces the delay time in actuation. However, further decreases in latency may produce even greater improvements to the Hand Mentor's functionality. The current design does not reach the clinically ideal latency range.

However, the reduction in actuation latency time should allow for a noticeable decrease in discrepancies between patient's knowledge of performance and results, and the Hand Mentor's own extrinsic feedback. This will reduce instances of underutilization of the Hand Mentor's feedback, and lead to greater patient engagement and results. These improvements should also facilitate functionalities formerly impossible, such as real-time active assistance during exercises, which will allow for improved performance and motor learning.

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