Title
Development and Testing of a Novel Class of Device for Rehabilitation and Assistance: Exochairs

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UNIVERSITY OF CALIFORNIA,
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Design and Evaluation of a New Class of Device for Therapy and Mobility: ExoChairs

DISSERTATION

Submitted in partial satisfaction of the requirements
for the degree of

DOCTOR OF PHILOSOPHY

in Biomedical Engineering

by

Yasemin Sarigul-Klijn

Dissertation Committee:
Professor David Reinkensmeyer, Chair
Professor Michael McCarthy
Professor Frithjof Kruggel
Professor Ken Mease

2017

DEDICATION

To

my mom, brothers, grandpa, uncle, aunt, and fiancé

in recognition of their unconditional support

I love you all

To my grandmother, who passed away from a walking disability:
Had I known
Your steps would one day slow to a stop
I would have taken my time and never run ahead on our walks.

You left on a cold January night
Yet I never need know fear

For in my heart

We walk hand in hand

For your love is always
near

This PhD is for you, grandma.

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ACKNOWLEDGMENTS

I would like to thank my mom for her constant support and encouragement through everything in life, as well as my brothers and grandfather, and uncle and aunt.

I extend my appreciations to my lab mates. Justin, Jaime, Sumner, Diogo, Joan Lobo, Joan, Quentin, and my collaborators on the GRAM project: Peter, Dr. McCarthy, and Curt Deckert. Go Red Lion Robotics! I want to thank The Cove at UCI, NSF I-Corps, Doug Crawford, Bill Morris, and everyone ant the Antrepreneur center for their support and help with the GRAM project. I couldn’t have done any of these things without any of you.

I give my thanks to my advisor Dr. Reinkensmeyer for allowing me the opportunity to work in this wonderful field. I’d like to thank my committee, Dr. Kruggel, Dr. Mease, and Dr. McCarthy for their support.

I thank my friends at UCI, Shramana G, Juan G, and Thomas B. I thank my friends Josh D, Jia T, and Mike B. I am grateful you have all been there for me.

I thank my fiancé Michael, whose unconditional love and support has helped me greatly in trudging through difficult times during my PhD.

Most of all, I want to thank my grandmother. You passed away ten years ago when I was 16 and couldn’t walk anymore. Though you no longer step upon this earth, you always walk hand in hand with me in my heart. This PhD is for you. I will always love you, Anne.

CURRICULUM VITAE

EDUCATION

Ph.D., Biomedical Engineering **December 2017**

University of California, Irvine

NSF I-Corp Recipient

M.S., Biomedical Engineering **September 2016**

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B.S., Biomedical Engineering **May 2013**

University of California, Berkeley

Golden Key International Honors Society

EXPERIENCE

CEO, Founder, Principal Engineer, Red Lion Robotics **September 2014-December 2017**
Lead, and manage team in designing and testing robotic devices for walking disability. Architect company vision, act as figurehead and procure funding for company

Graduate Researcher at Reinkensmeyer Laboratory, UC Irvine **September 2014-December 2017**
Develop, design and test robotic devices to help people with disability. Clinical testing of devices with people with stroke. Develop software to control devices. Conduct research and write paper

Mechanical Design Engineer at Kazerooni Laboratory, UC Berkeley **May-December 2013**
Develop testing protocol to assess exoskeleton efficiency; used EMG software to gather data. Redesign/Optimized exoskeleton with CAD. Interfaced with international clients to meet specifications.

Research Assistant at Deviren Laboratory, UCSF **August-December 2013**
Conduct research on paraspinal muscles modic changes post surgery in patients using MRI

Research Assistant at Stastickwiz Laboratory, **August 2012-May 2013**
Data analysis of results using wet lab techniques Performed data mining & analysis of weekly experiments

Programmer on Stats Design Project, UC Berkeley **August 2012-December 2012**
Co-lead a team of 5 members in developing a statistics research report Used web scraping techniques to acquire data; Coded XML skeleton for a Google Earth Visual representation. Created custom graphics

Intern at N&M Engineering Company Davis, CA **December 2008-December 2009**
Create technical drawings, gather feedback from engineers, and modify drawings to meet requirements Proofread engineering publications.

AWARDS AND HONORS

NSF I-Corps Recipient, UC Irvine	2017
Aging 2.0 Orange County Chapter Winner	2017
Beall Hardware Competition 2nd Place	2017
New Venture Competition Finalist	2017
Blum Hardware for Good Competition Semi-Finalist	2017
UCI Founder's Frenzy 3rd Bracket Team	2017
GPS-Biomed Elevator Pitch Competition 1st Place	2016
Paul Merage Business Competition Finalist	2016
Beall Hardware Competition 2nd Place	2016
Nominated: UCI Greatest Societal Impact (for Engineering Innovation) Award	2016
Nominated: UCI Most Dedicated Team Award	2016
Nominated: UCI Best IP Award	2016
UCI Best Female Innovator Award	2016
AGS Research Talk Competition Semi-Finalist	2015
UC Berkeley Biology Poster Symposium Honorable Mention	2013
Golden Key International Honors Society Member	2011

SCIENCE & TECHNOLOGY COMMUNICATION

Graduate Research Advocacy Day **April 2017**

Sacramento, CA

Represented the University of California, Irvine and advocated the importance of science funding at the State capitol. Advocated for California DREAMERS act as well as successfully advocated for bills to pass more funding for graduate students.

KUCI Get Notified! **April 2017**

Irvine, CA.

As a reward for winning the GPS-Biomed pitch competition, I spoke on the air with another student about entrepreneurship and communicating science effectively to reach broader audiences.

SoCal Robotics Conference Poster Presentation **June 2016**

San Diego, CA

Presented: "*Norman, S.L., Sarigul-Klijn, Y., & Reinkensmeyer, D.J. (2016, April). Design of a Wearable Robot to Enable Bimanual Manipulation after Stroke. Southern California Robotics Symposium 2016"*. Was chosen to pose for photos for the conference because of the work.

Course on Science Communication **January–May 2017**

University of California: Irvine. Irvine, California, USA

Took a ten-week program which focused on different methods to communicate science to lay audiences. We focused on key ways to improve public speaking and science advocacy.

1 Million Cups **June 2016**

University of California: Irvine. Irvine, California, USA

Presented on a novel robotic device for lower leg therapy and discussed commercialization to a crowd made up of medical professionals, business people, and investors.

Exercise Medicine and Sports Initiative **October 2015**

University of California: Irvine. Irvine, California, USA

Prepared a virtual reality demonstration with another graduate student and worked at a booth talking to people about robotics and science education.

PUBLICATIONS

Journal Articles

J1. Sarigul-Klijn, Y., Smith, B. W., & Reinkensmeyer, D. J. (2017). Design and experimental evaluation of yoked hand-clutching for a lever drive chair. *Assistive Technology, (in press).*

Conference Articles & Abstracts

1. Norman, S.L., Sarigul-Klijn, Y., & Reinkensmeyer, D.J. (2016, April). Design of a Wearable Robot to Enable Bimanual Manipulation after Stroke. *Southern California Robotics Symposium 2016*.

C2. Sarigul-Klijn, Y., Lobo-Prat, J., Smith, B. W., Thayer, S., Zondervan, D., Chan, V., Stoller, O., & Reinkensmeyer, D. J. (2017, July). There is plenty of room for motor learning at the bottom of the Fugl-Meyer: Acquisition of a novel bimanual wheelchair skill after chronic stroke using an unmasking technology. *In Rehabilitation Robotics (ICORR), 2017 International Conference on (pp. 50-55). IEEE.*

C3. Sarigul-Klijn, Yasemin Maria. "Gait Rehab Adaptive Machine: Design of GRAM, a Walking Linkage Powered Wheelchair for Lower Body Therapy and assistance" *ASME 2018 Design of Medical Devices Conference*. American Society of Mechanical Engineers, to be submitted.

C4. Sarigul-Klijn, Yasemin Maria, and Nesrin Sarigul-Klijn. "A Dynamic Response Model of Human Spine Under Superior-Inferior G Loads using a Voigt Viscoelastic Representation" *ASME 2013 Summer Bioengineering Conference*. American Society of Mechanical Engineers, 2013.

C5. Ozcan-Eksi, E & Feeley, Brian & Dang, Alan & Sarigul-Klijn, Yasemin & Deviren, Sibel. (2014). Poster 514 The Inter-Variability of Four Different Paraspinal Muscles' Measurement Techniques on Lumbar Spine MRI in Patients with Spinal Disorders. PM&R. 6. S366. 10.1016/j.pmrj.2014.08.882.

MENTORSHIP

Mentor - Undergraduate Students

Selina Eich **December 2016- March 2017**

ABSTRACT OF THE DISSERTATION

Development and Testing of a Novel Class of Device for Rehabilitation and Assistance: Exochairs

By

Yasemin Sarigul-Klijn

Doctor of Philosophy in Biomedical Engineering

University of California, Irvine, 2017

Professor David J. Reinkensmeyer, Chair

Stroke is one of the leading causes of disability in the United States. After a stroke, patients require both rehabilitation to reduce upper and/or lower body impairments, and assistance to help carry out daily living. The amount of conventional therapy performed is often limited due to cost and lack of intensity, therefore patients do not receive an effective dose to maximize gains.

Robotic training devices can promote greater dosage of therapy, but often are expensive and bulky, limiting their use to the clinic. In the home, wheelchairs provide low-cost mobility, but no means of therapy. Emerging hybrid assistive/rehabilitation devices like exoskeletons provide mobility with some clinical benefit but remain infeasible due to complexity and cost.

The premise of this dissertation is that simple, low-cost devices can be designed to promote both therapy and mobility, improving patient outcomes. This dissertation explored the design and experimental validation of three such devices, including a new class of hybrid assistive/rehabilitative device called an “exochair”. An exochair couples a rehabilitative exoskeleton to a wheelchair, with the goal of allowing a user’s therapeutic exercise drive around the chair, granting the user both assistance in daily living as well as therapy.

First, I report results from experiments that quantified the ability of young nonimpaired users to learn to drive a previously developed upper extremity exochair called LARA. The LARA chair is driven via levers with an arm support (i.e. by a simple exoskeleton) and is intended to help individuals with weak arms to self-propel and exercise the arms. I compared two designs of LARA, one that used a simple one-way bearing mechanism to provide propulsion (a technique that had been previously proposed), and a second that utilized a yoked hand-clutching scheme for driving (a novel technique suitable for people with unilateral arm/hand weakness such as occurs after stroke, but potentially more difficult to learn). Unexpectedly, I found that the participants learned to drive the hand-clutched device with similar learning time constants, and ultimately achieved similar over ground speeds, compared to the one-way bearing model. The hand-clutched approach increased the physiological effect of exercise, cognitive engagement, and maneuverability, showing its potential as a hybrid assistive/rehabilitation device for people with stroke.

In a follow-up study, I then quantified the ability of people with severe arm impairment after chronic stroke to learn to drive the hand clutch design. All participants increased their propulsion speed across five days of training. Unexpectedly, they showed a rate of motor learning that was two times higher than the average value reported for learning of a variety of motor tasks by unimpaired individuals, including the rate of learning of LARA by unimpaired users. They also improved on clinical arm movement scores by an amount comparable to previous robotic therapy device studies and showed high levels of motivation and self-ratings of competence that increased to high levels across training. Thus, the exochair unmasked a latent motor learning ability for individuals who have a severe arm impairment after stroke. This is significant because such arm impairment normally prevents such individuals from learning functional tasks with the impaired arm. People can learn to drive an exochair, it is rewarding, it assists in functional ambulation ability, and this learning process improves clinical arm movement scores.

Based on these experiments, I explored the application of classical machine learning algorithms to assist people with bilateral hand impairment in driving LARA. I showed the feasibility of an automated clutch system, comparing it with a hand switch approach. This approach could potentially relieve the burden of hand-clutching when using LARA for longer periods of time.

Next, I present results of a low-cost hand exoskeleton designed for assisting in hand extension post-stroke, with a demonstration of use by an unimpaired subject. This work explored an alternate way functionality could be increased for patients via wearable devices.

Finally, I extended the exochair concept to a hybrid assistive/rehabilitative device for leg movement called GRAM. My design concept for GRAM was that users could manipulate a linkage with their legs, mimicking the natural kinematic pattern of walking, in order to propel the wheelchair. I tested GRAM in a case study with an unimpaired individual, demonstrating that it was possible to learn to propel GRAM using this technique.

These results demonstrate the potential of exochairs such as LARA and GRAM to provide hybrid assistance/therapy in a simpler and less costly way than is currently possible with other technologies. Future research will study the long-term functional and therapeutic benefits of exochairs in larger patient population.

CHAPTER 1: INTRODUCTION

1.1 STROKE

A. Stroke Disability Facts

Roughly 6 million Americans currently live with stroke making it the leading cause of disability in the United States [1]. Strokes occur when there is an occlusion of blood to the brain, either via a blockage or a blood vessel hemorrhaging. The resulting impairments caused by stroke in the descending neural pathways lead to a wide range of disability in the upper and lower extremities that impede activities of daily living. For those over 65 years of age, over half of these individuals face long-term disability from these impairments [1], [2].

Post-injury, some spontaneous recovery naturally occurs via a process called neuroplasticity. This process, a massive reorganization of cortical and sub-cortical function, tends to be especially prominent in brain areas involved in motor control [3]. Neuroplasticity can be further enhanced via physical rehabilitation. As a result, early rehabilitation efforts are crucial for patients to regain the most independence possible and have been shown to be effective as a means to regain motor function in both subacute and chronic stroke patients [4], [5]. Typically, post-injury patients will undergo several months of physical therapy to achieve the maximum motor recovery possible.

However, conventional therapy has several limitations. Most prominently, regimens are limited by high cost, which prevent patients from receiving an effective dose of therapy [6], [7]. Indeed, the potential recoverable function by a patient during conventional therapy varies greatly. Task specificity and therapy intensity are shown to be the main factors in recovery following stroke [8]. It is not surprising then that the actual amount of therapy that patients receive, an average of just 32 repetitions per session for upper extremity, which falls far below what is believed to be needed for healing, is insufficient to provide best outcomes [9], [10].

B. Motor Learning and Stroke

Motor learning is a concept that will be widely discussed in this dissertation. Simply put, motor learning is a change in movement ability that occurs in response to experience. For example, some individual practices shooting baskets, and through enough practice learns the coordination to eventually score a point. Motor learning includes several complex processes like skill acquisition and motor adaptation [11], [12].

Motor learning has been widely studied in healthy subjects. For example, a pivotal paper by Newell et. al. found that motor learning tasks could be modeled by the power law of practice [13]. The relevant equation is below and is used several times throughout the dissertation (see Chapter 3, Chapter 4, and Chapter 6).

$$T = BN^{-\alpha} \tag{1.1}$$

Here T is the time to finish the task and N is the amount of trials, and B and α are constants. In particular, B is the baseline, i.e. the first trial's performance time and α is the learning rate [14].

The main result found by this study was that this law could be applied to motor tasks irrespective of the task or time scale of practice. From this paper, the mean rate of learning for healthy subjects learning a motor task tends to fall between 0.2-0.3 [13]. In the context of stroke therapy, the effects of stroke on motor learning have not been widely studied [11]. For example, one paper found that stroke subjects of varying severity had corresponding variety in their ability to learn and perform an elbow flexion task [15]. This suggests having a stroke may alter motor learning; however, this is an area that needs more research.

1.2 ROBOT-ASSISTED MOTOR REHABILITATION

A. Design Considerations for Robotic Therapy Devices

Rehabilitative robotic devices such as therapy exoskeletons are seen as attractive alternatives to conventional therapy. These devices can automate the task of rehabilitating patients. They also can relieve the immense physical strain involved in delivering therapy. Furthermore, robotic devices can provide more quantitative measures of patient performance, which grants insight into the neural mechanisms behind stroke recovery [6], [16], [17]. These devices are expensive and bulky, however, making their use outside of the clinic infeasible [17]–[19].

Certain design considerations must be taken to ensure robotic devices are effective in promoting patient engagement and neuroplasticity. In particular, robotic devices must avoid a type of energy minimization and disengagement that tends to occur when patients use therapy robots, a phenomenon called “slacking.” This “slacking” has negative consequences for rehabilitation because high levels of engagement are needed to incite motor learning. High engagement is also needed to improve other health areas, like the cardiovascular fitness of patients, and improvement in these areas also promotes improved therapy outcomes. [20]–[22]. As an example, a study compared two wrist treatments after stroke [23]. One system had subjects remain passive in a wrist robot that provided therapy, while the other had patients use a wrist orthosis that was driven by their own muscle activation. It was found spasticity was reduced in both groups because there was a reduction in muscle activation during both therapies. However, it was found that the EMG-controlled robot improved muscle coordination and activation in subjects as well because they were actively engaged in the control of the device. This shows the importance of designing robotic systems for therapy that prevent patients from “slacking”.

Another problem in robotic rehabilitation besides cost is providing patients with the appropriate motivation to do therapy. Coupled with the previously mentioned issue of ineffective dosage of therapy post-injury, there is a clear need for patients to have systems which deliver salient and copious therapy for optimal recovery [6], [24], [25]. Overall, then, considerations into device efficacy must be weighed to justify their use. Currently, research suggests robotic devices are generally equal to conventional therapy in motor outcomes though the potential of such systems cannot be understated.

B. Upper Extremity Robotic Technologies for the Arm

There exists a wide range of upper extremity robotic devices with varying levels of complexity. The first device introduced was the MIT-MANUS which is a two-degree-of-freedom (DOF) robot that grants free motion to the patient in the horizontal plane for the shoulder and elbow joints. It featured an impedance controller to assist users. Clinical studies with the MANUS comparing it to a control group that had limited exposure to the robot showed the group using the robot had improved strength in their elbow and shoulder that was retained 3 years post-trial. Another study with 20 chronic unilateral stroke patients comparing two modes of robotic therapy with the device found both groups experienced improved motor function, strength and Fugl-Meyer (FM) scores [6], [26]–[28]. Similarly, positive results have been reported for the 6 DOF robot MIME comparing neuro-developmental therapy to therapy with the robot [29], [30]. The 3 DOF GENTLE/S robot comprising of an overhead support system with a haptic arm also showed robot-mediated therapy to be beneficial [31]. Less sophisticated devices like the Bi-Manu-Track arm trainer and ARM-guide also showed similar promise in stroke patients. The Bi-Manu-Track arm trainer also reports the greatest increase in FM score for this type of device [32], [33].

C. Upper Extremity Robotic Technologies for the Hand

For finer motor control, particularly finger extension, which is a prominent movement impairment after stroke, there are a variety of devices which exist in various designs that confer differing benefits [34], [35]. Passive devices like HANDSOME utilize a four-bar mechanism with a spring to provide the correct assistive torques for patients to accomplish a physiologically natural extension trajectory. Purely passive devices like this confer the benefit of lower costs due to a lack of motors but give up the ability to precisely control changing torques in response to the environment and patient needs [34], [36].

Active hand robotic devices have a wider variety of force transmission systems. Some examples of these active systems are the HEXOSYS (Hand EXOskeleton SYStem) which drives two fingers connected to an under-actuated linkage via electric motor [37]. Wege et al. developed an example of a cable drive system. In this device, each joint is articulated via Bowden cable and powered by an electric motor, and a single motor is used to pull two cables per joint [38]. Traditional actuators such as DC motors provide a reliable way to articulate hand extension in stroke patients. However, care must be taken to ensure proper motors are selected with appropriate compliance for human-robot interaction. Issues like friction and backlash, as well as the added weight and cost of using these sorts of motors, are the primary drawbacks of such systems [34], [37], [38].

FIGURE 1.1: EXAMPLES OF UPPER EXTREMITY ROBOTIC TECHNOLOGIES. **TOP LEFT:** THE MIT-MANUS. **TOP RIGHT:** THE MIME ROBOT. **BOTTOM LEFT:** HANDSOME, A PASSIVE HAND EXOSKELETON. **BOTTOM RIGHT:** HEXOSYS, AN ACTIVE HAND EXOSKELETON.

D. Lower Extremity Robotic Technologies – Naturalistic Gait Task

There are a variety of robotic technologies for the lower extremity that attempt to emulate a natural gait task in patients, and these devices can be divided into overground and stationary systems. The AutoAmbulator and the Lokomat are two stationary systems which use robotic arms and an exoskeleton respectively to guide patients through a natural gait motion [39], [40]. A set of robots in this category developed by our lab include ARthuR and PAM. With ARthuR the leg motions are controlled by a moving coil, and PAM is an overground treadmill trainer which allows for 5 DOF of hip movement, the most in this class of device [22], [41]. The Mechanized Gait Trainer and LOPES are two more examples of devices in this class [42], [43].

A clinical result comparing the AutoAmbulator between robotic and conventional therapy found no significant difference in timed walking tests [40]. The Lokomat has been widely used in several studies, but clinical results from the Lokomat are similarly mixed. One study reported that a mixed approach with the Lokomat and general therapy yielded better neurological improvements and walking scores than conventional alone. A different study using the Lokomat vs. conventional therapy in subacute stroke patients found improvements in timed walking tests, but not functional walking itself. A later study found the Lokomat gave better motor scores and timed walking tests than conventional therapy. There was no difference in self-selected walking speed for a study observing chronic stroke patients for the Lokomat vs. traditional treadmill therapy, whereas another similar study actually found conventional therapy outperformed the Lokomat [16], [44]–[51]. Results from the LOPES device with chronic stroke found significant increases in joint range of motion, walking distance and speed [52].

Overground systems include the KineAssist, which is a mobile base to catch patients if they fall as they try to walk. Similar to the KineAssist are systems like the Walk Trainer. There are also wearable overground systems, like the Hybrid Assistive Leg (HAL), ReWalk, and BLEEX exoskeletons, which are meant for higher functioning individuals due to the less stable nature of their designs. For overground systems, results from the aforementioned overground exoskeletons are limited mostly to feasibility studies [40], [53]–[55].

E. Lower Extremity Robotic Technologies – Alternate Gait Task

Asides from more traditional stationary systems like the Lokomat described above, there are also devices in this class that utilize other alternate trajectories of gait-like patterns. For example, the Gait Trainer is another device which uses foot plates in a gait-like motion. While the task did not exactly match walking, the hemiparetic subjects studied still reported improvements in overground walking velocities and other measures [42]. Other devices in this footplate subclass of stationary trainers are the Haptic Walker, GaitMaster, and G-EO systems [6], [17], [18].

More prominent in the difference of task, there also exists split-treadmill devices, which allow stroke patients to experience different feedback to each side. These devices have been studied mostly in the context of altering gait patterns to assist in correcting asymmetric stride length after stroke. The rationale of these devices is a that learning a novel motor task can give the patient more ability in the general skill of learning a new gait pattern then a traditional approach focused just on straight-ahead walking like the Lokomat [56], [57]. Through this ability, the patient then may eventually adopt a completely healthy gait pattern which is more robust to real-world dynamic conditions [56].

Indeed, studies have noted the similarities between different types of leg exercises such as elliptical motions, compared to overground walking and noted the potential uses of these alternate gait trajectories. An observational study of how humans learn to walk noted that rarely do young humans walk in a straight path only, but rather perform a variety of movement [58], [59]. This collaborates with the idea alternate gait patterns may also be beneficial to patient recovery. As a specific example regarding split-belt training, one study using 13 chronic stroke subjects found a protocol which initially exaggerated step length, improved stride length asymmetry significantly throughout the study, and these differences approached significance at the 1 and 3 months follow-up [60]. Thus, positive clinical results from devices in this section evidence the potential usefulness of devices that use different gait patterns to more effectively correct or explore how patients recover post-stroke. See Figure 1.2 for examples of these types of devices.

FIGURE 1.2: LOWER EXTREMITY ROBOTIC TECHNOLOGIES. **FAR RIGHT:** THE LOKOMAT, A STATIONARY NATURAL GAIT ROBOTIC DEVICE. **MIDDLE RIGHT:** THE REWALK OVERGROUND NATURAL GAIT ROBOTIC DEVICE. **MIDDLE LEFT:** THE GAIT TRAINER, A STATIONARY ALTERNATE GAIT ROBOTIC DEVICE. **FAR LEFT:** A SPLIT-BELT TREADMILL, A STATIONARY ALTERNATE GAIT ROBOTIC DEVICE.

F. Summary of Current Work in Robotic Therapy Technologies

These positive clinical results in both upper and lower extremity serve to illustrate the tremendous potential of rehabilitative robotic devices for therapy. However, systems to be deployed at home have so far been limited to small wrist or hand exoskeletons [61], [62] likely due to the previously described tremendous cost and bulk associated with robotic therapy. Thus, there exists a need for more accessible devices. Lower extremity robotic devices utilize a mix of natural gait tasks and alternative tasks to assist recovery, and these alternative tasks could potentially promote greater transfer to real-world walking after stroke.

1.3 ASSISTIVE DEVICES - WHEELCHAIRS

In parallel to this, patients also require assistance in mobility while they recover. While there are some technologies for upper limb assistance like mobile arm supports, mechanical devices that counterbalance forces so patients can more easily perform tasks of daily living, development for upper limb has been limited [63]. In the United States alone, however, the wheelchair serves 3.3 million people in the effort to provide assistance for the lower limb [64]. While wheelchairs have evolved in the last several decades, with various new designs and sizes, 90% of manual wheelchair users still drive conventional push-rim chairs for mobility [65].

However, there are physiological downsides to prolonged usage of such vehicles. Research shows extended time in a wheelchair can negatively affect patients by decreasing blood circulation, increasing contractures, and generally compounding previous medical issues [66], [67]. Traditional push-rim wheelchair users also report increased shoulder problems due to the nature of the chair's propulsion which leads to higher forces in the shoulder joint [68], [69]. Efforts to relieve these issues have been undertaken by several groups and commercial ventures who have designed modified wheelchairs with lever or pedal transmissions. While studies show promise of some of these alternate transmissions in improving vehicle efficiency and lowering physical strain, therapeutic effects for stroke have not been widely studied [68], [70]–[72].

1.4 HYBRID DEVICES

To increase time spent in therapy, some attempts have been made to combine assistive and therapeutic devices in the form of "exercise wheelchairs," which include pedal, lever, and hand and hub crank chairs.

A. Upper Extremity Wheelchair-Based Devices

Hub crank chairs are used in racing wheelchairs and are less practical due to complicated steering and braking, but use a similar hand motion to the hand-rim pattern in traditional wheelchairs [70]. In non-wheelchair users, the efficiency of this model was studied, and gross mechanical efficacy was >3% higher than a traditional wheelchair [73]. Similar studies by other groups confirm the higher efficacy of this mode of wheelchair propulsion. Likewise, hand crank wheelchairs have been reported to also lower physical strain and improve mechanical efficiency. Studies indicate lever drives offer more efficient modes of transportation but at the cost of maneuverability and size [70]. Another design hurdle with lever chairs is that max speed is limited by the repetition of the push, and furthermore, it is difficult to take breaks in between repetitions [74].

Assist systems exist for more traditional manual wheelchairs designs, such as push-rim-activated power assists. With these systems, users push the push-rim and an electric motor on the wheel activates and assists with propulsion [75]. However, more advanced control systems or techniques to develop them, like machine learning, have been limited mostly to power wheelchairs with the purpose of providing mobility as opposed to increasing theraputic opportunity [76].

Under the lens of rehabilitation, few studies exist examining these alternate drive systems for stroke patients. Basic research on how stroke patients use traditional wheelchairs exists. These studies look at the strategies patients employ to drive chairs, how walking ability correlates to wheelchair propulsion, as well as how posture effects patient propulsion [77]–[79].

Of the few studies regarding alternate designs, typically these are done in chairs that utilize unilateral propulsion with the patient’s unaffected side, and these include altered push-rim models driven by the good hand and foot [74], [80]. A study found that a unilateral hand wheelchair granted users quicker efficiency in completing tasks of daily living [81]. Likewise, research has been done showing the energy expenditure was lower and comfort was higher with the same unilateral design in nonimpaired subjects [74]. These designs do not encourage use of the affected side, limiting their effectiveness as therapy tools.

FIGURE 1.3: A MODIFIED WHEELCHAIR DRIVEN BY ONE HAND AND ONE FOOT FOR STROKE PATIENTS, TAKEN AS IS FROM [82].

Our group proposed a solution to this problem with our lever drive chair called LARA (Lever-Actuated Rehabilitation and Ambulation), which requires coordinated bimanual activity to propel the chair. The therapeutic background of LARA was founded in a previous pilot study with a stationary version of the device called RAE. In this study, our group found that users with severe arm impairment (FM score = 21.4 points ± 8.8 SD out of 66) could move and synchronize the levers to the resonance of the device and that repetitive exercise with RAE had a therapeutic benefit [83], [84]. LARA was developed from RAE by incorporating one-way bearings into the levers. In a previous study, our group found that people with severe arm impairment could also drive LARA overground [85], [86]. However, this version of LARA had shortcomings, as it was unable to back up or turn in place, key maneuvers for daily wheelchair use in a rehabilitation unit.

As a solution, we developed a second version of LARA with a novel drive system called "yoked hand clutching". Users operate this system by gripping a single clutch handle with their strong hand, and this simultaneously actuates both clutches on each side of the chair. This attaches the push levers to their respective wheelchair wheel so by the correct timing of the clutch a person can drive the chair analogous to how one would operate a traditional push-rim wheelchair. By properly timing the phasing of the levers and the clutching by the good hand, users can use this system to turn in place and back up. It was unknown, however, what the baseline performance level of either the hand-clutched or one-way bearing versions of LARA was in healthy subjects in terms of overground speeds, exercise effect or motor learning. Likewise, it was unknown if the hand-clutched version of LARA could be operated by people with stroke, and what the associated performance was in terms of overground speeds, exercise effect, motor learning, and clinical outcome.

B. Lower Extremity Wheelchair-Based Devices

For lower body wheelchairs, there have been several groups who have prototyped various pedal models, which are the main approach to leg based wheelchair exercise [87], [88]. Some studies in this area include one group who investigated the physical cost index of a pedal chair for stroke patients [89]. Another group compared three knee-extension wheelchairs for stroke. This study found that a four-bar linkage model was preferred to hand-propelled models [89]. Another group investigated three unilaterally propelled wheelchairs for stroke patients and found their knee-propelled design functioned best [90]. Yet another group investigated a knee-extension controlled wheelchair and found their population of unimpaired female drivers could operate it successfully [91]. Some of these groups have also explored various control algorithms to assist in the operation of these pedal wheelchairs [92]–[94]. For example, one group used a regenerative braking system to assist users in their pedal wheelchair [95]. Commercially, there also exists a pedal wheelchair device called “Profhand” which utilizes a hand control system for added maneuverability [96].

Groups have also combined other technologies in rehabilitation with the concept of a pedal wheelchair. Recently, a group developed a pedal-driven wheelchair that utilized a leg exoskeleton to assist impaired individuals with propulsion [97]. Similarly, there have been several groups which have investigated the feasibility of pedal wheelchairs used in conjunction with functional electrical stimulation (FES) to stimulate the leg muscles and assist patients [98], [99]. A study by one group investigated a leg-propelled wheelchair that used FES to assist movement; this model requires less effort to drive than prior leg driven chairs but was too large to be feasible for daily use [100]. One study comparing 17 patients after stroke using a pedal wheelchair and the same chair with FES, found both conditions reduced spasticity, but the FES assisted device could reduce spasticity in even higher-level patients [101].

FIGURE 1.4: A COMMERCIAL PEDAL WHEELCHAIR FOR LEG EXERCISE IN STROKE PATIENTS [96].

C. Limitations with Wheelchair-Based Devices

While these first results in exercise wheelchairs are promising there are still issues to be addressed. Many designs as discussed above for upper and lower extremity were unilateral, and research shows that use of the impaired limb is critical to recovery. The focus on pedaling in lower-extremity wheelchair devices also leaves unresolved questions regarding how different types of leg exercise could promote patient outcomes in this context. Perhaps for example, a walking motion, where each leg is uncoupled from the other, could promote a better therapeutic outcome since it would disallow patients from using their strong side to pedal for their weak. More studies would need to be done to investigate the true therapeutic effect of these current devices. Therefore, it is critical to address the unique issues of stroke recovery with a tool designed for the task.

1.5 GOALS OF THE DISSERTATION

The development of accessible devices for therapy and mobility have great potential to improve patient outcomes and daily functionality for individuals living with stroke. In this dissertation, I explore possible solutions to mitigate the main limiting factors involved with robotic devices for stroke recovery and function – cost, complexity, motivation, and accessibility via the design of a novel class of device for wheelchairs, called an "exochair." Exochairs are therapeutic exoskeletons that couple to a wheelchair's propulsion via a drive mechanism. I evaluate a previously designed exochair LARA for stroke rehabilitation and detail the feasibility of a novel lower-body exochair named GRAM. I also discuss the design and testing of another low-cost hand exoskeleton called RHED for use in a home environment.

In Chapters 2 and 3, I validate the upper extremity exochair, LARA in unimpaired subjects as well as people with stroke. In Chapter 2, I demonstrate the differences in motor learning of a hand-clutched version of the LARA exochair over a simpler one-way bearing design and discuss potential applications for use in inpatient stroke rehabilitation. In Chapter 3, I use the hand yoked version of LARA and demonstrate feasibility in subjects with chronic stroke as well as demonstrate that there was, unexpectedly, an accelerated rate of motor learning in these subjects compared to non-impaired subjects. In Chapter 4, I explore the feasibility of several paradigms to promote drivability of LARA for patients with very high level of bilateral hand impairment using classic machine learning algorithms and compare on and offline efficacy to a simple switching system. In Chapter 5, I discuss the development of a low-cost hand exoskeleton called RHED for the home environment to facilitate hand extension after stroke. In Chapter 6, I discuss the theoretical development and design process of a lower-body exochair called GRAM, as well as present motor learning results from an unimpaired subject to assess feasibility. In Chapter 7, I finally conclude with future work and directions.

1.6 RESEARCH CONTRIBUTIONS

- Previous literature showed lever-drive wheelchairs lower physical strain but are less maneuverable than push-rim wheelchairs. This dissertation shows that a solution to this issue, the clutch LARA, is better for stroke therapy, and also shows the associated performance of two LARA models in healthy subjects.

- Previously, this dissertation showed healthy people could drive the clutch LARA, and previously it had been assumed people with severely paretic arms after stroke cannot learn novel skilled tasks with that arm. This dissertation then shows people with stroke could learn the novel task of driving the clutch LARA with both arms, with a doubled learning rate compared to healthy subjects.

- Previous literature had developed limited ways for people with hand impairment to propel wheelchairs. This dissertation shows a person could learn to manually propel a lever drive chair when aided by an automated clutching control system. Clutch state identification accuracy was 90% online, 85% offline.

- Previous work with LARA showed arm exochairs could provide mobility and exercise in healthy subjects; this dissertation shows that exochairs can use leg actuation to provide mobility and exercise.

CHAPTER 2: DESIGN AND EXPERIMENTAL EVALUATION OF YOKED HAND-CLUTCHING FOR A LEVER DRIVE CHAIR

Note: Chapter 2 has been published as the following and is presented here in an edited format: Sarigul-Klijn, Y., Smith, B. W., & Reinkensmeyer, D. J. (2017). Design and experimental evaluation of yoked hand-clutching for a lever drive chair. Assistive Technology, (in press).

This chapter builds on the previous work done on the LARA exochair by experimentally validating the performance of young healthy subjects to learn two versions of LARA, one that uses a one-way bearing drive mechanism and another that uses a novel yoked clutching system. In this Chapter, we study twenty-two unimpaired novice adults learning to navigate a figure eight track during six training sessions over two weeks. We find that participant mean speed improved roughly 60% for both chairs, with similar exponential improvement time constants (3 days) and final speeds. However, speed improvement mostly took place overnight rather than within the session for hand-clutching, and the physiological cost index was also about 40% higher. These results indicate that while hand-clutching is no more difficult to learn than a lever drive, it is reliant on overnight improvement. Also, its increased maneuverability comes with decreased efficiency. We discuss how the yoked clutch may be particularly well suited for individuals with stroke during inpatient rehabilitation. This work then forms the basis for the work done in Chapter 3.

2.1 INTRODUCTION

The goal of this study was to develop and test an alternative method for propelling a lever drive wheelchair, “yoked clutching”, which could potentially be used by people with severe impairment of one hand resulting from hemiparesis after stroke, among other potential users from among the over 3.3 million people in the U.S. who use a wheelchair [64]. This is important because, despite rapid evolution over the last 40 years in many aspects of wheelchair design, including weight, wheel camber, and adjustability [65], 90% of people who use a manual wheelchair still use the conventional technique of push-rim propulsion [70]. Push-rim propulsion remains difficult for people with a stroke and others who have lost hand function. Lever drive chairs use an alternate transmission mechanism that has the advantages of lowering physical strain and improving efficiency compared to push-rim propulsion [65], [68], [70], [71], [102]. They also have the advantage of not requiring grasp and release of a push-rim, so they can be used more easily by people with hand impairment.

Despite these advantages, lever drive wheelchairs remain unpopular for reasons including poor aesthetics, social acceptability, difficulty in transferring, asymmetric function, increased weight/difficulty with stowing for transport [65]. One main practical contributing factor to the lower popularity of lever chairs is decreased maneuverability, where for the rest of this work our working definition of maneuverability will refer to the ability to perform simple maneuvers useful for daily living in many living and working environments: turning in place, going forward, and going backward. As an illustration of this lack of maneuverability: some lever drive chairs can only turn about one wheel and are difficult to back up or turn in place. Other lever drives address these problems by allowing the user to shift between different modes, such as forward and reverse, by manipulating shifter knobs [67]. However, compared to a push-rim operated chair, such interfaces are more cumbersome. For a person with a stroke, operating such interfaces bilaterally, with a severely impaired hand, is not an option.

Here, we study a solution to the lever drive maneuverability problem in which the user actuates a clutch between the lever and wheel on each stroke with his or her hand, by manipulating a clutch handle. The result is something very similar in nature to conventional push-rim propulsion: it requires the user to use his or her hand to, in effect, "grip" and "release" the wheel on every stroke. However, instead of gripping and releasing the push-rim, the user squeezes and releases the clutch handle mounted on the lever that in turn grips and releases the wheel through the clutch. This approach has the advantage of allowing the arms and hands to be positioned in a more ergonomic position, which allows people with severe arm impairment to manually propel the wheelchair. For example, we found in a recent pilot study that individuals with severe hemiparesis after a stroke could bimanually propel a lever drive chair [85]. This bimanual engagement of both the affected and non-affected side could benefit people with hemiparesis since use of the affected side is key to promote rehabilitation and regain function. Further, as we show here, the user can operate the yoked clutch with only one hand, which is suitable for people with severely reduced function of one hand (but some motion of both arms), compared to grasping and releasing moving push rims with both hands.

This paper first explains the design concept of a hand-clutched wheelchair, including how the clutching can be implemented with a Bowden cable, an electric clutch, or a hydraulic clutch. In initial testing, we discovered that the hand-clutched chair could be maneuvered well when the clutches on both wheels were yoked to a single clutch lever. We provide a simple feasibility demonstration of this claim using video capture of an experienced user operating the hydraulic yoked clutched version in constrained office space.

Using a single clutch handle to actuate both clutch levers may be non-intuitive to some users. However, a "yoked" system such as this could provide benefit to people with hemiparesis over a traditional push-rim or individually clutched lever chair since it would allow users to engage both the impaired and unimpaired arms for maneuvering, but only require the less-impaired, ipsilesional hand for clutching, thereby promoting use and recovery. To determine if novice users have difficulty learning this technique, we measured the amount of time required for motor learning (quantified as improvement in speed over days of practice), while also estimating efficiency (defined in this paper as lower physiological cost index of using the device) for a hydraulic yoked clutch chair compared to a conventional one-way bearing lever drive.

2.2 METHODS

A. Design of a Hand-Clutched Wheelchair

In this section, we describe the design of the hand-clutched chair, which we developed from a lever drive chair that at first used one-way bearings, modifying the chair to include clutches that allow the user to join the levers to the wheels at will to propel the wheels. The initial lever drive chair was the "LARA" (Lever-Assisted-Resonant-Assistance) wheelchair that we developed previously [84], [85]. LARA has two identical levers mounted to the wheel hubs that are used to drive the wheels. The levers are also connected to the wheelchair frame with springs, which creates a resonant system that aids in energy efficiency and helps support and provide appropriate limits to the range of motion of the arms [84]. Mass added to the ends of the levers lowers the resonant frequency of the system to a range typical of forward-backward stroking in steady-state propulsion, about 1 Hz.

We initially designed LARA with two transmission modes, selectable with a mechanical lever mounted behind the wheel hub. The first mode is the one-way bearing mode, and it uses a one-way bearing to allow forward propulsion of each wheel with a forward lever stroke. The bearing disengages during a rearward lever stroke and during coasting. This also allows turning about one wheel, when braking is applied to that wheel. The second mode is the stationary mode, and it decouples the lever from wheel rotation, allowing the lever to rotate around the wheel axle independently of the wheel. This allows the user to exercise the arms by moving the levers back and forth without propelling the chair, a form of exercise which we have found to reduce arm movement impairment of individuals with severe hemiparesis after stroke [82], [83].

To implement hand-clutching, we placed the wheelchair transmission in stationary mode and used bicycle brake levers attached to the ends of the levers (Figure 2.1) to actuate clutches mounted on each side of the chair. This clutch was implemented using a bicycle brake caliper and brake disk. When the user squeezes one of the brake levers (about 29 N of force is required to activate the clutch in the hydraulic version of the clutch), the brake caliper, which is mounted to the wheelchair lever, grabs the bicycle brake disk, which is mounted to the wheelchair wheel. Thus, when the user squeezes the brake levers (which we will call "clutch handles" for the rest of this paper for consistency with their actual function), then she can propel the chair forward a small amount by simultaneously pushing forward on the levers, now mechanically coupled to the brake disk, thus turning the wheel. However, to keep forward momentum, the user must then release the clutch handle at the end of the stroke, and then move the lever backward, independently of the wheel, allowing a return stroke of the lever. The result is something very similar in nature to conventional push-rim propulsion during which a push-rim is alternatingly grasped and released as the arms move forward and backward.

We implemented three versions of the hand clutch. The first used the clutch levers and Bowden cables to actuate the clutch. The second replaced the clutch levers with electric rocker switches that are read by a microcontroller (Arduino), which then activates a servomotor (Hitec HS-7950TH

Ultra Torque Servo) that pulls on the Bowden cables. In this electronically-clutched version, very small finger forces trigger the switches and activate the clutches. This version also allowed us to use a single switch mounted on one lever to trigger both clutches at the same time, which we found also allows a good level of maneuverability in a constrained space, as we demonstrate below. In a third version, we replaced the Bowden cable bike brake system with a hydraulic bike brake system. This system is stiffer, because there is no cable stretch, and also allows one clutch lever to actuate both clutches when the sides are hydraulically joined through a hydraulic switch (Figure 2.1). In this paper, we study the hydraulic clutch version.

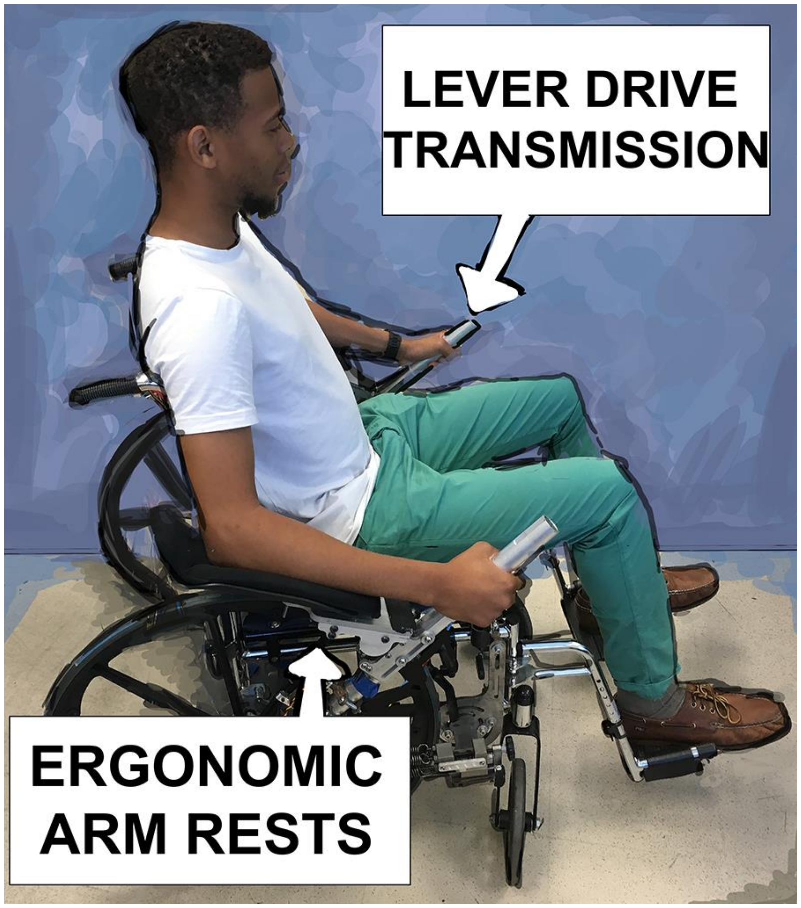

FIGURE 2.1: THE LARA WHEELCHAIR HAS TWO IDENTICAL LEVERS, FITTED WITH ARM SUPPORTS FOR THE USER TO REST THEIR ARM. THESE LEVERS ARE ATTACHED TO THE FRAME OF THE CHAIR VIA SPRINGS, AND TO THE CHAIR'S WHEEL VIA A ONE-WAY BEARING, OR A ONE-WAY CLUTCH, WITH EACH MODE SELECTABLE WITH A FLIP LEVER.

We discovered during initial testing that the chair could be driven and maneuvered well when both clutches were connected to one clutch lever or "yoked". Since we were interested in wheelchair design for individuals with stroke, who often have one paralyzed hand, we decided to study this "yoked clutch" mode. Given our target population's unique user needs, we neglected other modalities of wheelchairs (individually clutched lever chairs or traditional push-rim chairs) because yoked lever clutching only requires one hand for the more dexterous demand of clutching, and this could be done using the nonimpaired hand, while engaging both the impaired and unimpaired arms for propulsion. The potential for engaging the affected side to promote stroke recovery in this way promoted further study into the feasibility of the device as a way to provide both recovery and address maneuverability for people who have had a stroke and other disabilities.

B. Experimental Methods

22 young, male, unimpaired participants volunteered for this experiment, which was approved by the UC Irvine Institutional Review Board. In order to evaluate motor learning, peak achievable speeds, and energy expenditure of yoked clutching versus a traditional one-way bearing lever drive, a cohort of 11 participants (average age = 22.3 years ± 2.8 SD) first learned to drive LARA with the one-handed hydraulic hand clutch. In a follow-up experiment, a second cohort of 11 participants (average age = 21.5 years+/- 1.9 SD) learned to drive the same wheelchair, but this time in the one-way bearing mode. The weight of the hydraulic version of the LARA chair itself was 311 N and the counterweights weighed approximately 18 N each.

Participants in the hand clutch group were asked to drive around a 14-meter long figure eight track, defined by tape lines on the floor, and to complete 10 laps each day for six days. Participants became faster each day, reducing their mean practice time each day. To match practice times between this hand clutch group and the one-way bearing group, the one-way bearing group was instructed to complete an unspecified number of laps over a period of time equal to the average practice time calculated for the hand clutch group on the corresponding day of training.

Participants' lap times were recorded using a stopwatch application (Apple iPhone native app), with each new lap beginning when participants crossed the starting line. Heart rate was measured before and immediately after training for each group using a smartphone application ("Heart Rate" by Azumio for Apple iPhone). We verified the accuracy of the app in measuring heart rate with a Welch Allyn Propaq Cs hospital-grade pulse oximeter device. Over a test period of 5 minutes using three different test subjects in a resting condition (sitting in a chair), it was found that the mean difference between the app and monitor was 0.13 beats ± 0.7 SD. We also had the same subjects perform three rounds of light cardio exercise (jogging in place for thirty seconds), and in this condition, we found the mean difference between the app and the monitor was 0.67 beats ± 0.57 SD.

C. Data Analysis

Mean velocity for each lap (hereafter referred to as "velocity") was calculated by dividing lap times by the track distance. To quantify within-day improvement, the velocity for each participant's last lap of that day was subtracted from their first lap velocity. Between-day improvement, defined as performance improvements that accumulated without training between training sessions, was calculated by subtracting the velocity of each subject's first lap of the following day from the velocity of the last lap of the prior day.

The physiological cost index (PCI) of each drive type was calculated as:

$$PCI = \left(\frac{Ending\ HeartRate - Starting\ HeartRate}{Velocity}\right) \tag{2.1}$$

This equation for physiological cost index relates to a subject's efficiency and has been found to be a valid substitute for other measures of energy expenditure in submaximal (less than 85% max heart rate) exercise in young adults [102], [103]. In this experiment, maximal heart rates were not reported since measures were taken before and after exercise, so the heart rates for this measure are submaximal.

A mixed model ANOVA was used to compare the two groups for velocity, change in heart rate, and PCI. Day and group (i.e. hand clutch or one-way bearing) were considered to be fixed effects, with day a repeated measure. An interaction term between day and group was included. T tests were used for some post-hoc comparisons.

To find the time constant of speed increase for each drive type, participants' daily average speeds were fit using MATLAB's fit.m using the following exponential model:

$$v_k(t) = (v_{ss,k} - v_{0,k})\left(1 - e^{-\frac{t}{\tau}}\right) - v_{0,k} + \epsilon \tag{2.2}$$

Where $v_{0,k}$ is the average speed on day 1 and $v_{ss,k}$ is the asymptote of the speed curve, both for the kth participant, τ the learning time constant (assumed to be the same for all participants); t is the experiment day (with $t = 0$ on day 1), and ϵ is a normally distributed error term assumed to have uniform variance across participants and days of the experiment. Since each participant had a unique speed profile that caused large between-participant variance, each participant's daily speed was first normalized by subtracting that participant's mean speed across days, then dividing by their standard deviation of speed across days; both operations do not affect the time constant. The fit was then performed as a weighted least squares regression with this normalized data and a weighting that restored the assumption that ϵ was uniform across participants. The weighting for each data point for the kth participant was therefore equal to $w_k = SD_k^2$.

2.3 RESULTS

A. Example Level of Maneuverability of a Yoked Clutch Lever Drive With a Single Clutch Handle

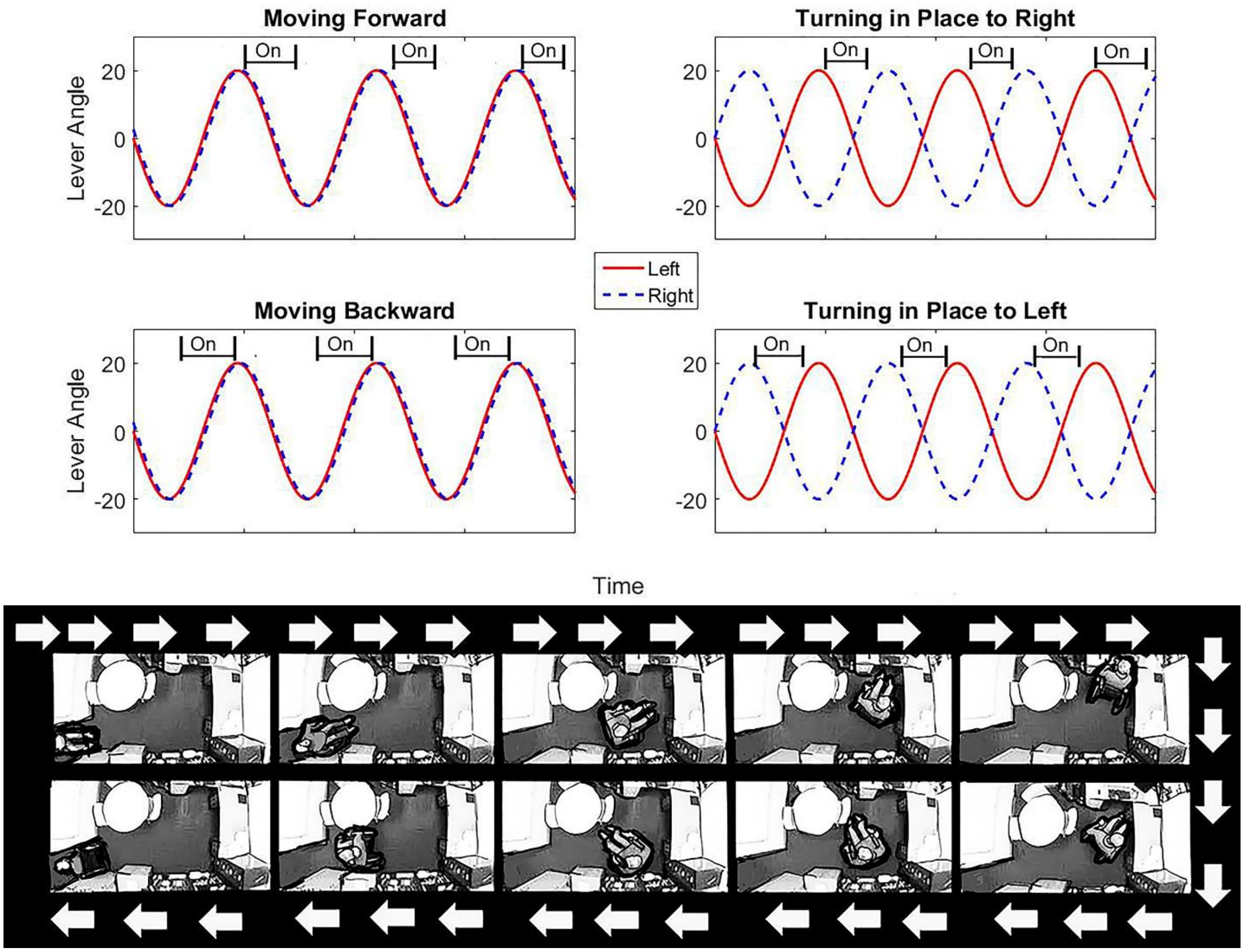

FIGURE 2.2: **Top:** Proper timing of clutching and lever phasing allows the users to move forward, turn, and back up. Moving the levers in phase and clutching during the forward strokes moves the chair forward. Moving the levers out of phase and clutching one lever during its forward stroke and the other during its backward stroke causes the chair to turn in place. **Bottom:** Example of an experienced user maneuvering LARA in and out of a crowded office space using a single clutch handle mounted on the right lever to actuate the clutches connecting the levers to both the left and right wheels.

Figure 2.2 shows an overhead view of a user maneuvering a chair into and out of an office using a yoked hand clutch with the hydraulically actuated clutch system. In this feasibility demonstration, the user controlled the two yoked clutches using his right hand in an environment which exemplifies one that is somewhat difficult to maneuver a wheelchair in. The user had practiced with the chair for about two hours in a variety of contexts and practiced moving in and out of the office three times before the run shown. By coordinating the gripping action of the right clutch handle with movement of the levers, the user successfully navigated the office, pulling up to the desk, backing up, turning in place, and moving through the doorway, without bumping objects. This simple example in an environment many wheelchair users may encounter, although not quantified, at least demonstrates that a moderately experienced user can achieve a useful level of maneuverability using clutches on each wheel yoked to a single clutch handle. Figure 2.2 explains more detail of how forward and backward propulsion and turning in place, can be achieved by varying the phasing of the lever motions with the action of the hand clutch.

B. Learning to Drive a Hand-Clutched Lever Drive

To quantify how difficult it is to learn to drive a yoked clutch lever drive, we compared two groups of trainees who had never driven a lever drive chair – one group learned to drive the yoked clutch transmission, and the other learned to drive the conventional, one-way bearing transmission. In both groups, all participants were able to learn to drive their version of the chair. On the first training day, it took the hand clutch group about five minutes on average to perform the required ten laps (Figure 2.3). As the hand clutch group became more proficient, they needed less time to complete the original ten laps (Figure 2.3). We matched the one-way bearing group practice times to the hand clutch group practice times, so the chair. On the first training day it took the hand clutch group about five minutes on average to perform the required ten laps (Figure 2.3). As the hand clutch group became more proficient, they needed less time to complete the original ten laps

(Figure 2.3). We matched the one-way bearing group practice times to the hand clutch group practice times, so the total amount of practice time decreased over days for both groups, providing a fair comparison.

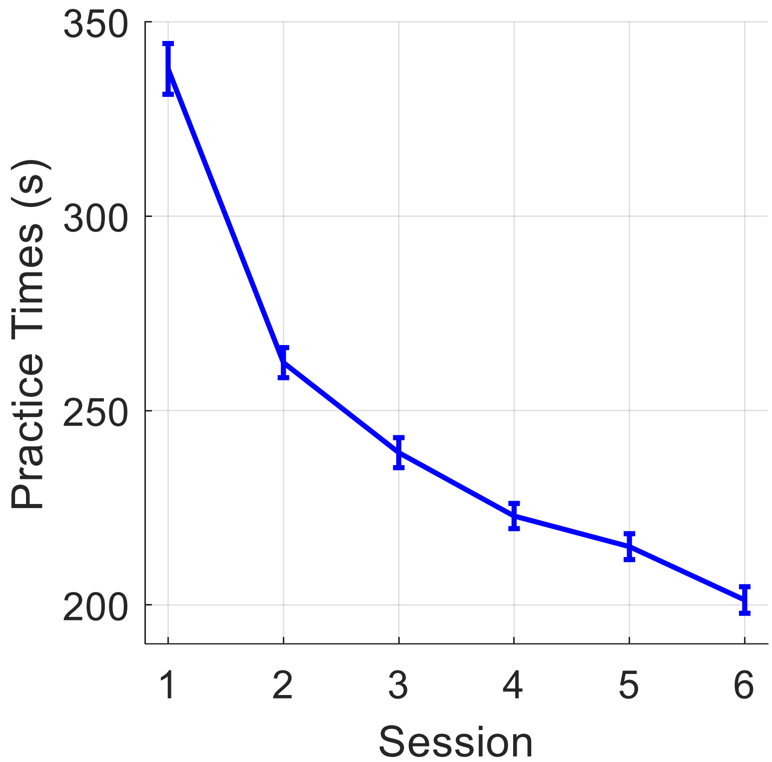

FIGURE 2.3: DAILY PRACTICE TIME. THE PRACTICE TIME FOR THE CONVENTIONAL TRANSMISSION GROUP WAS MATCHED TO THE AVERAGE PRACTICE TIME REQUIRED FOR THE YOKED CLUTCH GROUP TO COMPLETE 10 LAPS. ERROR BARS SHOW ± 1 STANDARD ERROR.

The mean speed achieved during training increased significantly for both groups across the six days of the study (Figure 2.4, mixed model ANOVA, $p<0.001$). Participants who used the hand clutch were 9.2% slower on the first day (Figure 2.4). By the last day, the participants who used the hand clutch achieved speeds that were 3.9% slower than the participants who used the one-way bearing, (Figure 2.4). The interaction term and the group term of the speed curves were not significantly different from each other (mixed model ANOVA, $p = 0.8, 0.5$).

We fit an exponential model (two degrees of freedom: time constant and offset) to the individual improvement curves to estimate the time constant of improvement for each subject. The one-way bearing group had a mean time constant of 3.23 days with a 95% CI of [2.38 5.10] and the hand-clutched group had a statistically comparable time constant of 3.18 days with a 95% CI of [2.30 5.18].

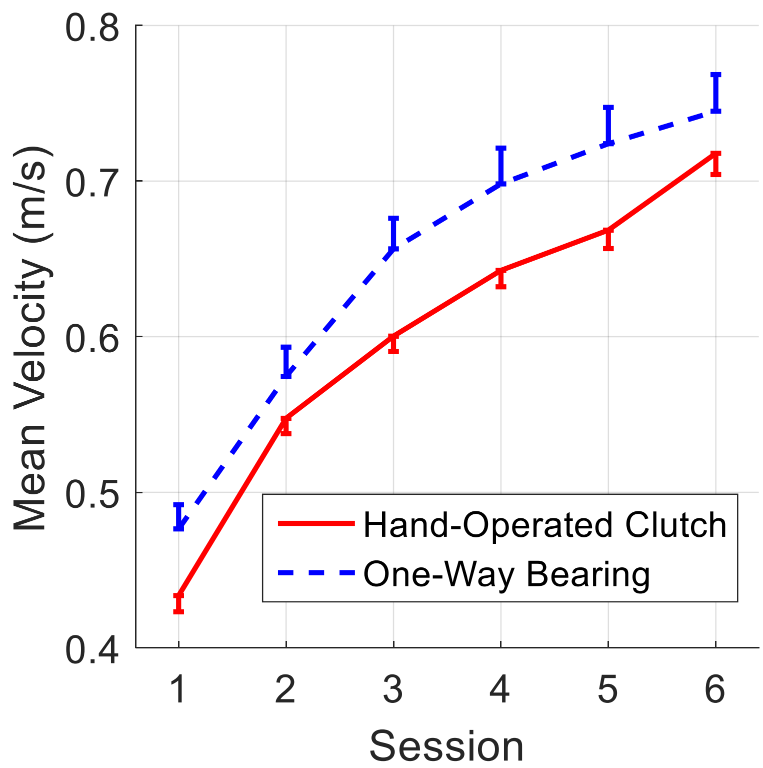

FIGURE 2.4: AVERAGE WHEELCHAIR SPEED. VELOCITY FOR BOTH GROUPS ACROSS TRAINING DAYS AS THEY COMPLETED LAPS ON THE FIGURE EIGHT TRACK. THE ONE-WAY BEARING GROUP STARTED FASTER AND ENDED WITH SLIGHTLY HIGHER VELOCITIES COMPARED TO THE HAND-CLUTCHED GROUP. THE DIFFERENCES IN VELOCITY WERE NOT STATISTICALLY SIGNIFICANT, HOWEVER, BOTH GROUPS SHOWED A SIGNIFICANT INCREASE IN VELOCITY OVER TIME. ERROR BARS SHOW ± 1 STANDARD ERROR.

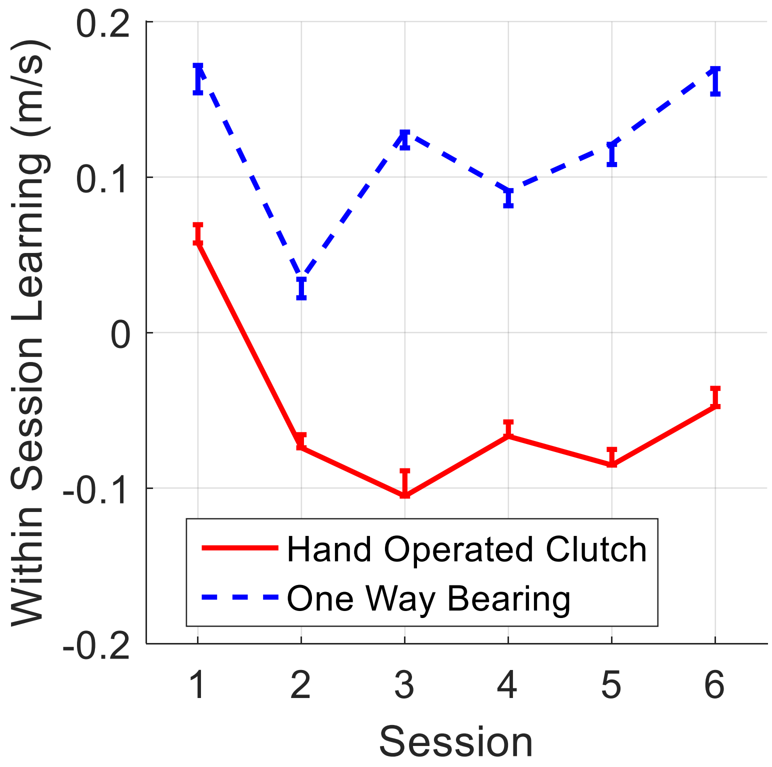

FIGURE 2.5A: WITHIN-SESSION VELOCITY CHANGE FOR EACH GROUP. THE ONE-WAY BEARING GROUP MADE IMPROVEMENTS WITHIN THE SESSIONS AND THE HAND CLUTCH GROUP DID NOT.

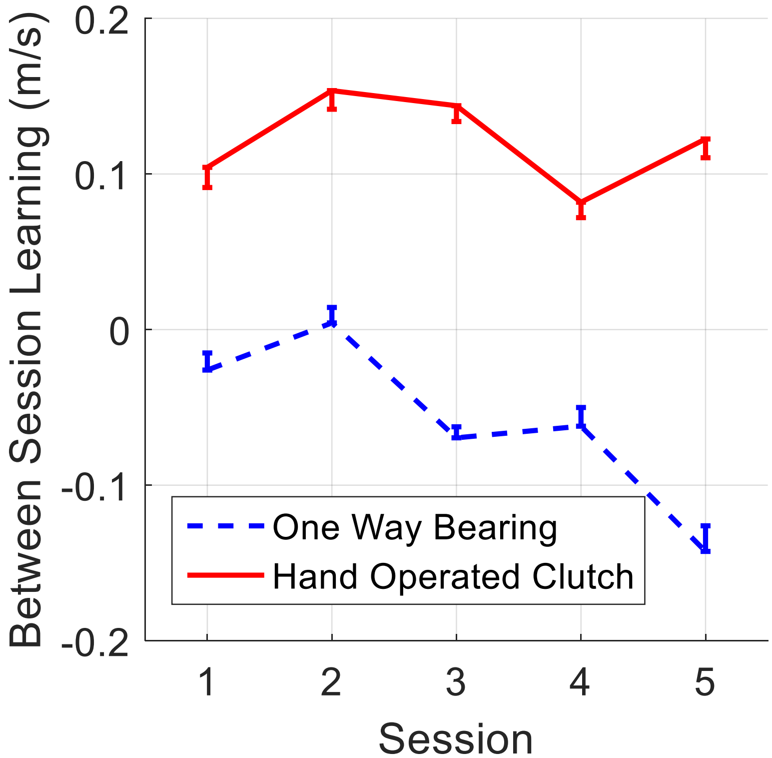

FIGURE 2.5B: BETWEEN-SESSION VELOCITY CHANGE OF EACH GROUP. THE HAND CLUTCH GROUP SHOWED BETWEEN-DAY LEARNING, WHEREAS THE ONE-WAY BEARING GROUP DID NOT. ERROR BARS SHOW ± 1 STANDARD ERROR.

C. Within-Day Versus Between-Day Improvement

We were interested in how much of the improvement took place within a training session, versus how much took place between-day. For each subject, we defined within-day improvement as the change in velocity from the first to last lap and between-day improvement as the change in velocity from the last lap of the previous day to the first lap of the next day. Positive values correspond to improvements in performance (Figure 2.5). A mixed model ANOVA analysis showed a trend toward significance for the between group effect ($p = 0.07$), suggesting that the one-way bearing group exhibited greater within-day improvement than the hand clutch group. There was no significant across-day effect ($p = 0.14$) or interaction between group and day ($p = 0.29$). The one-way bearing showed an increase in velocity within each session, while the hand clutch group showed a reduction in velocity.

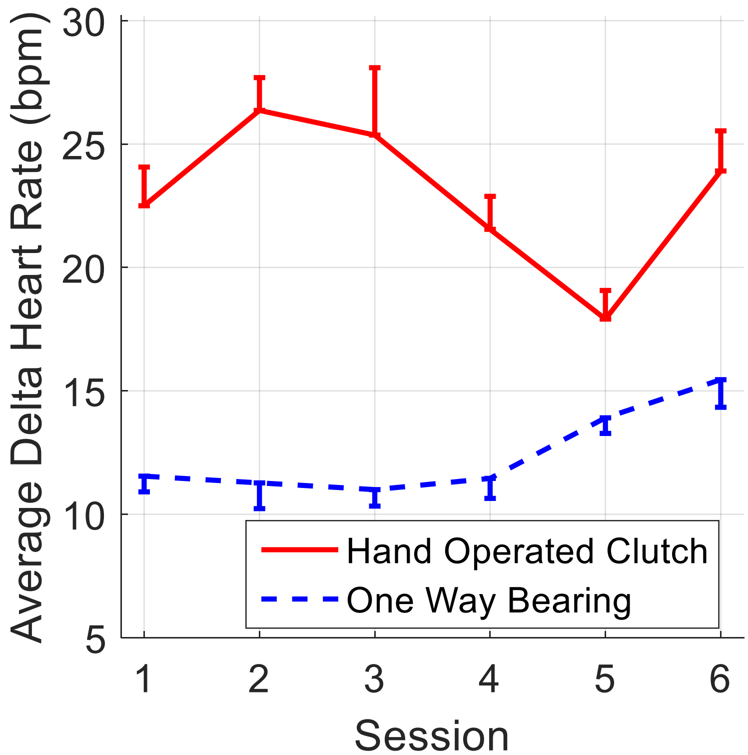

FIGURE 2.6: THE AVERAGE CHANGE IN HEART RATE PER DAY FOR EACH GROUP. THERE WAS A SIGNIFICANT INFLUENCE OF TIME ON THE CHANGE IN HEART RATE. THE TWO CURVES ARE SIGNIFICANTLY DIFFERENT AND HAVE DIFFERENT RATES OF CHANGE. ERROR BARS SHOW ± 1 STANDARD ERROR.

The same mixed model ANOVA analysis was performed for the between-day improvement and revealed a significant between group effect ($p = 0.04$), suggesting that the hand clutch group exhibited greater between-day improvement than the one-way bearing group. Again there was no significant across-day effect ($p = 0.6$) or interaction between day and group ($p = 0.27$). The hand-operated clutch group began each day with a greater velocity than they finished the previous day with, consistent with between-day improvement. The one-way bearing, began each day with a lower velocity than they finished the previous day with.

C. Changes in Heartrate and Physiological Cost Index

Training with both devices caused significant increases in heart rate, measured by subtracting heartrate measured before practice from that measured immediately after practice (Figure 2.6, mixed model ANOVA, p=0.03). That is, the change in heart rate was significantly different for the day term. There was also a significant difference for the group term (p=0.007) and the interaction term (p=0.03). By the last day subjects' heart rates increased with the brief training bout using the hand clutch by 23.9 beats per minute (bpm) ± SD 12.8 bpm, versus 15.5 bpm ± SD 7.1 bpm for the one-way bearing (Figure 6).

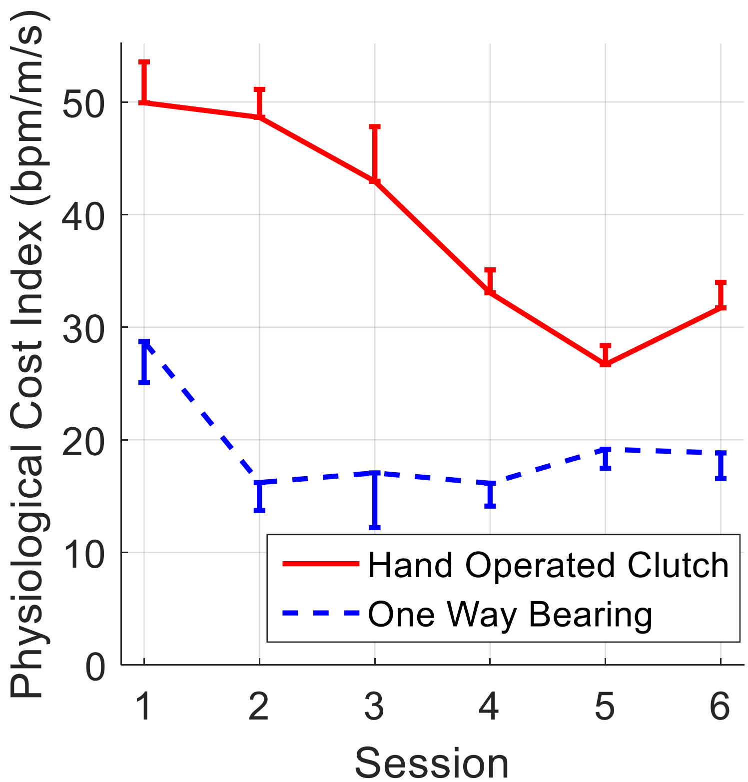

FIGURE 2.7: PCI FOR EACH GROUP. BOTH GROUPS HAD SIGNIFICANT REDUCTIONS IN PCI, AND PCI FOR THE ONE-WAY BEARING GROUP WAS SIGNIFICANTLY LESS. ERROR BARS SHOW ± 1 STANDARD ERROR.

PCI significantly decreased over time for both groups (Figure 2.7, mixed model ANOVA, $p = 0.005$, day term). On the first day, the hand-operated clutch group had a PCI that was 49.9 bpm/m/s ±36.3 SD compared to the one-way bearing group which reported a PCI of 28.7 bpm/m/s ± 27.8 SD. On the last day, the hand-operated clutch group had a PCI of 31.7 bpm/m/s ± 25.0 SD compared to the one-way bearing group which had a PCI of 18.8 bpm/m/s ± 9.9 SD. Also, there was a significant difference for the group term ($p=0.02$) and the interaction term ($p < 0.001$).

2.4 DISCUSSION

We studied use of a yoked hand clutch to propel a lever drive chair. As we demonstrated, yoked clutching allows an experienced user to maneuver a lever drive chair in and out of a constrained office by spinning in place, backing up, and moving forward. The key to this level of maneuverability is to vary the phasing of the levers with the timing of actuating the clutch lever, as detailed in Figure 2.2. We then studied if this propulsion technique is difficult to learn compared to conventional lever drive propulsion. The main findings were that 1) the one-way bearing and hand clutch group had overground speeds that were not significantly different from one another, with comparable time constants of improvement; but also that users 2) were less energetically efficient (i.e. a higher PCI metric despite slower speeds) when using the hand clutch device and 3) manifested a different form of speed improvement, between-day improvement for the hand clutch group and within-day improvement for the one-way bearing group.

A. Speed and Efficiency of Yoked Hand-Clutching

Yoked hand clutching resulted in noticeably slower speeds compared to the conventional lever drive, but users were less than 5% slower at the end of training, and this speed difference was not statistically significant between groups. Thus, the increased types of maneuverability (turning in place and backing up) possible with yoked hand clutching did not cause a large speed loss.

On the other hand, yoked hand clutching was associated with a greater increase in heart rate during training, and a significantly larger Physiological Cost Index (about 40% greater). This difference in PCI between groups decreased as training progressed, suggesting that part of the increased physiological cost of hand-clutching arose because of inefficiencies in coordinating the hand grasping and arm movements, especially early in learning. Another factor is that actuating the hydraulic clutches we used here required significant hand force generation. Actuating the clutch on each stroke could lead to grip fatigue, which could explain the difference in PCI between groups. It may be that using an electrically actuated clutch, with an electric rocker switch, for example, that requires only very small finger forces may improve efficiency. We also predict that the inefficiency of hand-clutching will be a function of the degree to which coordinated action of the hand and arm levering motion has been learned, and should decrease with extended learning or perhaps with simpler track configurations (such as a straight versus figure eight track).

B. Motor Learning of Hand-Clutching

Both groups increased their speed by over 60% on average over the six days of training, with only a few minutes of training each day. This indicates that hand-clutching, as well as lever drive propulsion in general, is a motor skill that requires time to master. When considering the additional coordination required to operate the hand clutch, it was somewhat surprising that the hand clutch group had no greater difficulty improving their speed than did the one-way bearing group. The velocity of both groups increased in an exponential manner with statistically comparable time constants.

However, we observed a different of pattern of within-day and between-day performance improvements for the two groups, with the yoked hand clutch not improving within-day, but rather between days, and vice-versa for the lever drive. One possible explanation is that the yoked hand clutch group experienced more hand fatigue during practice, due to repeatedly having to actuate the clutch. If so, this may have limited their speed improvements within-day. The next day, they would have recovered from the hand fatigue, and this may have boosted the between-day improvement. This possibility could be confirmed or eliminated by future research using an electronically actuated clutch that requires only very small hand forces. Another possible explanation is that hand clutch group experienced greater between-day improvement because of the higher complexity involved in coordinated the hand-clutching while pushing the levers. Indeed, learning motor skills of greater complexity benefits from having subjects sleep between sessions, presumably to consolidate learning [104]. Either way, these results suggest that a training regimen for learning hand-clutching should include training sessions on multiple days to allow between-day consolidation.

C. Study Limitations, Future Research, and a Key Application

Neither training group reached an asymptote in speed. Future studies should examine extended training periods. These extended training periods could also give insight into how grip fatigue could be a limiting factor in performance for the hand clutch group, and whether extended practice could improve hand strength and thus reduce fatigue, even to the point where the clutch users could be faster than the one-way bearing users. Another limitation of this work is the small sample size; with more participants, it is possible the slightly lower velocity of hand-clutching could become statistically significant. The course chosen for participants to navigate in this study was chosen so that they would alternate between forward propulsion and turning in both directions and because it was "doable" using both the one-way bearing and yoked hand clutching configurations of the wheelchair. However, course design may have influenced the results. Future studies should be longer or differently configured tracks then to assess these effects. Additionally, we would have to investigate these same metrics against existing one-arm lever drive systems to make a more comprehensive comparison, although such drives do not exercise both arms, which is a primary goal of this approach, applications in stroke rehabilitation.

Another limitation of this study is that we did not compare hand-clutching to a traditional push-rim, thus leaving unanswered the question of whether the efficiency gains associated with a lever drive compared to a push-rim chair are negated by the addition of hand-clutching. Further, the version of the lever drive used here incorporated counterweights whose added mass could affect efficiency [84]. Furthermore, a needed direction for future research is to quantify the level of maneuverability of a yoked clutch chair versus these other types of wheelchairs. Some lever drives use shifting to spin and back up, and it would be enlightening to understand if yoked clutching can provide better performance.

Finally, the use of heart rate as a measure of exertion has known limitations, and the study could be repeated while continually measuring heart rate during training and/or by using V02 measurements. A baseline fitness test between groups should also be administered in future works to eliminate the variable of individual's fitness levels affecting the heart rate measure during the experiment. Although we used a random population of college-aged students, in the future participant biometrics such as height and weight should be noted as well, since the driver size could affect the efficiency of the chair. Indeed, using a chest-belt heart rate monitor would likely help future work as an iPhone app has limitations, such as not providing real-time measurement of heart rate.

A key direction for future research is to examine motor learning of hand-clutching by individuals with arm impairment after a stroke. Indeed, our research group became interested in lever drive wheelchairs when we found that people with severe arm impairment after stroke could bimanually propel a lever drive chair [85]. This is significant because people with severe arm hemiparesis after stroke spend much of their day in wheelchairs during inpatient rehabilitation. They are typically taught to use their "good" arm and leg together, in order to move about in the wheelchair. This presumably contributes to further disuse atrophy of the hemiparetic upper extremity. Further, it has been suggested that individuals with a stroke do not exercise their arm enough for optimal recovery within current rehabilitation practice [105].

In pilot clinical and home-based testing, we found that repeated use of the hemiparetic arm to move a lever of a lever drive chair (while the chair remains stationary) reduces arm movement impairment; that is, levering motions are a form of therapeutic arm exercise [82], [83]. Thus, we hypothesize that use of a lever drive chair during inpatient rehabilitation could greatly increase the amount of arm movement exercise early after stroke. In a study currently underway, we are finding evidence that people with stroke can learn to operate the hand clutch and maneuver the chair, demonstrating the feasibility of this approach. Grip force strength is near normal for the ipsilesional hand for most people with stroke, so they typically have adequate hand strength to squeeze the yoked clutch with their 'good hand'. We envision the patient using a lever drive chair to ambulate around the rehabilitation unit, requiring many movements with the hemiparetic arm. In this context, maneuverability, including the ability to back up and get into and out of tight rooms, is important. Yoked hand clutching potentially provides this maneuverability for people with one severely impaired hand.

Within this stroke rehabilitation context, the decrease in efficiency of propulsion may actually be desirable, since cardiopulmonary exercise promotes motor recovery after stroke [106]. The hand-clutched group had an average increase in heart rate of 23 bpm +/- 3 SD, which put their heart rate at the lower end of their target heart zone range for cardiovascular exercise. Further, the fact that learning hand-clutching is a complex motor skill is attractive from the perspective of wanting to challenge patients with substantial motor learning tasks during rehabilitation [107]. We speculate that yoked hand clutching could someday play an important role in both wheelchair mobility and upper extremity rehabilitation of individuals recovering from a stroke.

CHAPTER 3: THERE IS PLENTY OF ROOM FOR MOTOR LEARNING AT THE BOTTOM OF THE FUGL-MEYER: ACQUISITION OF A NOVEL BIMANUAL WHEELCHAIR SKILL AFTER CHRONIC STROKE USING AN UNMASKING TECHNOLOGY

Note: Chapter 3 has been published as the following and is presented here in an edited format: Sarigul-Klijn, Y., Lobo-Prat, J., Smith, B. W., Thayer, S., Zondervan, D., Chan, V., Stoller, O., & Reinkensmeyer, D. J. (2017, July). There is plenty of room for motor learning at the bottom of the Fugl-Meyer: Acquisition of a novel bimanual wheelchair skill after chronic stroke using an unmasking technology. In Rehabilitation Robotics (ICORR), 2017 International Conference on (pp. 50-55). IEEE.

This chapter builds on the previous chapter by testing the potential of the yoked hand clutch version of LARA by validating its performance in a group of people with chronic stroke (n = 5, upper extremity Fugl-Meyer scores: 31,30,26,22,8). In this study, over six daily training sessions, each involving about 134 training movements with their "useless" arm, the users gradually achieved a 3-fold increase in wheelchair speed on average, with a 4-6 fold increase for three of the participants. They did this by learning a bimanual skill: pushing the levers with both arms while activating the yoked clutches at the right time with their ipsilesional (i.e. "good") hand to propel the wheelchair forward. They perceived the task as highly motivating and useful. The speed improvements exceeded a 1.5-factor improvement observed when young, unimpaired users learned to propel the chair. The learning rate also exceeded a sample of learning rates from a variety of classic learning studies. These results suggest that appropriately designed assistive technologies (or "unmasking technologies - UTs") can unleash a powerful, latent ability for motor learning even for severely paretic arms, which in the context of stroke have often been assumed unable to learn novel, skilled behaviors that incorporate use of that arm. While UTs may not reduce clinical impairment, they may facilitate large improvements in a specific functional ability. This chapter then serves as a validation of the potential of the yoked hand version of LARA for the purpose of stroke rehabilitation, as seen in Chapter 2, and provides credence to the exochair concept as being helpful for stroke recovery.

3.1 INTRODUCTION

Stroke is a leading cause of disability in the United States with nearly 800,000 people experiencing a stroke each year [108]. Approximately half of people who have a stroke end up with severe arm impairment [108]. Arm impairment can be reduced with intensive rehabilitation, such as what is made possible with robotic therapy devices, but the average improvement is small [109]. Further, arm movement recovery, measured as a change in the upper extremity Fugl-Meyer (UEFM) score over time, follows a well-defined trajectory for most patients, which is predictable from early EUFM score [110], [111]. The small improvements in EUFM score observed with intensive therapy coupled with the predictable progression of this score suggests limited use-dependent motor plasticity after stroke [15] – that is, a relative inability to alter limb impairment fate through training.

This raises the question, "How much motor learning is possible with a severely impaired limb?" Here, we studied this question using a novel lever-drive wheelchair called "LARA" or "Lever-Actuated Rehabilitation and Ambulation," that requires coordinated bimanual activity to be propelled. In a previous pilot study with a stationary version of LARA, a device called RAE, we found that users with severe arm impairment (FM score = 21.4 points ± 8.8 SD out of 66) could move the levers and synchronize to the resonance of the device, created by springs attached to the levers, and that repeatedly moving the levers had a therapeutic benefit [83], [112]. We then made RAE into LARA by incorporating one-way bearings into the levers and found that people with severe arm impairment could also drive LARA overground [84], [113]. This is important clinically because it means that stroke inpatients could potentially bimanually propel their wheelchairs, greatly increasing their arm activity beyond the low amounts that are typical during routine therapy [105], and thus potentially enhancing use-dependent plasticity [114]. With this clinical application in view, a practical problem with the first version of LARA was that it had limited maneuverability because the user could not back up or spin in place, which would make it difficult to use for transportation in a rehabilitation unit. Thus, for the present study, we developed a second version of LARA with a novel drive system we call “yoked hand clutching” [115] (Figure 3.1). Yoked hand clutching requires users to grip a single clutch handle with their ipsilesional (i.e. “good”) hand that simultaneously actuates both clutches: one on the left and one on the right, which attach the push levers to their respective wheelchair wheel. Somewhat counterintuitively, users can use this system to turn in place and back up by timing the pumping motion of the two levers with clutching by the good hand [115]. Learning to time the bimanual pumping and unimanual clutching is a motor skill that requires practice. Here, we studied how long it took people with severe arm impairment to learn this skill and thereby improve their speed of overground travel around a figure eight course using LARA

Table 3.1 Participants' Demographic and Clinical Information

FM Score	Age	Gender	Injury side	Preferred Arm	Type of Stroke	Days since stroke
31	62	M	R	R	H	439
26	53	M	R	R	I	469
22	59	M	L	R	H	1297
30	65	F	L	R	I	551
8	69	M	L	R	I	367

M: male, F: female, R: right, L: left, H: hemorrhagic, I: ischemic

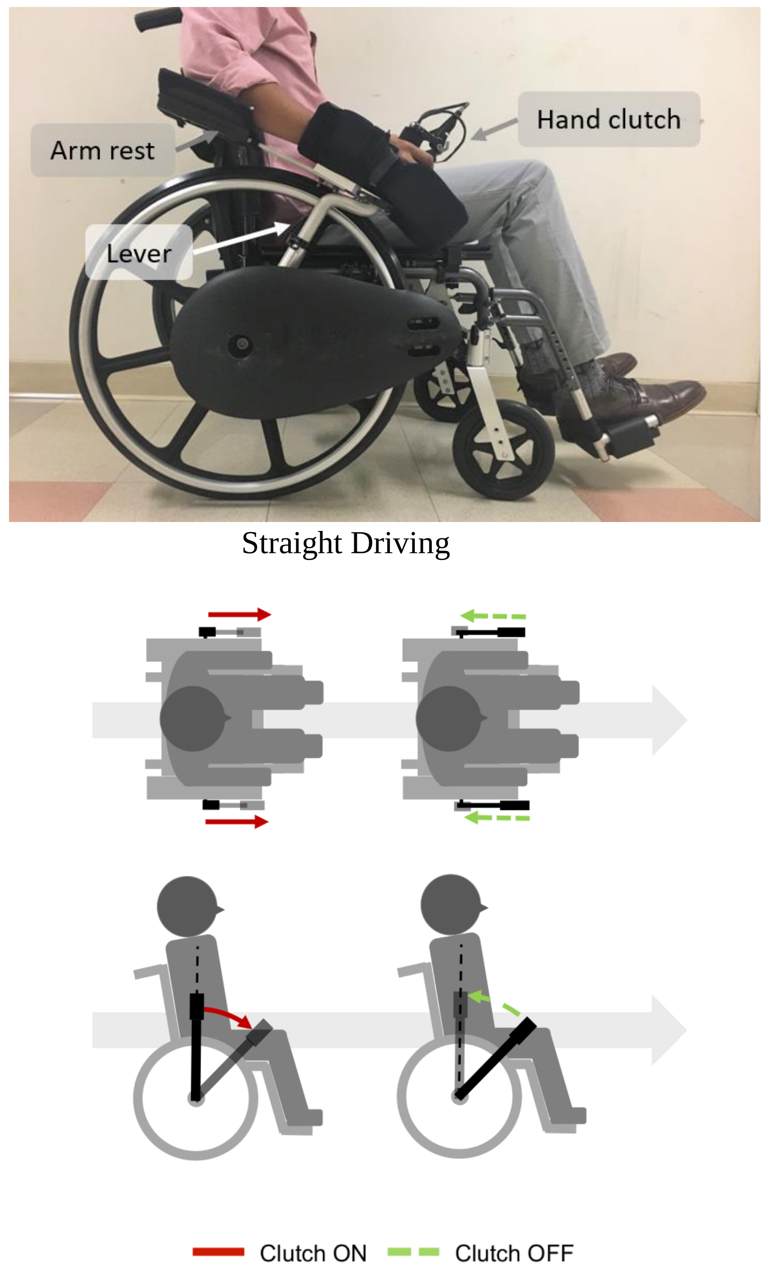

FIGURE 3.1: The Yoked Hand-Clutch version of the LARA wheelchair. **Top:** When the user is not squeezing the hand clutch, the levers rotate freely around the axels without moving the wheels. When the user squeezes the hand clutch, both levers become attached to each wheel (i.e. the single "yoked" hand clutch activates both of the clutches between the levers and wheels on both sides), so moving the levers now turns the wheels. **Bottom:** Schematic drawing showing how to use the clutch to drive the LARA wheelchair in a straight line. The user activates the clutch (ON, red) when pushing the levers forward and releases it (OFF, green) when pulling the levers back. The user can turn or back up by changing the phasing of the single hand clutch relative the two lever motions.

3.2 METHODS

A. Experimental Protocol

Five volunteers (Table 3.1), four males and one female, (mean age = 61.6 years ± 6.1 SD) with chronic stroke (1-3.6 years post-stroke) and severe arm impairment (mean EUFM score = 23.4 points ± 9.3 SD) provided informed consent to participate in this experiment, which was approved by the UC Irvine Institutional Review Board.

In this motor learning experiment, we were interested in skill improvement, measured as driving speed, as the primary outcome, but we also recorded clinical assessments as secondary measures as described below. Participants navigated a figure eight track with a 14-m long path designated with masking tape on the floor. This was an identical protocol to a previous study with younger, unimpaired participants [115]. To match practice times with that previous study (in which participants completed a fixed number of laps each session rather than a fixed training duration like this study), participants practiced for the following number of minutes each session for six training days spaced over two weeks: Session 1: 5 min 37 sec; Session 2: 4 min 22 sec; Session 3: 4 min 1 sec, Session 4: 3 min 42 sec; Session 5: 3 min 35 sec; Session 6: 3 min 21 sec. This reduction in practice reflected the previous study's participants increase in speed.

Participants' lap times were recorded with a stopwatch, with each new lap beginning when participants crossed tick marks on the track, which were spaced to be a quarter of the total track distance. A chest heart rate belt (Blu Beets Bluetooth Wireless Heartrate Monitor) was worn during the study to measure resting and peak heart rates. Resting heart rate was taken before the experiment began just after the participant had settled in the chair, and peak heart rate was measured while driving through the course. Gyroscopes (MPU 9250) attached to each lever recorded data at 5 Hz on a microcontroller (Arduino M0 board with AnyCubic Data Logging

Shield).

The UEFM and Box and Block Tests were evaluated by an experienced physical therapist before Sessions 1 and 6. Motivation was evaluated using the Intrinsic Motivation Inventory (IMI) after each session [116]. Limb spasticity was evaluated for the shoulder, elbow, wrist, and hand by the Modified Ashworth (MAS) before and after each session and a mean score was reported as the average of these four. Before and after each session pain was evaluated with the Visual Analog of Pain Scale (VAPS). Use and ease of use of the impaired limb outside of training was evaluated during Session 1 and 6 by self-reported Motor Activity Logs (MAL).

B. Data Analysis

The average speed was determined by dividing the number of laps completed by the amount of time required to complete them. The gyroscope data was analyzed in terms of peak lever speed on each pump, instantaneous pumping frequency, and instantaneous pumping synchrony. To identify lever pumps, the MATLAB function “findpeaks” was used with the threshold option of “MinPeakProminence”. Threshold levels were tuned manually for each subject. The peak lever speeds were found as the peak value identified by “findpeaks”. The instantaneous pumping frequency was calculated as the inverse of the time difference between peaks. The pumping synchrony, a measure of bimanual coordination advantageous for forward propulsion, was determined by performing a zero-lag cross-correlation between the unimpaired and impaired side gyroscope data with a window of 10 samples centered at each peak location of the unimpaired side gyroscope data.

Due to an error with the memory card, all gyroscope data was lost for Sessions 1 and 2, and two subjects did not have any gyroscope data recorded on Session 6, so for these participants, their data from Session 5 was used for the Session 6 instead.

To compare learning rates with those from a sample of previously published learning experiments, MATLAB's "fit" function was used to fit power curves (curve fit option 'power1') to the mean data of the five subjects over the six sessions with:

$$T = BN^{-\alpha} \tag{3.1}$$

where T is the time to finish the task and N is the amount of trials, and B and α are constants. In particular, B is the baseline, i.e. the first trial's performance time and α is the learning rate [14]. This same equation was also fit to a previously collected dataset acquired from young, unimpaired adults (mean age = 22.3 ± 2.8 years, N = 11) who drove the same version of LARA with the same experimental protocol [115]. See Chapter 1 section 1.1 for a discussion on motor learning. Statistical comparisons were carried out with the Friedman's test, Wilcoxon Rank Sum (WRKS) test and Wilcoxon Sign Rank (WSR) test.

3.3 RESULTS

A. Learning to Drive the Yoked Clutch Lever Drive Wheelchair

The participants significantly increased their mean speed over the six training sessions (Figure 3.2, Friedman's test, $p < 0.001$). The average increase in speed was 305 % ± 197 SD, with three participants achieving an increase over 400% and two around 200%. The learning rate, estimated as the exponent of the power curve fit, was $\alpha = 0.62$ for the people with a stroke. This rate was twice that of the mean learning rate reported for a sample of classic learning experiments ($\alpha = 0.28 \pm 0.16$ SD; Figure 3.3) [13]. The young unimpaired subjects who learned to drive LARA had a learning rate similar to that found in the classic experiments ($\alpha = 0.29$; Figure 3.3).

B. Features of the Lever Movement

From Session 3 to 6 (the two sessions for which we obtained gyroscopic data from both levers) we found an increase of about 50% in wheelchair speed that neared significance (WSR test, $p = 0.06$). We plotted histograms of peak pump speed, pump frequency, and pump synchrony from this data, in order to gain insight into the mechanisms of speed improvement (Figure 3.4).

First, it was clear from these graphs that there was substantial pumping activity by both limbs on both days (mean # of pumps = 124 ± 48 SD for the impaired and 156 ± 37 SD for the unimpaired limbs on Session 3, and 124 ± 44 SD for the impaired and 153 ± 25 SD for the unimpaired limbs on Session 6). The number of pumps was not significantly different between arms (WRKS test, $p > 0.16$), and did not change significantly for either arm from Session 3 to Session 6 (WRKS test, $p > 0.98$). As confirmed by video recordings of the sessions, many of these pumps did not move the chair, especially in the early sessions. That is, all participants struggled to coordinate the clutching with the lever movement, especially the

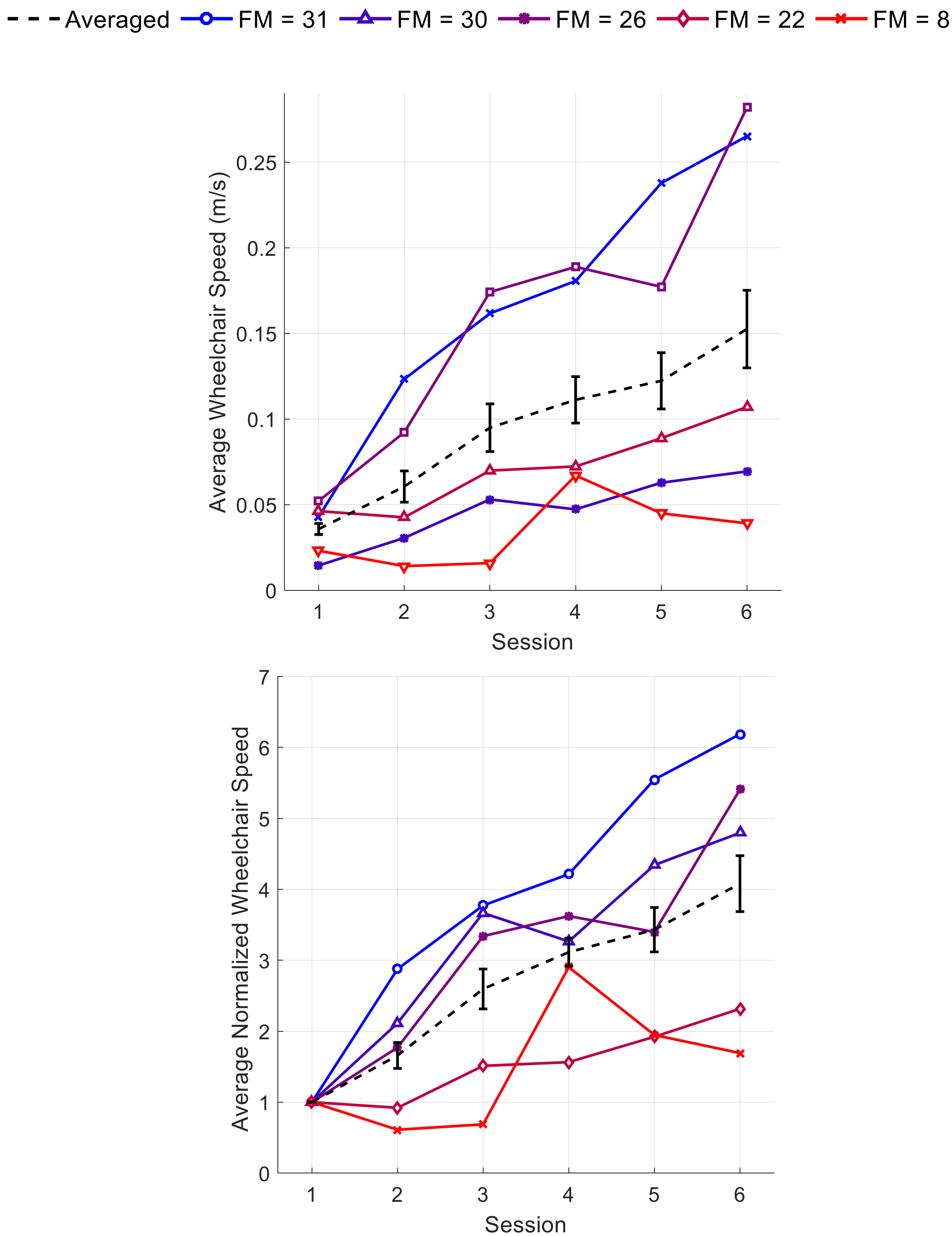

FIGURE 3.2: WHEELCHAIR SPEED ACROSS SESSIONS. **TOP:** AVERAGE WHEELCHAIR SPEEDS OF THE FIVE PARTICIPANTS AND AVERAGE WHEELCHAIR SPEED OF THE GROUP ALONG THE SIX SESSIONS. THE PARTICIPANTS WITH STROKE SIGNIFICANTLY INCREASED THEIR MEAN SPEED OVER THE SIX SESSIONS OF THE STUDY (FRIEDMAN'S TEST, $p < 0.001$). **BOTTOM:** WHEELCHAIR SPEEDS OF THE FIVE PARTICIPANTS NORMALIZED TO SPEED IN THE FIRST SESSION. BARS INDICATE STANDARD ERROR.

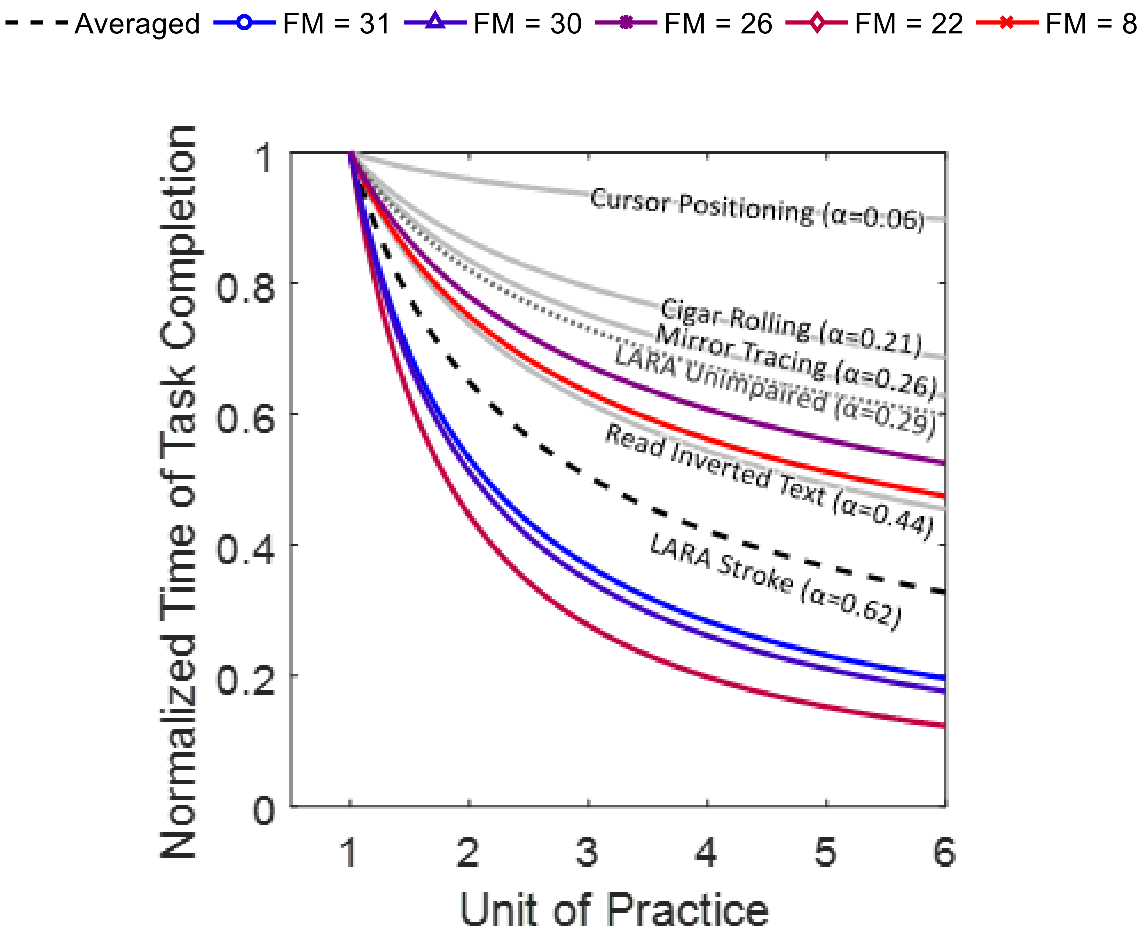

FIGURE 3.3: LEARNING CURVE FITS FOR LARA AND FROM A VARIETY OF OTHER MOTOR LEARNING STUDIES. FOUR REPRESENTATIVE LEARNING CURVES OF TWELVE CLASSIC EXPERIMENTS [13] (GREY) TAKEN FROM TOGETHER WITH AVERAGE LEARNING CURVES FROM IMPAIRED (BLACK DASHED) AND NONIMPAIRED PARTICIPANTS (DARK GREY DASHED) DRIVING THE LARA WHEELCHAIR, AS WELL AS THE INDIVIDUAL LEARNING CURVES FROM IMPAIRED PARTICIPANTS (RED TO BLUE). THE Y-AXIS IS NORMALIZED TO THE TIME REQUIRED TO COMPLETE THE TASK AFTER ONE UNIT OF PRACTICE. THE X-AXIS IS NORMALIZED TO MULTIPLES OF THAT UNIT OF INITIAL PRACTICE. THE POWER CURVE FIT TOOK ON A MASSIVE MOTOR LEARNING RATE FOR THE STROKE PARTICIPANTS. THIS RATE IS MORE THAN TWICE THAT OF THE MEAN LEARNING RATE REPORTED FOR A SAMPLE OF TWELVE CLASSIC LEARNING EXPERIMENTS (A = 0.28 ± 0.16; FIGURE 3.3). THE RATE FOR UNIMPAIRED PARTICIPANTS WAS SIMILAR TO THIS AVERAGE. THE LEARNING RATES FOR EACH SUBJECT FROM HIGHEST FM SCORE TO LOWEST: A = 0.91, 0.97, 0.36,1.2,0.41 WITH R-SQUARED VALUES: 0.75, 0.68, 0.43, 0.63, 0.86.

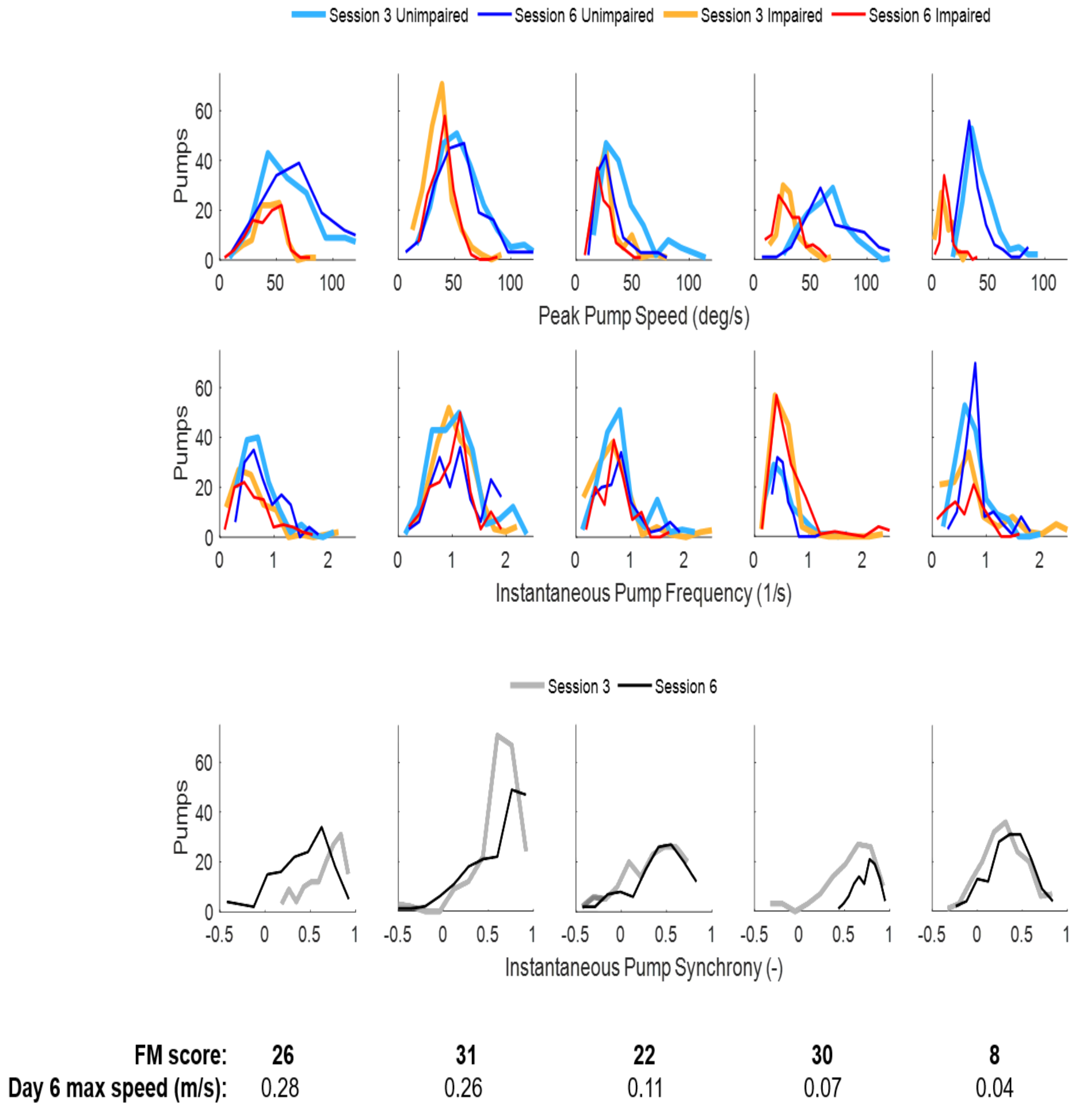

FIGURE 3.4: EXPLORATORY HISTOGRAMS DERIVED FROM LEVER GYROSCOPES FOR THE 5 PARTICIPANTS WITH STROKE ON SESSION 3 AND 6, ACROSS WHICH THEIR AVERAGE SPEED INCREASED 50%. TOP: PEAK PUMP SPEED FOR THE IMPAIRED (RED/ORANGE) AND UNIMPAIRED (BLUE/CYAN) ARMS FOR SESSION 3 (DASHED LINES) AND SESSION 6 (SOLID LINES). MIDDLE: INSTANTANEOUS PUMP FREQUENCY FOR THE IMPAIRED (RED/ORANGE) AND UNIMPAIRED (BLUE/CYAN) ARMS FOR SESSION 3 (DASHED LINES) AND SESSION 6 (SOLID LINES). BOTTOM: INSTANTANEOUS PUMP SYNCHRONY BETWEEN IMPAIRED AND UNIMPAIRED ARMS FOR SESSION 3 (GREY DASHED LINES) AND SESSION 6 (BLACK SOLID LINES). NOTE THAT THE GRAPHS ARE ORDERED FROM LEFT TO RIGHT FROM FASTEST TO SLOWEST PARTICIPANT ON DAY 6. SLOWER PARTICIPANTS AND PERFORMED MANY MOVEMENTS OF THE LEVERS WITH THE CLUTCH "OFF", AND THUS THESE MOVEMENTS DID NOT TURN THE WHEELS.

The peak pump speeds tended to be slower for the impaired limb (Figure 3.4 top row); specifically, the impaired side was significantly slower in Session 3 for participants with EUFM score 31, 26 30 and 8, and in Session 6 for participants with EUFM score 31, 26, and 8 (WRKS test, $p < 0.05$). The only participants that showed a significant change in peak pump speed from Session 3 to 6 were the participant with EUFM score 26, who showed a significant increase in his unimpaired arm pump speed (WRKS test, $p = 0.02$), and the participant with EUFM score 31, who, in contrast, showed a significant increase in his impaired arm pump speed (WRKS test, $p < 0.005$). Participants showed a preference for a pump frequency at or just below 1 Hz, and this frequency stayed roughly constant across Sessions 3 and 6 with no significant changes between arms or over Sessions for any participant (Figure 3.4 middle row). Increases in pump speed or frequency do not seem to be able to explain the 50% increase in speed from Session 3 to 6.

In terms of pump synchrony (Figure 3.4 bottom row), all participants showed predominantly positive correlations, concentrated at or above 0.5, indicating they tended to move the limbs in phase with each other. All participants altered their synchrony distribution with training, but the change was significant only for the participant with EUFM score 30 (WRKS test, $p = 0.04$). She increased her synchrony, as was the trend for three of the other participants. The fastest participant showed a different trend. He increased the percentage of pumps with zero synchrony – i.e. with one limb holding steady while the other limb pumped. From video analysis, it was clear that by Session 6 he had learned that he could drive the chair faster by pushing harder with his unimpaired arm, but that this caused steering error. To correct for the steering error, he used his impaired arm to hold one wheel still during a brief bout of backward pumping with his unimpaired arm, correcting the orientation of the chair. This resulted in more zero correlation pumps.

In summary, we observed subtle changes, if any, in peak pump speed and frequency, and a trend toward more synchronous arm movement from Session 3 to 6. Based on this data, we speculate that it is the coordination changes and improvements in clutching (not observable from the gyroscopic data but apparent in the video recording), that primarily explain the 50% increase in speed. A rough indicator of clutching efficiency, given that peak pump speeds and frequencies, stayed about the same, is distance traveled per pump of the impaired arm; this number increased from 0.18 m/pump ± 0.16 SD to 0.22 m/pump ± 0.14 SD from Session 3 to 6, a 22% increase that was not significant (WRKS test, $p = 1$).

C. Clinical, Physiological, and Motivational Outcomes

Learning to drive the wheelchair had very little effect on the clinical arm impairment: from Session 1 to Session 6 participants increased their UEFM scores by 2.2 points ± 1.1 SD, an improvement that neared significance (WSR test, $p = 0.06$, Figure 3.5A). Box and Blocks scores improved by 2.2 blocks ± 2.5 SD, a non-significant improvement (WSR test, $p = 0.3$, Figure 3.5B).

We assessed the short-term physiological effect of exercise with LARA using a heart rate monitor. On average, the participants significantly increased their heart rate by 36 beats per minute (BPM) ± 10 SD while they drove the chair (Figure 3.5C, measured on Session 6, WSR test, $p = 0.02$).

The IMI subscore for Competence increased significantly over the six training sessions (WSR test, $p < 0.05$), while the Effort and Usefulness subscores did not change (Figure 3.5D). On a scale of 1 to 7 of the IMI, where 7 represented “very true” and 1 represented “not true at all”, the participants scored the perceived usefulness/value of LARA as 6.8 points ± 0.4 SD, their perceived competence as 5.6 points ± 0.63 SD, and their perceived effort/importance as 5.8 points ± 2.2 SD, at the end of Session 6.

As assessed by the Visual Analog Pain Scale participants did not experience any pain or pain in Sessions 1 to 6 (mean VAPS score = 0 points ± 0 SD). Similarly, as assessed by the Modified Ashworth Score, upper limb spasticity did not change significantly from the post-Session 1 to the post-session 6 (MAS = 0 points ± 0.18; WRS test p = 1). There was also no significant change of limb spasticity within each session, (mean score pre-Session 1 = 1.75 ± 0.85 SD, mean score post-Session 1 = 1.7 points ± 0.82 SD, mean score pre-Session 6 = 1.8 points ± 0.89 SD, mean score post-Session 6 = 1.7 points ± 0.78 SD, WRS test p = 0.98, p = 0.85).

By the end of Session 6 participants increased their Motor Activity Log Amount of Use score to 1.2 ± 1, a non-significant improvement from the Session 1 score of 0.9 points ± 0.8 SD (WSR test, p = 0.55). Similarly, participants increased their MAL Quality of Use score to 1.3 points ± 0.9 SD, a non-significant improvement from the Session 1 score of 0.9 points ± 0.9 SD (WSR test, p = 0.69).

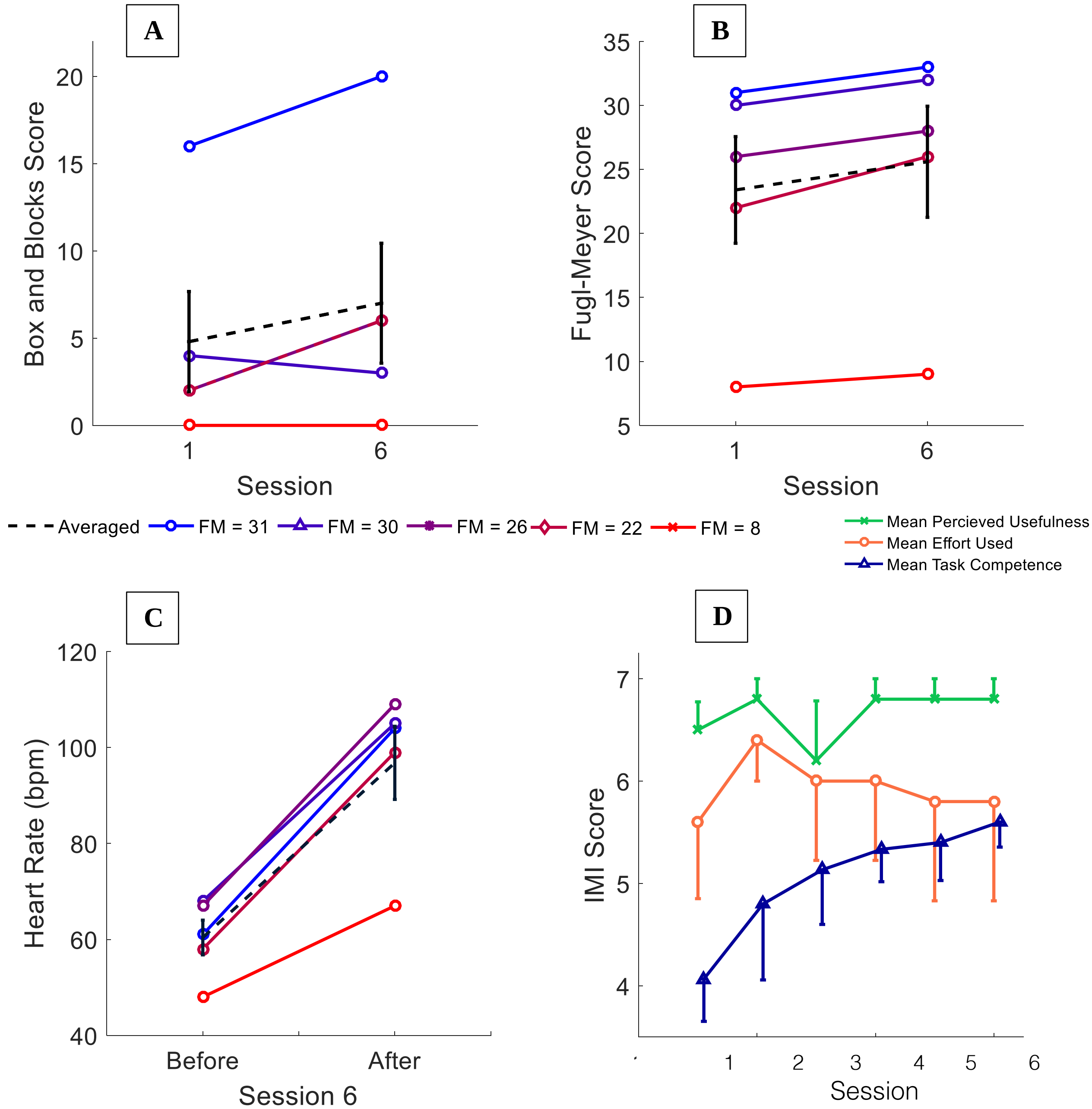

FIGURE 3.5: Clinical and Physiological measures. **A)** Change in Box and Blocks scores between Sessions 1 and 6. Between the first and last session of training, there was an average improvement of 2.2 ± 2.5 SD blocks, a non-significant change ($p = 0.1$). The participants with FM = 26 and FM = 22 both had Box and Blocks score of zero. **B)** Change in UEFM scores between sessions 1 and 6. Between the first and last session of training, there was an average improvement of 2.2 points ± 1.1 SD, a significant improvement ($p = 0.01$). **C)** Participants' heart rate increased by 36.4 BPM ± 10 SD while training with LARA on Session 6, a significant increase ($p = 0.02$). **D)** IMI subscore for Competence (triangles) increased significantly over the six training sessions ($p < 0.05$), while Effort (circles) and Usefulness (x's) scores did not change. Bars indicate standard error

3.4 DISCUSSION

How much skill learning is possible with a severely impaired limb? In the context of the LARA assistive device, we believe it is apt to call the amount of learning we observed “massive”. The learning rate, defined as the exponent in the power law fit to the speed improvements, was two times greater than the learning rate exhibited for the same task by young, unimpaired participants. It was also two times greater than the learning rates identified in a wide variety of tasks studied in classic motor learning experiments. While this learning of a bimanual skill did not manifest as a substantial reduction in clinical impairment, it translated into a large improvement in function – here, the ability to bimanually propel a wheelchair. Participants rated themselves increasingly competent at driving LARA and rated their new ability highly useful and valuable.

These results can be compared to the recent results from [15], in which participants with stroke learned to flex the elbow to a target. Participants were divided into unimpaired, mild, and moderate stroke groups. The moderate stroke group improved their speed-accuracy trade-off through learning, enough to move them up a classification level – i.e. to the level of performance that the mild stroke group exhibited before further training – even though their impairment level, assessed clinically did not change. Thus, learning was possible, even to the extent that the arm looked more normal for a specific task, but it did not “transform” the overall impairment status of the arm.

The present study adds to this the finding that a much larger amount of learning is possible given the “right task”. The functionally-meaningful and clear goal of propelling forward combined with the arm support that LARA provides are likely key factors here. Further, while it may not alter clinical impairment status of the arm, this learning can still be transformative because it can result in a new functional ability – here, the ability to bimanually propel a wheelchair. This is important for people with a stroke. By using LARA for wheelchair mobility, people can not only incorporate their “useless” arm into a meaningful task, but they can also greatly increase the amount of arm activity they experience throughout the normal course of the day. We hypothesize that this increased activity will improve long-term outcomes. In this context, LARA can be viewed as a dual assistive and rehabilitative device, an emerging paradigm in rehabilitation therapy.

The mechanism behind this massive learning was likely that participants became better at coordinating the arms with the hand-clutching. Future studies should instrument the hand clutch and wheelchair wheels as well as the levers to understand better the process of learning to drive LARA.

An interesting finding was that training with LARA for only a few minutes increased heart rate to the low end of the target exercise zone for the participants’ age group [117]. This suggests that LARA not only provides a platform for motor learning, but also a platform for cardiovascular exercise, which may promote health and fitness even in the early stages after stroke [106]. Likewise, LARA did not increase spasticity or pain, making it a feasible exercise device for people with severe arm impairment.

These results suggest an important goal for future work in rehabilitation engineering: developing assistive technologies (ATs) that unmask the robust but latent motor learning ability that people with severe impairments still possess; we propose calling such ATs “Unmasking Technologies” (UTs). We hypothesize that there are many UTs still undiscovered and that they can help people to achieve unexpected levels of function. It is not necessary that UTs do everything for the user; and indeed, such an approach may be counterproductive. For a UT, what matters most is not how well users do with it on the very first day, but, rather, how well they learn how to use it by exercising and improving their preserved abilities. There is plenty of room for learning even by the most impaired individuals.

CHAPTER 4: AN APPLICATION OF MACHINE LEARNING FOR GRASP-FREE YOKED CLUTCHING: A COMPARISION BETWEEN AUTOMATED AND SWITCHING SYSTEMS

Note: This following chapter is an unpublished manuscript.

This chapter builds on the previous chapter by extending the accessibility of the yoked hand clutching model of LARA with an automated assist system built with machine learning. The basis of this chapter is as discussed previously in Chapter 1: time spent actively performing therapy during inpatient rehabilitation is a crucial commodity limited by the time of the therapists and hence cost. For patients with severe disability, greater assistance may be needed in order to use therapeutic devices. In the context of LARA, we found previously in Chapter 3 that some individuals with stroke lacked the complex interlimb coordination or grip strength needed to master LARA to the same proficiency as others in their cohort, thereby limiting their potential gains (see Figure 3.2, page 61). In addition, people with bilateral hand impairment, such as is common after cervical spinal cord injury, would not be able to operate the yoked hand clutching system. Further, in the case that a user might desire to use LARA for longer periods throughout a day, removing the burden of hand clutching could potentially reduce fatigue.

In this chapter, we discuss the development and testing of a data-driven approach to assistance with an automated, electrically actuated, version of LARA to provide grip-free automated clutching. We discuss design rationale for classifier selection and the development of an online validation tool. Then we report results of an unimpaired subject learning the device over three days of practice and compare their success to using a switching system using a simple electric rocker switch. We find that the subject was able to drive around an environment representative of an inpatient setting using a K-Nearest Neighbor (KNN) classifier that was able to perform to 85% online accuracy.

4.1 INTRODUCTION

The LARA exochair facilitates extended use of the impaired arm in gross motions for inpatient rehabilitation, as well as providing the necessary mobility via yoked hand-clutching [115]. However, yoked hand-clutching has been shown to be a complex motor skill, and for some patients with chronic stroke, a limiting factor to wheelchair speed increase, device mastery and ergo possible recovery [118]. Thus, for such patients who lack the needed interlimb coordination to clutch or grip strength, an assistive system that could automatically clutch for them would provide an appropriate regression until they master more basic maneuvers. Other applications for automated hand clutching are for individuals with bilateral hand impairment, and to reduce fatigue for extended use of LARA.

It was seen in Chapter 3 chronic stroke subjects utilize individual-specific strategies to operate the chair, making heuristic based algorithms more obscure to implement. For such scenarios where rule-based approaches are unclear, machine learning approaches have classically been used to provide a data-driven way to generate robust control algorithms [119]–[121]. In stroke rehabilitation, such approaches have been proven viable in other applications such as brain-computer interfaces for example [122], [123].

This chapter first describes the design of an actuated LARA for automated clutching and data collection. Then the rationale for the chosen machine learning classifier is explained. The development of an online validation system is described, and finally, the off and online performance of the finalized weighted KNN classifier is examined in a case study of an unimpaired female over the course of three days, with final recommendations given.

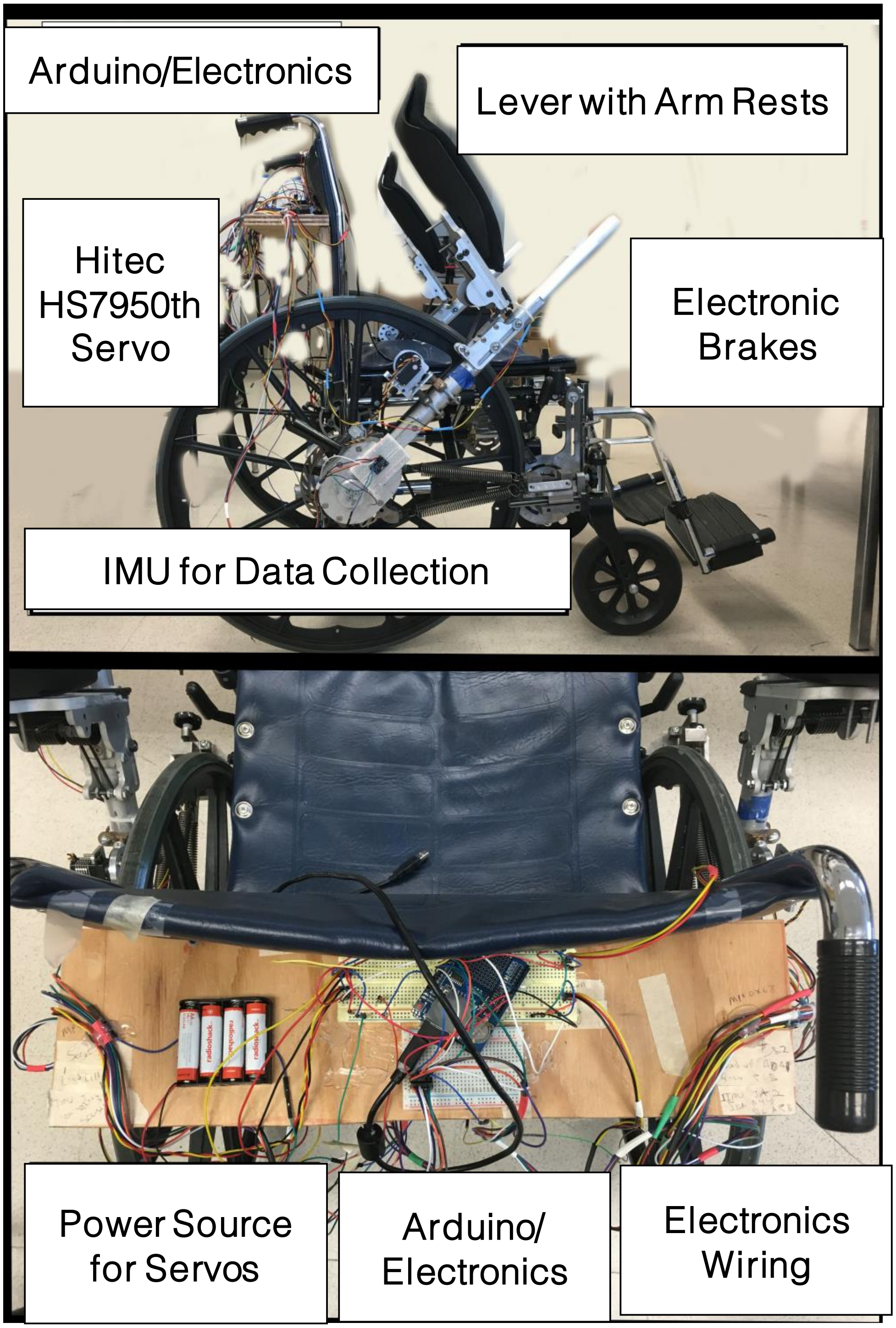

FIGURE 4.1: THE MACHINE LEARNING DATA COLLECTION SETUP ON THE LARA EXOCHAIR. AN ARDUINO MICROCONTROLLER READS BRAKE STATES FROM ELECTRIC ROCKER SWITCHES AND COLLECTS LEVER ANGLE, ANGULAR VELOCITY AND ANGULAR ACCELERATION FROM IMUS MOUNTED ON EACH SIDE OF LARA

4.2 METHODS

A. Design of Automated Version of LARA

The original one-way bearing version of LARA as described in Chapter 2 was modified to be electrically actuated by servos (Hitec HS-7950TH Ultra Torque Servo). These servos were attached to a Bowden cable transmission that connected the servos to pull bike disk brakes via activation of small electric rocker switches at the end of each lever. When users trigger the switch (referred to here on as "clutch") and release, the bicycle brake disk, which is mounted to the wheelchair wheel, is grabbed by the brake calipers which are on each respective lever. Thus by timing the correct activation and release of the clutch, users can propel themselves in a very similar manner to how one grabs and releases the wheelchair rim in traditional wheelchair propulsion [115]. For this version of LARA, each electric rocker switch commanded a microcontroller to activate the servomotors (Arduino M0 board). An SD card and shield system stored the data from two IMU sensors placed on each lever to obtain the lever kinematics (AnyCubic Data Logging Shield, MPU 9250). The IMU's attached to each lever recorded data at 5 Hz on the microcontroller (Figure 4.1).

B. Data Analysis

A dataset recording the left and right lever kinematics, as well as left and right servo states (0 for off, 1 for on), was taken using an experienced unimpaired female driving LARA. The subject drove the chair around a crowded room representative of a daily environment for an average wheelchair user for 10 minutes. She performed all maneuvers the LARA wheelchair was capable of and would be needed for use in an inpatient setting: propelling forward, turning in place left and right, and backing up in a mixed, randomized order. This data was imported into MATLAB 2015b and analyzed using the Classification Learner Application (Figure 4.2).

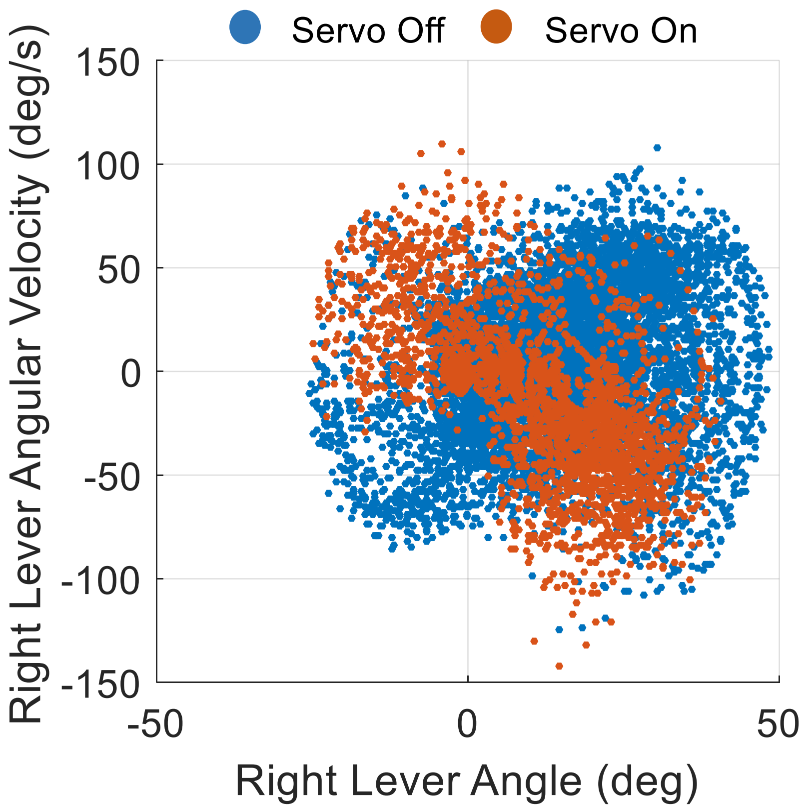

FIGURE 4.2: A SCATTERPLOT OF DATA TAKEN ON AN UNIMPAIRED FEMALE PERFORMING VARIOUS MANEUVERS ON THE LARA EXOCHAIR. THE ORANGE DOTS SIGNIFY THE CLASS 1 OR SERVO STATE "ON" LABEL, THE BLUE DOTS SHOW THE DATA THAT WAS CLASSIFIED AS CLASS 0 OR SERVO STATE "OFF".

C. Classifier Selection Rationale

Supervised learning algorithms were selected for exploration because labeled data was available, and design needs dictated for increased classification speed to provide rapid and accurate clutching [120]. Feature subtraction, a technique used to improve classification performance was performed manually by removing features until only a relevant set remained in order to optimize performance [124], [125]. Initially, the feature set included every feature the IMU's could provide: gyro and accelerometer readings in the x, y, z-axis, as well as derived features such as Euler angles, and angular accelerations. During this process, PCA was not used since it did not provide a boost in classification during the iterative feature selection process.

Feature engineering was also used to introduce an additional state variable, based on whether the user had crossed a threshold lever angle where previous work had shown clutching generally occurred. The Classifier Learning Application includes a breath of classic machine learning algorithms, including K-Nearest Neighbor (KNN) and neural networks, and the interested reader can refer to [126] for more detail. To select the proper algorithm, each type of classifier was run with each feature set during the subtraction process, and the accuracy of classification was used as the deciding metric. The final feature set was reduced to just 12 features and included lever angles, angular velocities, and accelerations as well as the state variables as some of the features.

The weighted KNN algorithm performed best throughout each feature subtraction iteration. This non-parametric approach works by computing the distance of incoming data points against a stored database of labeled data to cluster the data without making underlying assumptions of data linearity. However, this algorithm is computationally expensive and can be slow for large datasets since online it must compute a distance metric for each incoming data sample against every point in the stored database [120], [121]. Thus, only a small subset of the training data was used to provide best computational performance (Figure 4.3).

Finally, 10-fold cross-validation was used to manually tune algorithm-specific parameters (distance metric, number of neighbors) until they reached optimality. In this validation technique, the dataset is divided randomly into k groups or "folds" of equal size. The first fold acts as a validation set, and the classifier is tested on the remaining k-1 folds, and the misclassification rate is computed on the held-out fold. This process repeats k times, and a different fold is held-out; the average of the k estimates then produces the error. Through this method, the optimal parameters which had the lowest misclassification rate were calculated by iterating through different combinations of parameters. The optimal parameters were 3 neighbors and a City block, inverse distance metric. The offline results for each lever are presented in Table 4.1. Figure 4.5 showcases the offline validation of this algorithm as well [119], [120].

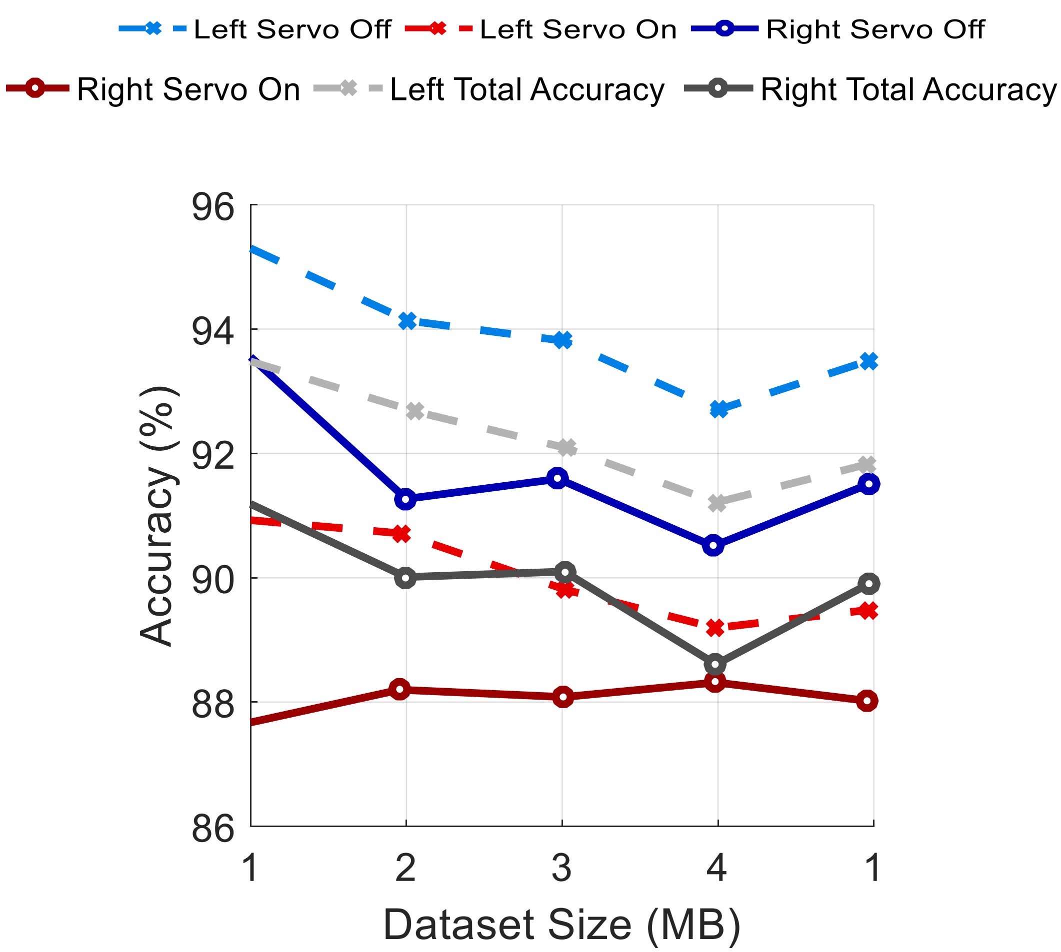

FIGURE 4.3: THE OFFLINE ACCURACY OF SERVO STATE, VERSUS DATASET SIZE. AS THE DATASET SIZE INCREASED, THE CORRECT CLASSIFICATION OF THE "ON" STATE OF THE SERVO TENDED TO BECOME MORE INACCURATE DUE TO COMPUTATIONAL LAG

D. A Validation System for Online Performance

A serial protocol between MATLAB and the microcontroller was used to validate online performance. The Arduino sent driver data via the serial port to a laptop that sat in the driver's lap (Lenovo W541 Thinkpad). MATLAB would classify this data in real-time using the chosen classifier, and send a servo command to the Arduino. In response, the Arduino would record and send the switch states from the driver as they drove, pressing the switches normally as if to actuate the servos as before. During online validation, the servos would not actuate and just send their recorded states to MATLAB so the intention to clutch could be compared with what the classifier would predict. MATLAB then would plot in real-time the predicted servo state, with the actual servo state, and performed an accuracy calculation (Figure 4.4A, Figure 4.4B).

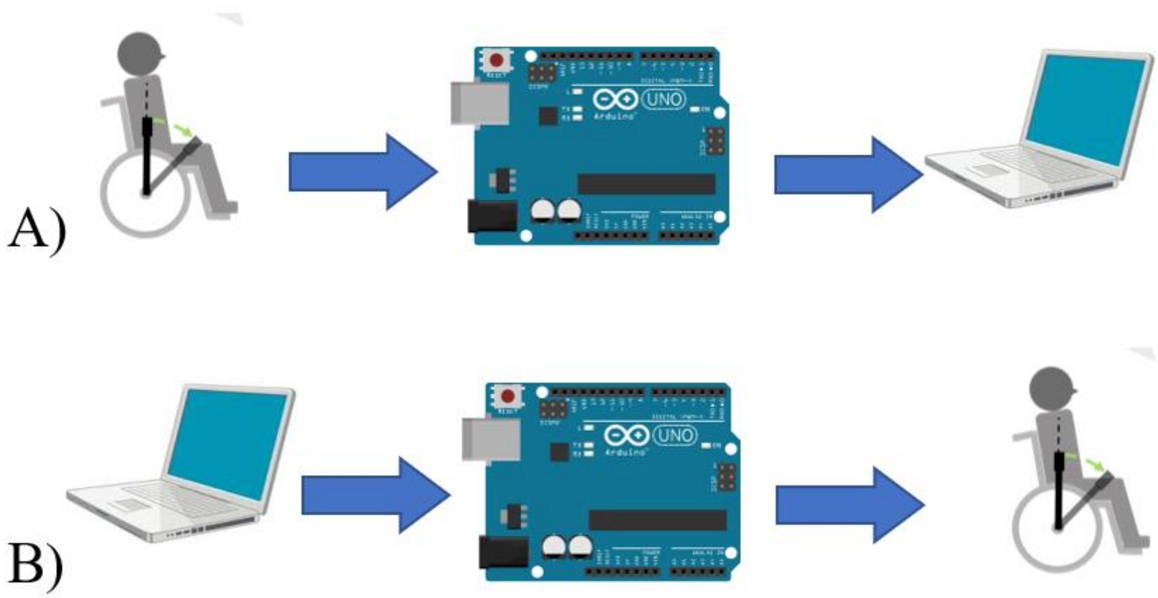

FIGURE 4.4A: DRIVER OPERATES LARA TAPPING THE SWITCHES TO INDICATE WHEN THEY WOULD USE THE CLUTCH, BUT THEY DO NOT HAVE VOLITIONAL CONTROL USING THESE SWITCHES WHILE AUTOMATED CLUTCHING IS ON. LEVER KINEMATIC DATA IS SENT TO THE MATLAB CLASSIFIER.

FIGURE 4.4B: CLASSIFIER IDENTIFIES WHEN TO TURN SERVOS ON. CLASSIFIER SENDS SERVO COMMAND TO ARDUINO WHICH CLUTCHES FOR THE USER. SIMULTANEOUSLY, THE CLASSIFIER CALCULATES THE ONLINE ACCURACY OF USER'S INTENT SENT PREVIOUSLY FROM (A) AND DISPLAYS ONLINE ACCURACY IN REAL-TIME.

E. Experimental Validation

Using the same environment that the data was collected in, an unimpaired subject drove in a circular pattern around the room for two minutes each day for three days to assess accuracy as well possible motor learning. Power curves (curve fit option 'power1') were fit using MATLAB's fit option to the performance accuracy:

$$T = BN^{-\alpha} \tag{4.1}$$

Here T is the time to finish the task, and B and α are constants and N is the amount of trials. α is the learning rate and B is the first trial's performance time or baseline [14]. See Chapter 1, section 1.1 for a discussion on motor learning.

4.3 RESULTS

A. Offline Results

Offline the classifier performed at 90% accuracy as assessed by the default option for k-fold cross-validation in the MATLAB Classification Learner application, which automatically validates classifiers models after creation using 5-fold cross-validation. (Figure 4.5, Table 4.1).

Table 4.1 Offline Performance of Weighted KNN Algorithm, With True/False Positive Rates for Each Class and Overall Accuracy

Algorithm Name	Right Servo State 0 Accuracy %	Right Servo State 1 Accuracy %	Overall Right Accuracy %	Left Servo State 0 Accuracy %	Left Servo State 1 Accuracy %	Overall Left Accuracy %
Weighted KNN	93.8/6.2	87.6/12.4	91.3	95.4/4.6	91/9	93.6

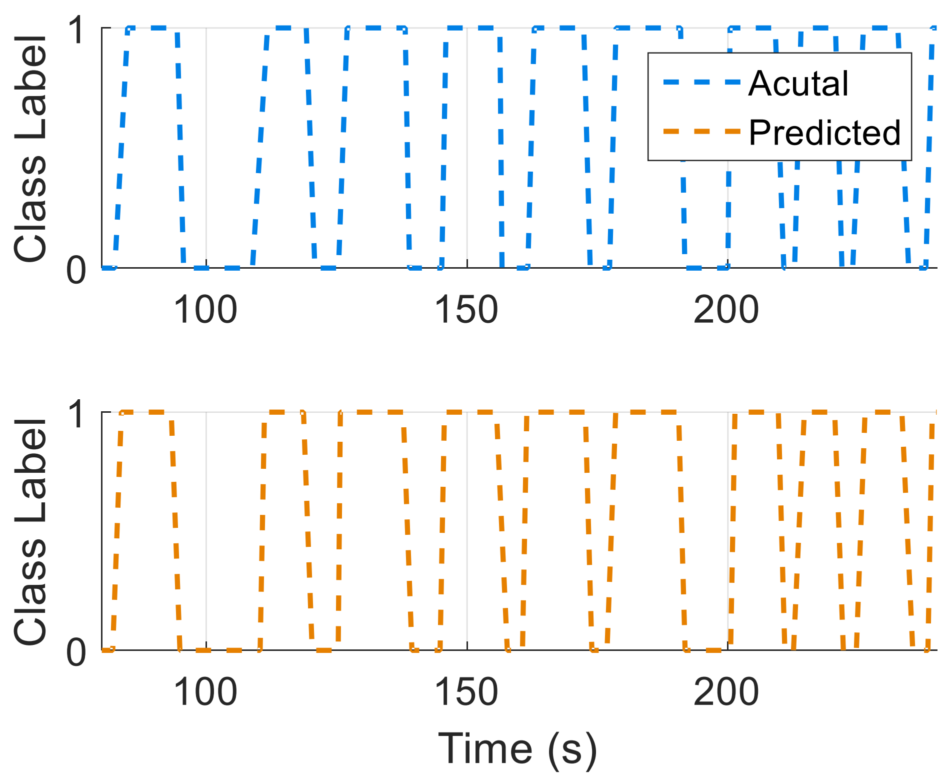

FIGURE 4.5: THE OFFLINE ACCURACY. ON TOP, IS A SUBSET OF THE OFFLINE DATA. BELOW IS THIS SAME DATA OVERLAID WITH THE PREDICTED SERVO STATES FOR THE RIGHT SERVO.

B. Online Results From an Unimpaired Subject

The subject was successfully able to drive around the represented environment and maneuver through obstacles. After three days of training, online performance was approximately 85% (Figure 4.6). This was a significant increase over time (t-test, $p<0.05$) (Figure 4.7). The rate of motor learning was α =0.14, which is not significantly different than mean learning rates reported previously for a range of motor tasks performed by unimpaired subjects in [13], (t-test, $p<0.05$). Of the three LARA tasks studied in this thesis, yoked hand clutching by unimpaired subjects, yoked clutching by impaired subjects, and unimpaired automatic clutching, the automatic clutching had the lowest learning rate (Figure 4.8).

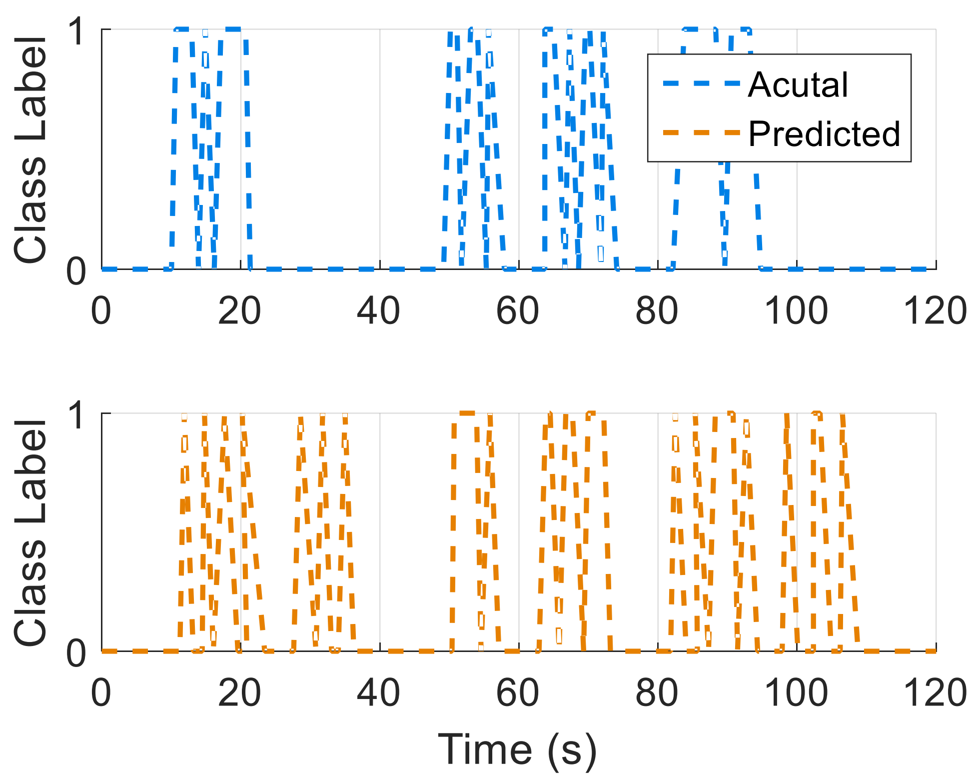

FIGURE 4.6: THE LEFT PREDICTED SERVO STATE VS ACTUAL LEFT SERVO STATE AS ACTUATED BY THE USER. THE ACTUAL REAL-TIME ACCURACY WAS 85%

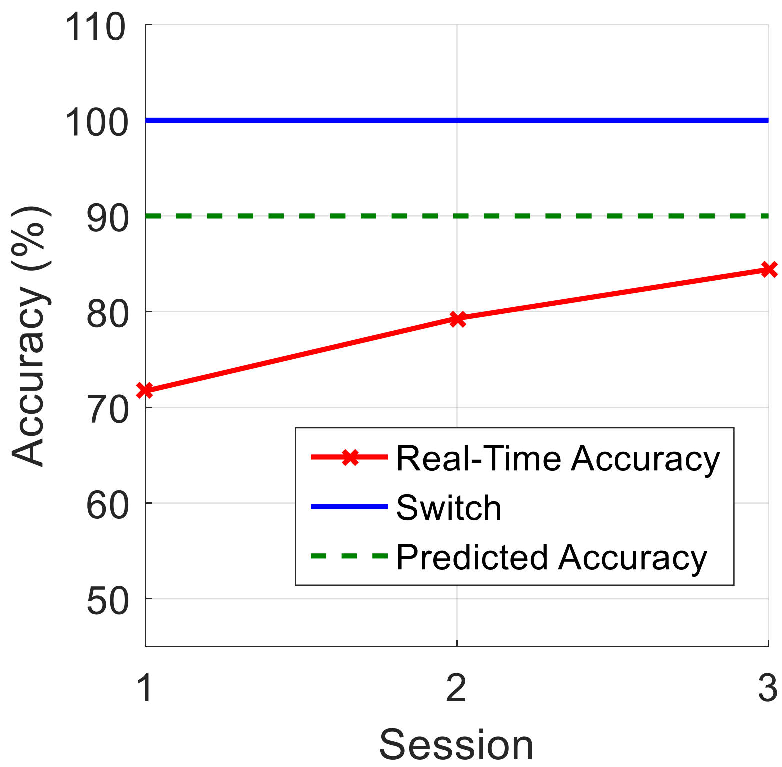

FIGURE 4.7: ALGORITHM ACCURACY OVER TIME. THE SWITCH HAD 100% ACCURACY FOR A SUBJECT WITH BASIC HAND STRENGTH. WHILE THE REAL-TIME ACCURACY OF THE AUTOMATED SYSTEM LAGGED BEHIND THE PREDICTED ACCURACY, OVER TIME THE SUBJECT LEARNED THE SYSTEM, INCREASING ITS ACCURACY.

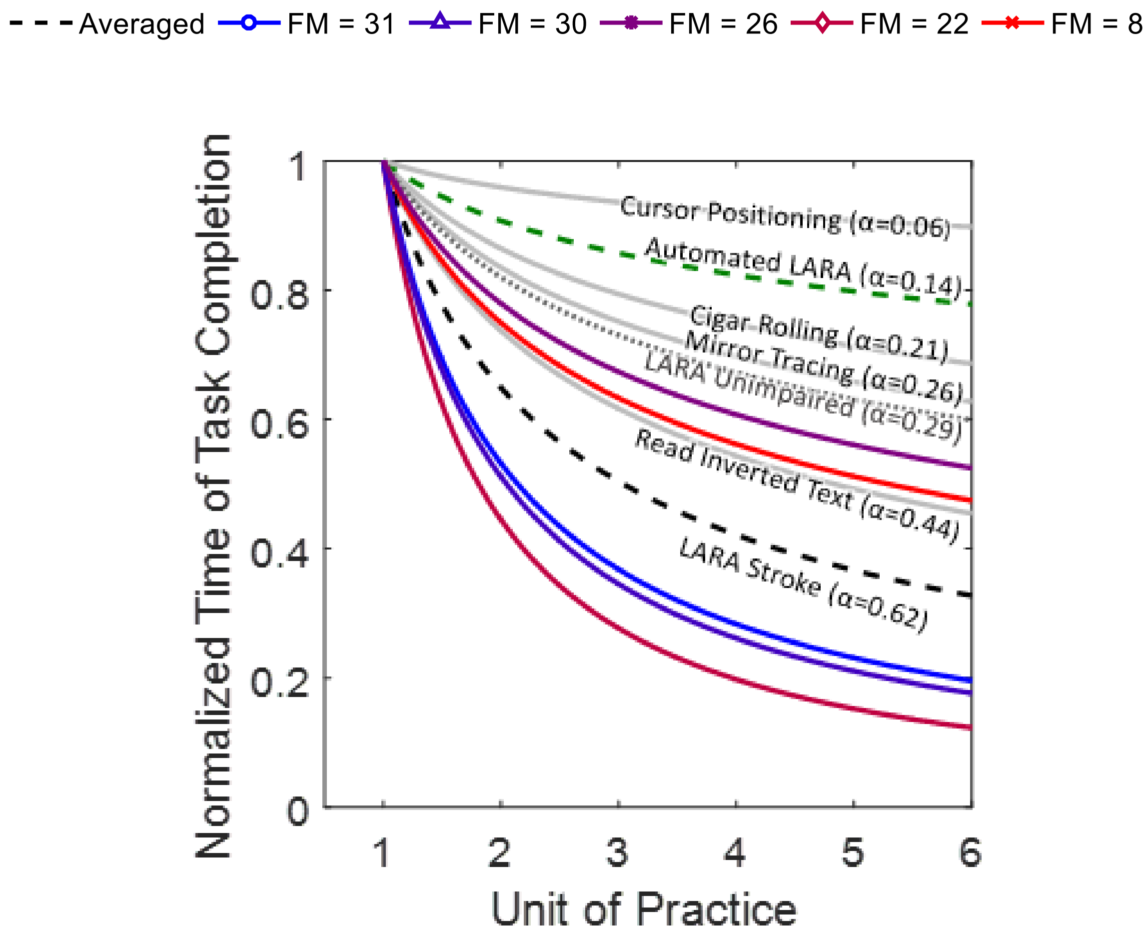

FIGURE 4.8: COMPARISON OF LEARNING CURVES FOR LEARNING THE YOKED HAND CLUTCH VERSION OF LARA IN NONIMPAIRED SUBJECTS (DARK GREY DASHED), STROKE SUBJECTS (BLACK DASHED, INDIVIDUALS IN RED TO BLUE) AND A SINGLE SUBJECT LEARNING THE AUTOMATIC CLUTCH (GREEN DASHED). IN GREY, A SAMPLING OF TWELVE CLASSIC EXPERIMENTS TAKEN FROM [13] THE X-AXIS IS NORMALIZED TO MULTIPLES OF THAT UNIT OF INITIAL PRACTICE. THE Y-AXIS IS NORMALIZED TO THE TIME REQUIRED TO COMPLETE THE TASK AFTER ONE UNIT OF PRACTICE. STROKE SUBJECTS HAD THE LARGEST LEARNING RATE, WHEREAS THE AUTOMATED CLUTCH HAD THE LOWEST OF THE THREE LARA TASKS. SEE FIGURE 3.4 FOR LEARNING GRAPH OF UNASSISTED LARA IN STROKE AND HEALTHY SUBJECTS, CHAPTER 3, PAGE 54.

C. Comparison of Automated and Switching Systems

Due to increased complexity from additional parts, the automated clutching system was costlier than the simple switch due to requiring a laptop to run the classifier system. However, the automated clutch provides for greater accessibility, if at lower accuracy for someone with no grip force.

4.4 DISCUSSION

How can technology improve accessibility for individuals with more severe impairment? In relation to LARA, automatic clutching can provide a possible solution for individuals with stroke who physically lack either strength or interlimb coordination to master the complex motor skill to clutch. At the same time, this system can preserve the increased mobility benefits of backing up given by yoked hand clutching over a simpler one-way bearing design [115].

However, such technology also comes at a cost, and the trade-off it provides in terms of ease of use to someone with a lower function needs to be counterweighed with financial accessibility which is a large limiting factor for therapy [127]. The automated system cost more than the electric rocker version of LARA. At the same time, for someone with at least basic hand function, its performance lagged behind 15% using a simple switch system in terms of accuracy. Similarly, the automated system had higher complexity than the simple switch in terms of equipment involved. There was also a significant change in performance accuracy that occurred with the automated system for this subject as they learned, which did not occur using the simple switch. The learning rate this subject experience was also not significantly different than the normative value for healthy subjects learning motor tasks. Of the three LARA tasks studied, it had the lowest learning rate at least for this subject.

For rehabilitation mental engagement has been shown to be a crucial key to improving motor outcomes after stroke [20], [128]. Thus, machine learning could provide a feasible way for a system not well served by traditional heuristic based assistive algorithms to give tunable difficulty/ This would be a way to transition more severe individuals to eventually using traditional yoked hand-clutching. For the end use case of LARA in an inpatient rehabilitation setting, the differences in cost could likely be mitigated and not affect accessibility, as it would not fall on the patient to pay for different versions of the device depending on the pricing model used by the center. It is worth noting however that any savings in the high costs of therapy could provide a great benefit to the patient, since insurances typically only cover therapy for several months, limiting gains for patients [129], [130].

Similarly, time spent actually performing therapy is a crucial commodity for a person with stroke recovering in an inpatient setting. For example, it was found that stroke patients only achieve about 32 repetitions per session while in this setting [105]. The extra bulk or complexity of the automated system could provide barriers that could slow workflow for therapists. This could further reduce therapy which is crucial for patients in early stages of recovery to regain function. Further studies would need to be performed using individuals with stroke to assess how setup times effect total time spent doing therapy between the two systems in an inpatient setting. Similarly, larger scale research with healthy and stroke subjects comparing the automated system and the with the original hand yoked clutch model studied should be investigated.

While the learning rate in this single subject was similar to the mean values reported for healthy subjects learning motor tasks, there are some confounding factors that need to be addressed in future work. First, this study only demonstrated the feasibility of the system, and more testing would need to be done with more subjects to assess if learning rate observed and results are generalizable. The subject was also unimpaired: some research suggests there are differences in motor learning after stroke [15], [118]. Similarly, the subject provided the ten-minute training data set for the study and tested it on themselves. Thus, this previous experience with the device could have masked greater learning a complete novice may have experienced. Indeed, in [115] the time constant of learning was only 3 days or approximately 15 minutes of driving spread out over those days, so this subject may have already overcome the initial learning curve associated with LARA. Thus, rates of learning should be investigated on a larger scale with individuals with stroke and completely naïve drivers and compared to experienced drivers to parse out these differences.

Overall, this work suggests important potential implications for rehabilitation. First, it adds to the body of work demonstrating the potential of machine learning approaches to improve therapy delivery. In fact, this particular application of machine learning to assist in driving a lever wheelchair is the first-of-its-kind as far as we know, as previously machine learning has been used more in mobility contexts, to create smart navigation systems in power wheelchairs, as opposed to assisting someone with bilateral hand impairment manually operate a lever drive chair [75]. Second, it highlights a key conflict in the field of robotic therapy: the trade-off between improving physical accessibility versus financial accessibility for patients. As seen in previous work detailing the development of LARA, we have shown that this device can provide similar clinical outcomes with much less device complexity as compared others in the field [118]. These facts along with this explorative piece imply that simple devices and approaches may be more warranted to make a real-world impact for patients affected by a costly injury. This also illustrates a continued need in rehabilitation robotics to balance the benefits of new technology with care to develop more low-cost tools and streamlined approaches to provide all patients with therapy at greater accessibility in every sense of the word.

CHAPTER 5: DESIGN OF A WEARABLE ROBOT TO ENABLE BIMANAL MANIPULATION AFTER STROKE

Note: Chapter 5 has been published as the following and is presented here in an edited format: Norman, S.L., Sarigul-Klijn, Y., & Reinkensmeyer, D.J. (2016, April). Design of a Wearable Robot to Enable Bimanual Manipulation after Stroke. Southern California Robotics Symposium 2016.

This chapter builds on the concept of low-cost, simpler devices to promote therapy with the design and performance testing of a simple hand exoskeleton for assistance after stroke. As discussed in Chapter 1, individuals with hemiparesis after stroke cannot use their hands together to achieve bimanual functions, as they lack the ability to extend the fingers in one hand, making that hand unavailable to play a supportive role. Wearable robots could potentially enable finger extension, but most previous devices are bulky and focus on multiple finger assistance. We developed a wearable Robotic Hand Extension Device (RHED) that combines the user's residual capacity for finger flexion along with robot-assisted finger extension, for only the thumb and index finger. Use of a Bowden cable allows the majority of mass to be placed on the user's forearm, while also preventing cable slack. For intuitive control, a magnetic ring on the unimpaired hand and a three-axis magnetometer on the impaired arm facilitate touch-free gesture commands. The user can choose different postures of the impaired hand, such as the four permutations of the open and closed thumb and index finger. By selecting an appropriate hand posture via gesture command, the user can incorporate the impaired arm into a supporting role, facilitating bimanual activities such as jar opening and toothpaste uncapping.

5.1 INTRODUCTION

Finger extension is a common movement impairment after stroke, the leading cause of disability in the United States [35], [131]. People with stroke often resort to compensatory strategies using the unimpaired limb, but most tasks require at least some degree of bilateral function (e.g. the impaired hand stabilizes a piece of paper while the other hand writes) [132]. Because activities of daily living often require bilateral actions, grasping function in the impaired hand is paramount to quality of life. In recent years, declining cost and size in assistive robotics have made them increasingly attractive for use by people with impairment. However, many of the existing robotic-assistive devices for people with stroke remain impractical for day-to-day use [34]. In this section, we introduce the design of the Robotic Hand Extension Device, or “RHED”.

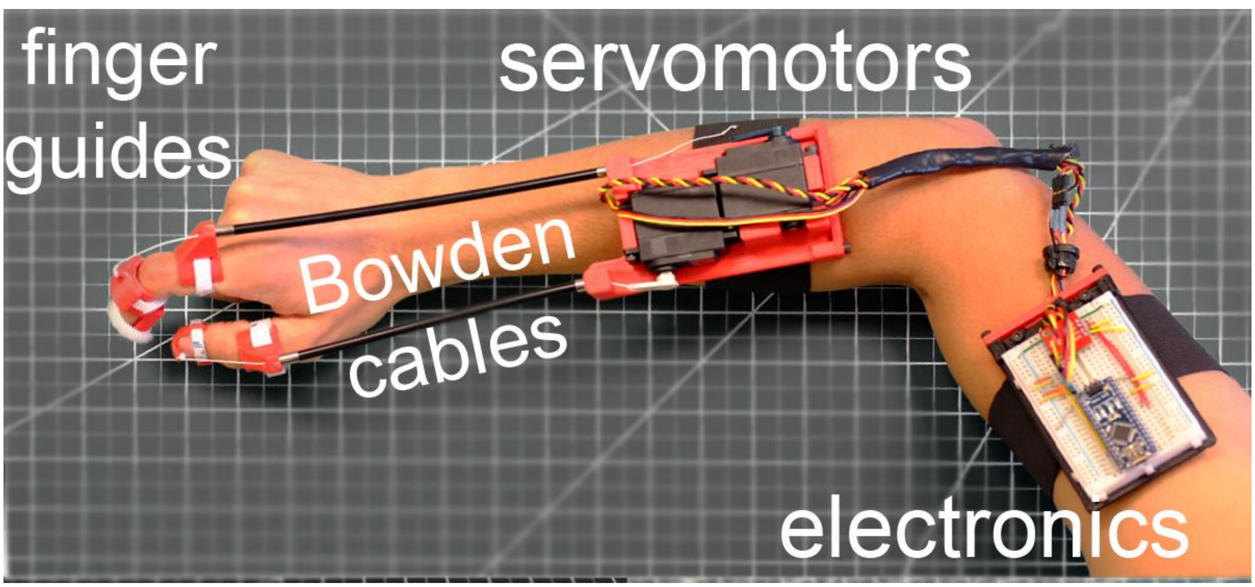

FIGURE 5.1: RHED DESIGN AND SETUP. TWO LIGHTWEIGHT SERVOMOTORS MOUNTED ON THE FOREARM PROVIDE ACTUATION FOR ASSISTANCE IN HAND EXTENSION.

5.2 DESIGN AND METHODS

We designed RHED to capitalize on the user's residual flexion ability by limiting actuation to solely extension. This allows the device to be simple, lightweight, and low-cost, ideal attributes for a wearable device. The user is uninhibited during flexion, encouraging dexterity in coordinated grasping tasks. However, when the user struggles to achieve full extension, RHED uses a unique cable assembly to transmit extension forces to the fingers remotely from actuators located on the forearm (Figure 5.1). Flexible/incompressible cable sheaths allow for wrist manipulation without causing cable slack. This allows RHED to facilitate finger and thumb extension without limiting the freedom of the wrist and remaining finger Many wearable assistance devices have relied on agonist/antagonist muscle activation via electromyography (EMG) or force-based intention sensing [34], [133]. However, people with stroke typically display abnormal muscle synergies, including coactivation of the flexor/extensor pair and inability to produce positive extension force, making these approaches suboptimal [35]. Additionally, past research has shown that rewarding abnormal synergies with robotic movement can result in their cementation through reinforcement learning [134]. For RHED, we use the unimpaired hand as a means of controlling three discrete grip strategies for the impaired hand. The user wears a magnetic ring on the unimpaired hand and a three-axis magnetometer on the robot/impaired arm. A simple wave of the unimpaired hand in one of three gesture commands results in the extension of the thumb and index finger, or a combination thereof (Figure 5.2).

$$posture = max\,index\left(\begin{bmatrix}\vec{F}\cdot\vec{F}_1\\ \vec{F}\cdot\vec{F}_2\\ \vec{F}\cdot\vec{F}_3\end{bmatrix}\right)\cdot\left(\left|\vec{F}\right| \geq \ F_{threshold}\right) \tag{5.1}$$

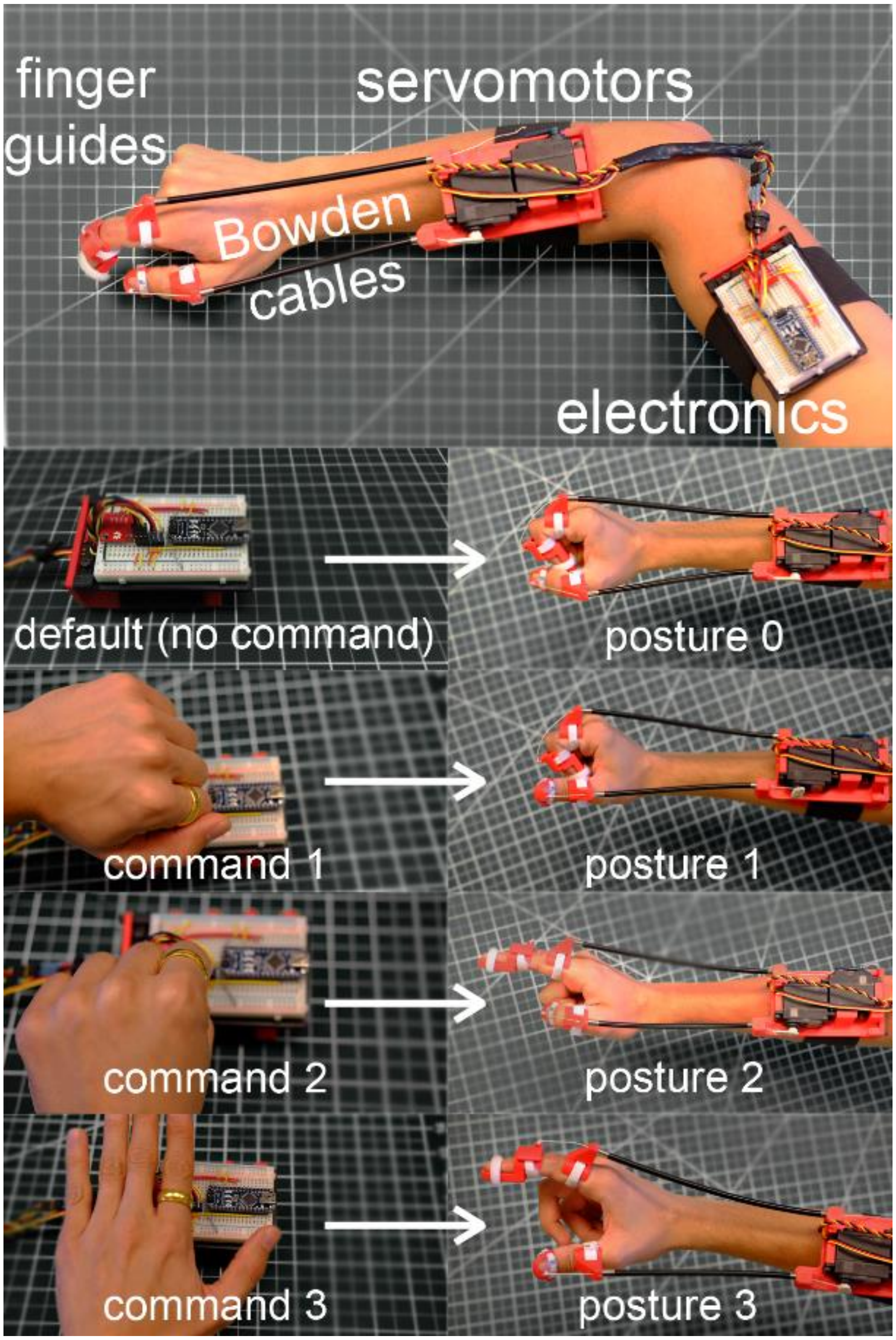

FIGURE 5.2: **TOP**, THE ROBOTIC HAND EXOSKELETON DEVICE (RHED). ARM MOUNTS HOUSED THE ELECTRONICS, BATTERY, AND ACTUATORS. INDIVIDUALIZED FINGER GUIDES STRAPPED TO THE PROXIMAL AND MIDDLE PHALANXES OF THE INDEX FINGER, AND TO THE PROXIMAL AND DISTAL PHALANXES OF THE THUMB. **LEFT,** AN EXAMPLE OF THE THREE DISCRETE COMMAND GESTURES. A MAGNETIC RING ON THE UNIMPAIRED HAND WAVES OVER A THREE-AXIS MAGNETOMETER. RIGHT, FOUR RESULTING GRIP POSTURES ACCORDING TO THE INPUT COMMAND. POSTURE 0 IS UNASSISTED. RHED ASSISTS COMBINATIONS OF THE USER'S FINGER AND THUMB IN EXTENSION FOR THE REMAINING THREE POSTURES.

As a special consideration for people with spasticity after stroke, the device was designed to be donned and doffed using only the unimpaired hand. People with spasticity typically exhibit difficulty with one-piece designs such as gloves, as it is difficult to extend the impaired fingers to a position in which to don the device. We designed finger joint cable guides attached by hook-and-loop straps. The guides attach to the proximal/middle phalanxes of the index finger and the proximal/distal phalanxes of the thumb. Rapid prototyping methods made individualized guides possible, allowing for comfort and ease of use (Figure 5.3).

FIGURE 5.3: EXAMPLE USES OF RHED IN ACTIVITIES OF DAILY LIVING. HERE THE USER ALLOWS RHED TO ASSIST IN OPENING A TUBE, SOMETHING A PERSON WITH STROKE WOULD HAVE GREAT DIFFICULTY DOING.

5.3 DISCUSSION

A primary design goal of RHED was to exceed the passive flexion torques of the user's fingers to allow extension that could support bilateral movement tasks. A statics analysis showed that RHED applies a minimum torque about the proximal interphalangeal joint that is greater than the maximum passive torque previously calculated in people with stroke [135]. This was verified in preliminary tests where RHED was capable of extending unimpaired subjects' index finger and thumb while they maintained a flexion force. The device can be seen assisting the non-dominant hand in extension to complete a supporting role in a bimanual functional task in Figure 5.3. This is just one example of many daily tasks that would now possible with an assistive device like RHED.

CHAPTER 6: GAIT REHAB ADAPTIVE MACHINE: FEASIBILITY AND DESIGN OF GRAM, A WALKING LINKAGE POWERED WHEELCHAIR FOR LOWER BODY THERAPY AND ASSISTANCE

Note: This following chapter is an unpublished manuscript.

This chapter builds on Chapters 2 through 4 by extending the exochair concept to the lower body, with the design and testing of an exochair called GRAM. As seen in Chapter 1, leg-based exercise is difficult to achieve, and typically involves the use of expensive, sometimes unsafe and mechanically complicated devices like exoskeletons which assist users in a walking motion. Here, we introduce a possible solution to fulfill this called GRAM (Gait Rehab Adaptive Machine). GRAM utilizes a 6-bar linkage that captures the trajectory of human gait and couples this motion to a wheelchair's propulsion. First, we discuss design considerations and theory. Then we show data from an unimpaired adult performing a test analogous to the 10-meter walk test used for therapy with GRAM and provide a comparison to the subject's performance on the task via walking. We found the subject was able to significantly increase their speed, repetition rate, and projected repetitions when using GRAM. The analysis also showed that GRAM was a task that could be improved upon most within in the session. This coupled with measures of heart rate and physical efficacy suggest GRAM's feasibility, and that it may be particularly well suited for a potential application for individuals with walking impairments like stroke to increase leg exercise opportunity and accessibility. This chapter then serves as an extension of the work done in Chapters 2-4 with LARA and provides a new approach to gait rehabilitation, which the current progress of was discussed in Chapter 1.

6.1 INTRODUCTION

Nearly half of individuals with stroke experience some form of long-term disability and stroke is one of the main causes of wheelchair use in the United States [63]. Early rehabilitation in the acute phase of stroke has been shown critical to promoting motor plasticity and patient outcomes. However, research shows that only 32% of the time is spent during inpatient rehabilitation in active therapy, while the rest of the time on other activities around the ward [136]. For walking impairment, it is especially important for patients to experience similar force loading and practice the patterning of gait in order to recover [137]. However, in a typical therapy session focused on gait rehabilitation patients only will roughly 300 steps on average, far below what has been thought needed for humans to learn how to walk [105], [138].

Currently, technology options to provide therapy include overhead treadmill systems like the Lokomat that assist patients in a walking motion or exoskeletons like the ReWalk, but these systems have considerable cost, complexity, and bulk, limiting the amount of clinics that can obtain them to provide to patients [17], [18]. Likewise, attempts to add therapy to wheelchairs for stroke patients have been mostly limited to pedal devices [88], [103]. These devices also have limitations because pedaling does not provide a similar muscle patterning to walking and some devices do not utilize the affected side in their control [139]. This second type of device does not promote use of the impaired limb, which can further disuse and presumably impairment.

This paper details the design and feasibility evaluation of a novel wheelchair called GRAM (Gait Rehab Adaptive Machine) for potential application in walking impairment recovery in stroke and other injuries. First, we discuss the theory for using a single degree of freedom 6-bar linkage that simulates the trajectory of human gait as a propulsion mechanism for a wheelchair. Next, we demonstrate feasibility in an unimpaired subject to drive the chair. We explore physiological outputs and overground speeds for a single subject in a distance test analogous to the 10-meter test used in physical therapy [140]. We discuss potential implications for rehabilitation and future work.

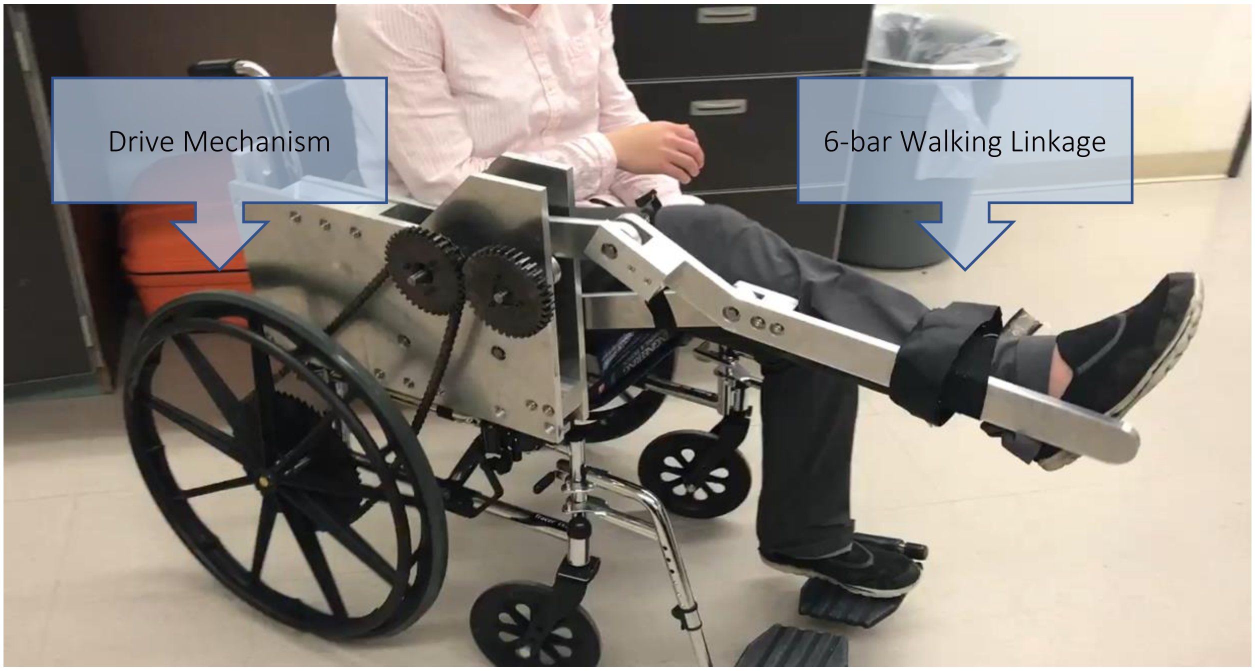

FIGURE 6.1: THE NONIMPAIRED SUBJECT IN GRAM. A SIX-BAR MECHANISM GUIDES THE LEG IN A WALKING TRAJECTORY, WHICH COUPLES TO THE WHEELCHAIR VIA A GEARING SYSTEM TO PROVIDE PROPULSION WITH EACH REPETITION.

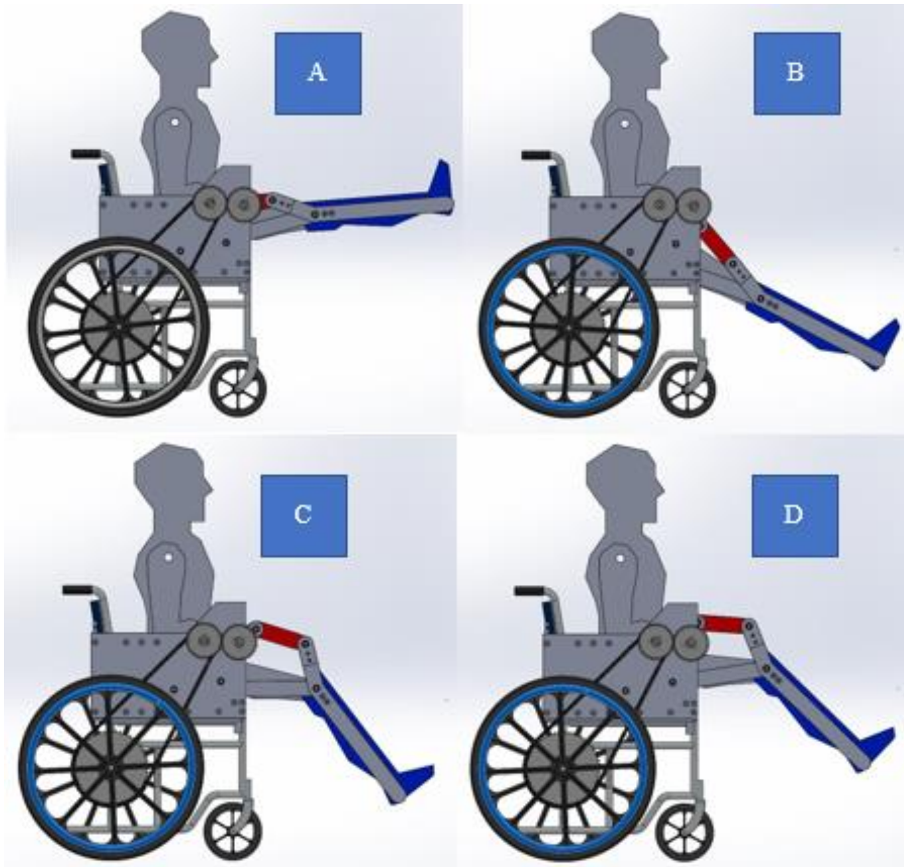

FIGURE 6.2: A CYCLE OF WALKING IN GRAM (A-D). EACH TIME A REPETITION IS COMPLETED, THE REAR WHEELS OF THE WHEELCHAIR TURN 1/3 REVOLUTION, ALLOWING THE USER TO PROPEL THEMSELVES WITH A WALKING MOTION.

6.2 METHODS

A. Design Rationale

The basis for GRAM (Gait Rehab Adaptive Machine) is to reduce complexity and costs associated with other leg therapy devices, all while providing the necessary practice with a relevant walking-like motion to be effective (Figure 6.1). Thus, a single degree of freedom six-bar linkage was used to because unlike other designs only a single actuator would be required to provide assistance if needed. Likewise, by coupling a walking motion to a wheelchair, users reduce time to therapy since no transfers are required like with other systems, increasing safety and therapy opportunity (Figure 6.2).

This mechanism was designed by recording motion capture data of a healthy person walking on a treadmill with a Vicon MX three-dimensional motion capture system. Markers were placed at the toe, ankle, knee, and hip to acquire this data as the user walked. This data was made up of 23 trajectories ranging between 199-210 data points, as seen in the following figure below.

FIGURE 6.3: THE MOTION CAPTURE SETUP WHICH PRODUCED THE DATA THAT GENERATED THE LINKAGE USED IN GRAM. TAKEN AS IS FROM [141]

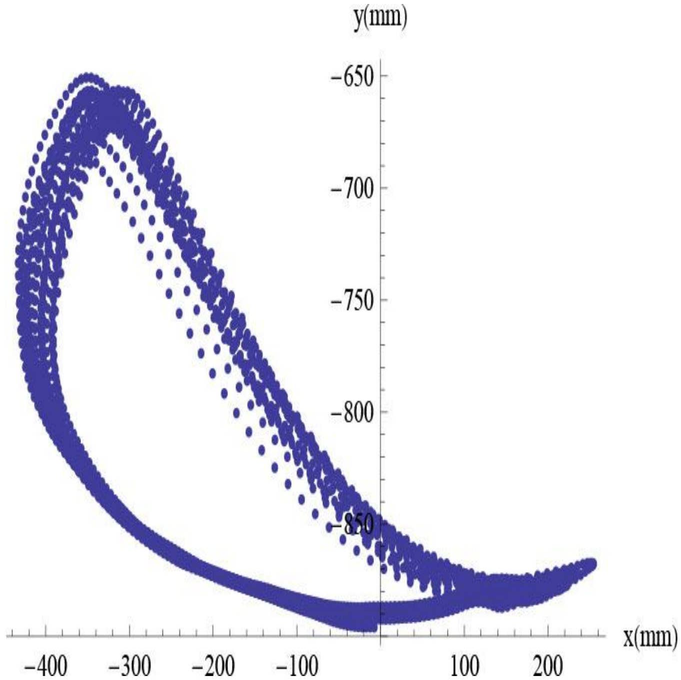

FIGURE 6.4: THE ANKLE COORDINATES OVER SEVERAL GAIT CYCLES, TAKEN FROM [141]

B-Splines were used to simplify the data. The motion capture data was used as control points on these B-splines in order to represent the ankle trajectory as a single parametric equation. 60 values for t between 0 and 1 were chosen and distributed around the curve evenly, the parametric equation was then evaluated at these points to obtain 60 precision points for the linkage synthesis. Other values like leg lengths were determined from the motion capture data. Figure 6.4 shows the ankle trajectory. The linkage was synthesized to match the volunteer's foot orientation and ankle trajectory [141]–[143]

The general procedure for this synthesis is as follows and is reproduced below from work by [141]–[143]. First, the linkage must be designed to match the proper path –in a step known as "path synthesis." Since the linkage in question was a Stephenson III six-bar, a set of N points, Pj, j = 0 … N-1 which represents the mechanism’s coupler curve or the desired trajectory for the linkage were defined. From these points, a set of 11 loop equations in the reference position were created. Since the linkage is meant to attach at the hip, a coordinate transformation on the ankle data was performed to ensure proper reference position.

This equation set was not traceable, and so it was further reduced to an 11-point path synthesis to simplify the problem, and further details can be found in [141]. From the solution of this, starting design parameter vectors were obtained and optimized to minimize the distance between the linkage to the coupler curve, to obtain a compact linkage. These solutions were sorted to yield the linkage used in GRAM, shown below in Figure 6.5 [141]–[143].

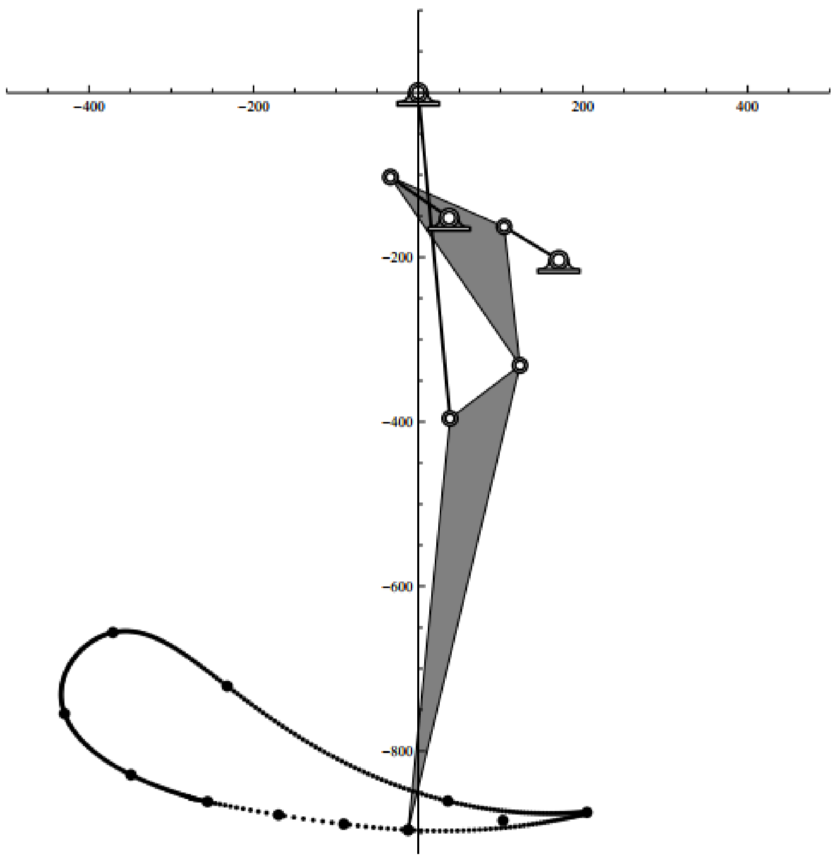

FIGURE 6.5: THE SIX-BAR LINKAGE USED AS A PROPULSION MECHANISM IN GRAM, TAKEN FROM [141]

B. Experimental Setup

Prior studies informed both the experimental setup and data analysis methodology for this work [115], [118]. An unimpaired female (age = 25 years) drove a wheelchair outfitted with a single GRAM unit on the right side with their right leg for 15 repetitions (about 0.6 meters distance traveled per repetition, for a total distance of about 9 meters). During this test, the contralateral arm remained in the subject's lap and their contralateral leg was on the wheelchair's footrest. The subject then walked for the same repetitions (15 repetitions of walking, where a repetition was counted as a heel strike of their right foot) for the same stride length as GRAM (0.6 meters) per repetition to provide walking data as a comparison for some of the analyses. In both tasks, the goal was to go as fast as possible, and in the case of the walking task, to do so without running. This test served as an analogy to the 10-meter walk test used in physical therapy to assess gait. Lap times were recorded with a stopwatch application (Apple iPhone native app), where each new lap began when the subject's right heel struck the ground or when the mechanism returned to its starting position. A chest heart rate belt (Polar H10 Wireless Heartrate Monitor) was worn during the study to measure resting and peak heart rates. Before each experiment, the subject sat still in a chair for 30 seconds to assess a resting baseline, then performed the test in GRAM. They took a 5-minute break, sitting still in a chair, to allow heart rate to return to normal, and then performed the same distance walking. The subject performed five sessions total, one session per day. The subject did three sessions in a row, took a two-day break, then returned for the remaining two sessions.

C. Data Analysis

For each repetition, mean velocity was calculated by dividing repetition time over distance (0.6 meters). Frequency of repetition was calculated as the inverse of repetition time. Total projected repetitions per day were extrapolated by multiplying the mean frequency of repetition by the time it took to perform the test the first day. Within-day improvement was calculated by subtracting the velocity of the last repetition from the time of the first repetition. Between-day improvement was calculated as the difference between the velocity of the first repetition of the day subtracted from the last lap of the prior day. Positive values in either within or between-day improvement corresponded to performance improvements. The learning rate for driving GRAM was assessed via power curves fit with MATLAB's "fit" function (curve fit option 'power1') to the average subject's time per repetition data over the five sessions with:

$$T = BN^{-\alpha} \tag{6.1}$$

B and α are constants where α is the learning rate [14]. B is the baseline, or first repetition time and N is the amount of trials T is the time to finish the task and α is the learning rate. See Chapter 1, section 1.1 for a discussion on motor learning.

Physiological cost index (PCI), an equation that relates the subject's efficiency of locomotion was calculated as:

$$PCI = \left(\frac{Ending\ HeartRate - Starting\ HeartRate}{Velocity}\right) \tag{6.2}$$

Change in heart rate was calculated by subtracting the peak heart rate during exercise from the average value from the baseline measurement for each day. The groups were compared using a linear model to ascertain differences in time to complete repetition, overground velocity, PCI, frequency of repetitions, projected repetitions and within and between-day improvements. There were four terms in this model, a term for day, group (GRAM or natural walking), a term for repetition (1-15 for the 15 repetitions done per day), and an interaction term between-day and group was used. For change in heart rate, a second linear model was used that had three terms, day, group and an interaction term between day and group. Some post-hoc comparisons were done with t-tests.

6.3 RESULTS

A. Learning to Drive a Wheelchair Driven by a Walking Linkage

To compare the difficulty of learning GRAM, we quantified a single subject in learning how to drive GRAM versus their performance in a walking task as a baseline. The subject was able to improve their time to complete 15 repetitions as computed by the linear model for the group, and the interaction term ($p<0.001$, $p<0.001$). On the first session, it took the subject about 27 seconds to perform 15 repetitions in GRAM. On the last day, it took 15.7 seconds to complete 15 repetitions with GRAM (Figure 6.6).

GRAM reported a learning rate of $\alpha=0.36$. The learning rate of GRAM was not statistically different than the mean learning rate found in a sampling of classic learning experiments in unimpaired subjects ($\alpha= 0.28 \pm 0.16$ SD; unpaired t-test, $p =0.68$, Figure 6.7) [13].

The group, day, interaction and repetition term of the velocity curves were significantly different from each other (linear model, $p<0.001$, $p<0.001$, $p=0.002$, $p=0.01$). The mean speed achieved during training increased significantly for both groups across the five sessions of the study (Figure 6.8). GRAM was 35 % slower than the walking task the first day, but by the last day, GRAM was only 17.6% slower (Figure 6.8). The overground speeds of GRAM fall in the range reported for manual wheelchair use of 0.48-0.8 m/s [144].

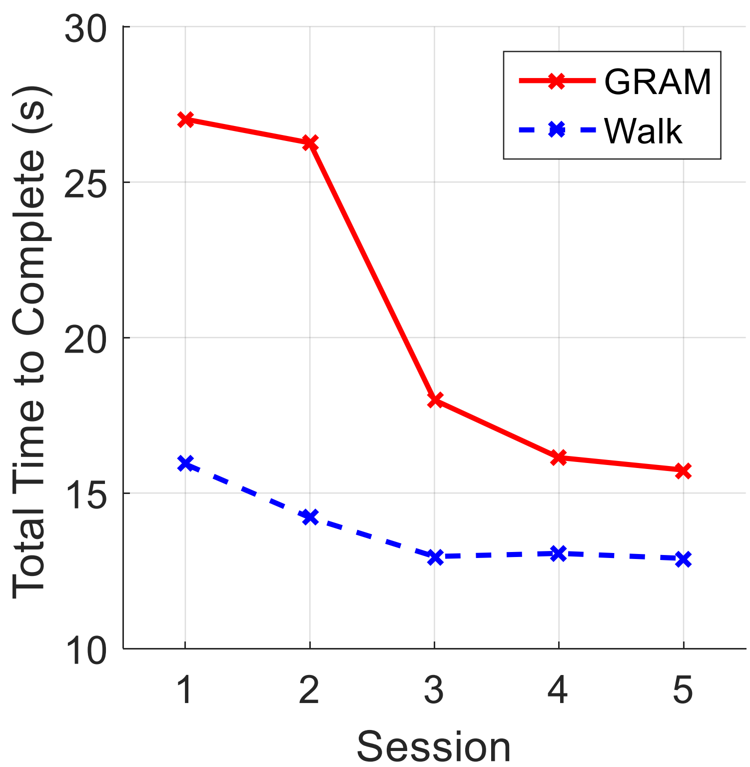

FIGURE 6.6: Total time to complete 15 repetitions. Both groups showed a significant difference in the group and interaction terms.

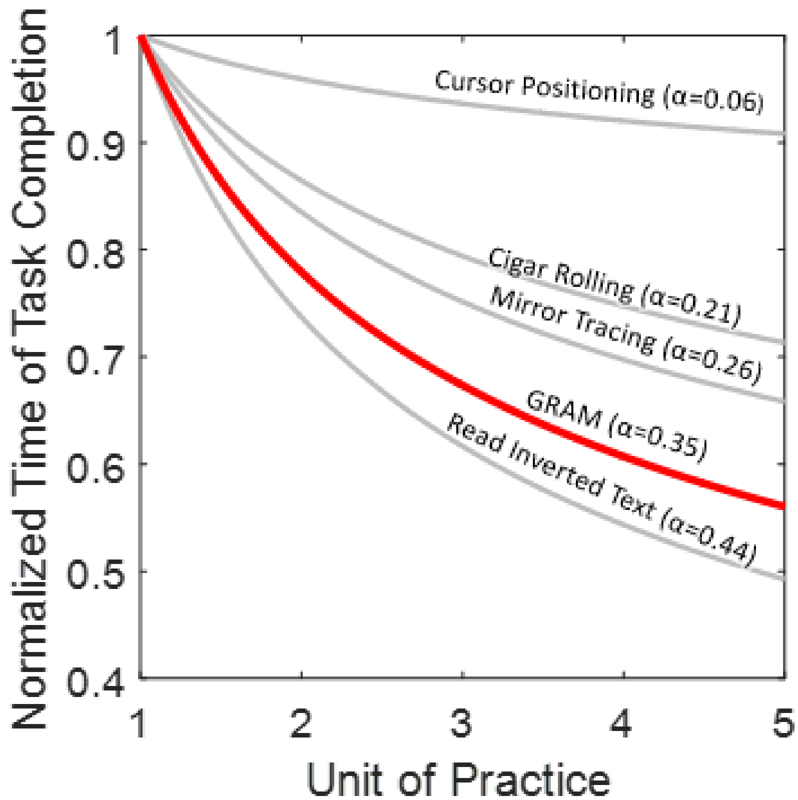

FIGURE 6.7: MOTOR LEARNING CURVE FITS FOR GRAM AND A SAMPLE OF OTHER MOTOR LEARNING STUDIES (GREY)[13] WITH THE SUBJECT'S INDIVIDUAL CURVES FOR GRAM (RED) THE X AND Y-AXIS IS NORMALIZED TO MULTIPLES OF THAT UNIT OF INITIAL PRACTICE, THE TIME REQUIRED TO COMPLETE THE TASK AFTER ONE UNIT OF PRACTICE RESPECTIVELY. THE RATE OF LEARNING FOR GRAM WAS NOT SIGNIFICANTLY DIFFERENT THAN THE LEARNING RATE OF THE MEAN OF THE TWELVE CLASSIC EXPERIMENTS.

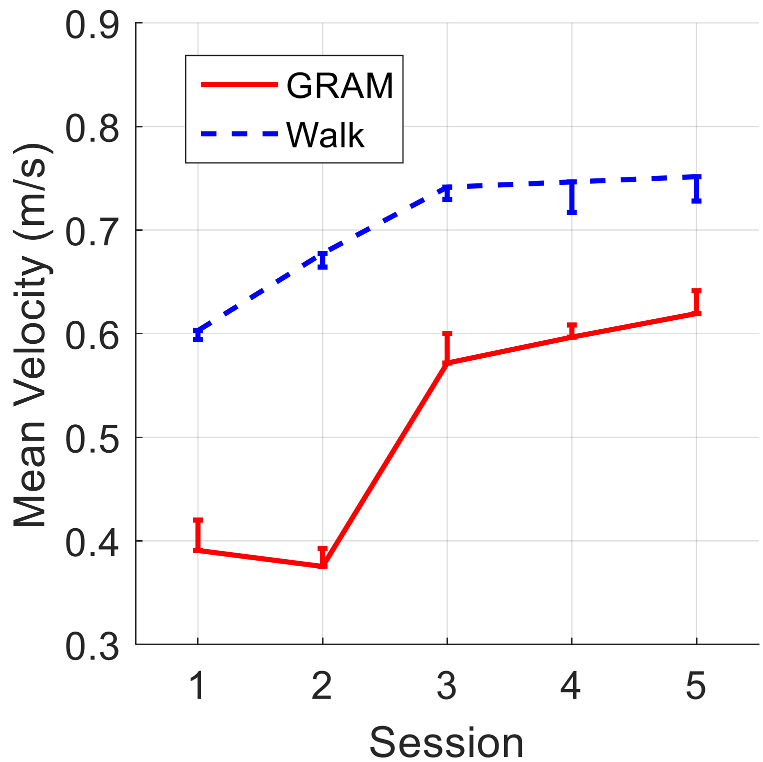

FIGURE 6.8: MEAN VELOCITY FOR GRAM AND THE WALKING TASKS. THERE WAS A SIGNIFICANT CHANGE OVER SESSION, BETWEEN GROUPS AND FOR THE INTERACTION AND REPETITION TERM. GRAM'S SPEED WAS IN THE RANGE OF SPEEDS REPORTED BY THE LITERATURE FOR MANUAL WHEELCHAIR USE (0.48-0.8 M/S) BY DAY 3 AFTER A BREAK WAS TAKEN FOR TWO DAYS [144]. ERROR BARS ARE +/- 1 STANDARD ERROR.

B. Between and Within-Day Improvement

The linear model analysis showed significance for the repetition term which represented within-day improvement ($p<0.001$). This suggests GRAM showed velocity improvement mostly within the session. (Figure 6.9A).

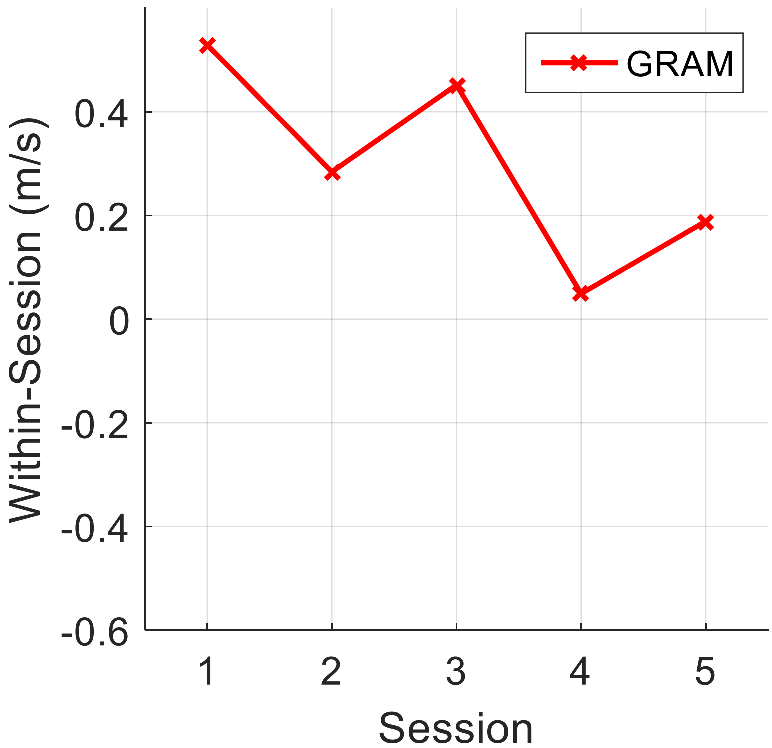

FIGURE 6.9A: WITHIN-DAY IMPROVEMENT. GRAM WAS SIGNIFICANT OVER THE REPETITION TERM. GRAM SAW MOST PERFORMANCE IMPROVEMENTS WITHIN THE SESSION.

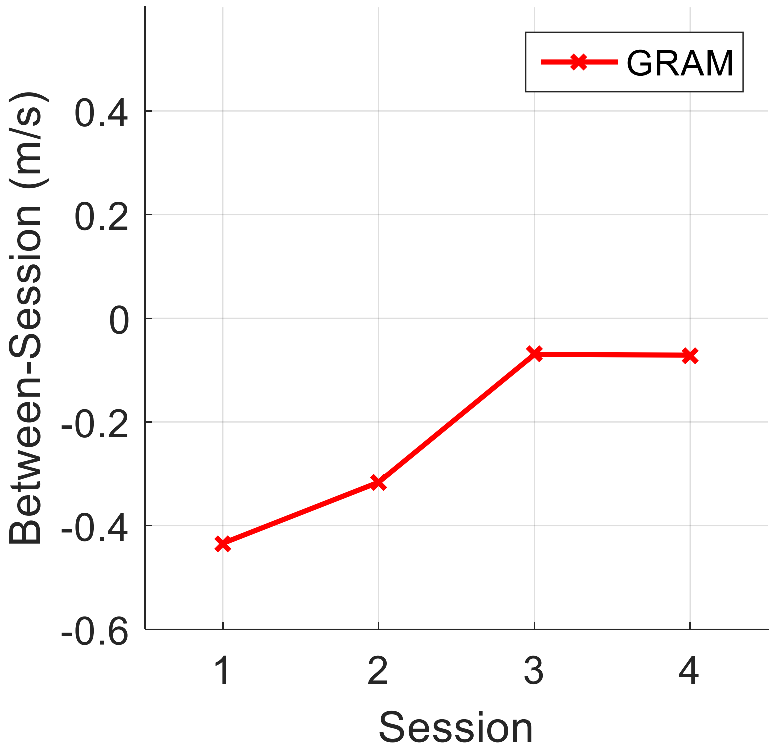

FIGURE 6.9B: BETWEEN-DAY IMPROVEMENT. THERE WAS A SIGNIFICANT DIFFERENCE FOR THE DAY TERM. GRAM DID NOT SEE MOST IMPROVEMENT BETWEEN SESSIONS.

A similar analysis was performed on the between-day improvement. Here, the linear model also showed significance for the day term which represented between-day improvement, ($p = 0.001$). GRAM started each day with a lower velocity then it finished the prior day with, suggesting most of the improvement had to occur within the session instead (Figure 6.9B).

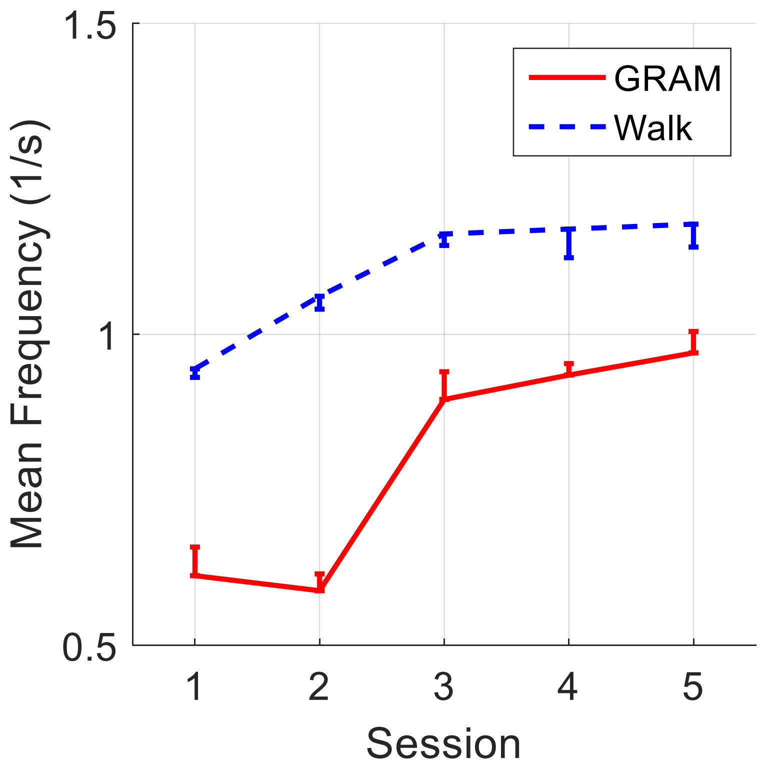

FIGURE 6.10: MEAN FREQUENCY OF REPETITIONS. THERE WAS A SIGNIFICANT DIFFERENCE IN REPETITION FREQUENCY FOR THE GROUP, SESSION, REPETITION AND INTERACTION TERM. ERROR BARS ARE +/- 1 STANDARD ERROR.

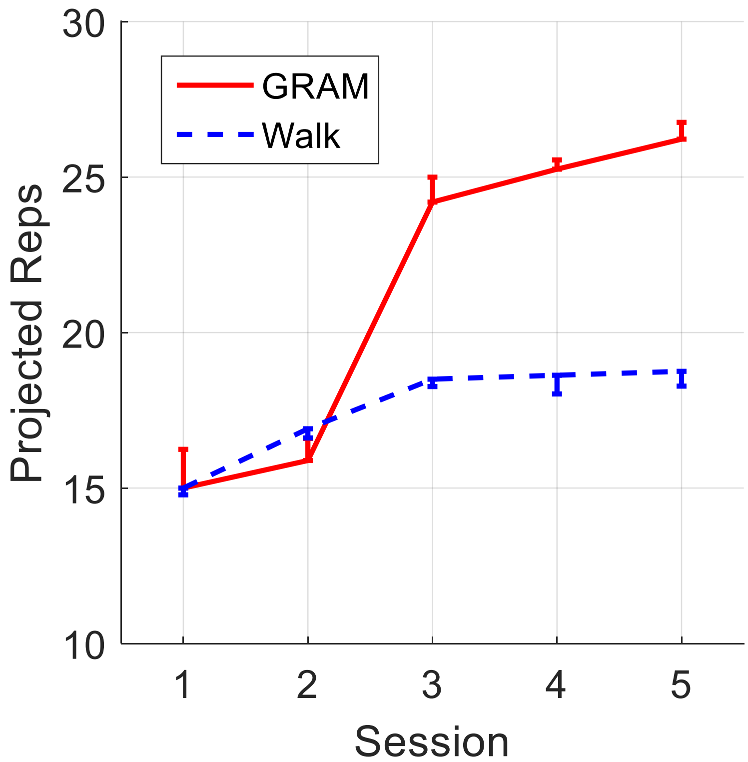

FIGURE 6.11: PROJECTED REPETITIONS OVER TIME, CALCULATED BY MULTIPLYING MEAN REPETITION RATE BY TIME ON THE FIRST TRAINING SESSION TO COMPLETE FIFTEEN REPETITIONS. THERE WAS A SIGNIFICANT CHANGE IN PROJECTED REPETITIONS BETWEEN GROUPS, SESSION AND FOR THE INTERACTION AND REPETITION TERM. BY THE LAST DAY OF THE STUDY, GRAM HAD A PROJECTED REPETITION COUNT OF 26 REPS IF GIVEN THE FULL 27 SECONDS OF THE FIRST DAY TO PERFORM THIS. ERROR BARS ARE +/- 1 STANDARD ERROR.

C. Frequency of Repetitions and Projected Repetitions

The rate of repetition was significantly different between groups and increased over days and for each group (Figure 4, linear model, $p < 0.001$, $p < 0.001$). Similarly, the linear model showed there was significance for the interaction term ($p = 0.002$) and the repetition term ($p = 0.01$). By the end of the study, GRAM had a frequency of 0.9 Hz while the walking task had a frequency of 1.2 Hz. The projected repetitions per day, calculated by multiplying the mean daily repetition frequency by the time it took on the first day to complete the required repetitions, were significantly different between groups, over days, and for the interaction and repetition terms (Figure 6.10, linear model, $p < 0.001$, p<0.001 p=0.002, p=0.01). GRAM started with a repetition count of 15 reps on the first day and ended with a projected repetition count of 26 repetitions. In contrast, the walking task had a projected repetition count of 18.8 reps by the last day (Figure 6.11).

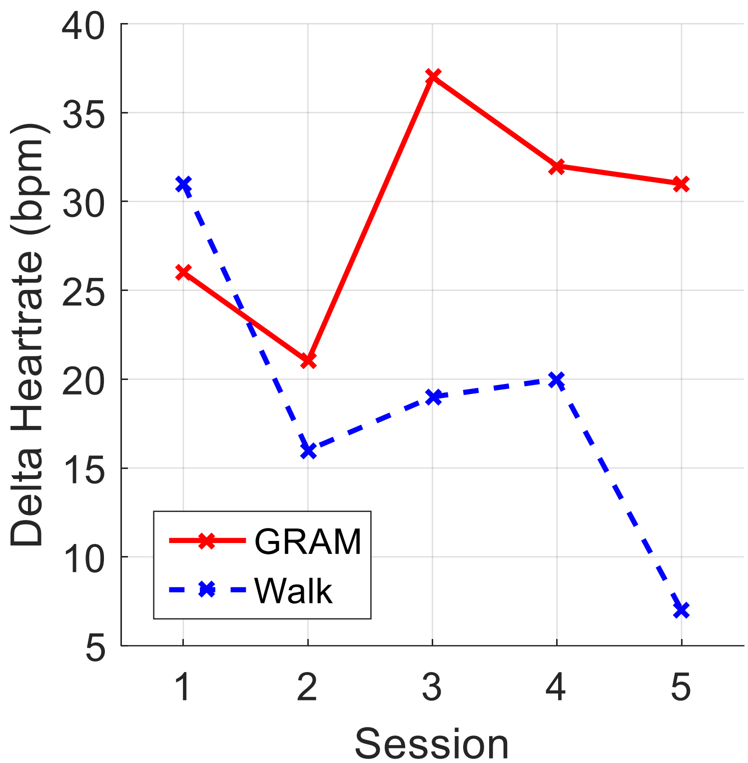

FIGURE 6.12: DELTA HEART RATE. ON THE LAST DAY OF THE STUDY, GRAM HAD A CHANGE IN HEART RATE OF 31 BPM VERSUS THE WALKING TASK WHICH HAD A CHANGE OF ONLY 7 BPM.

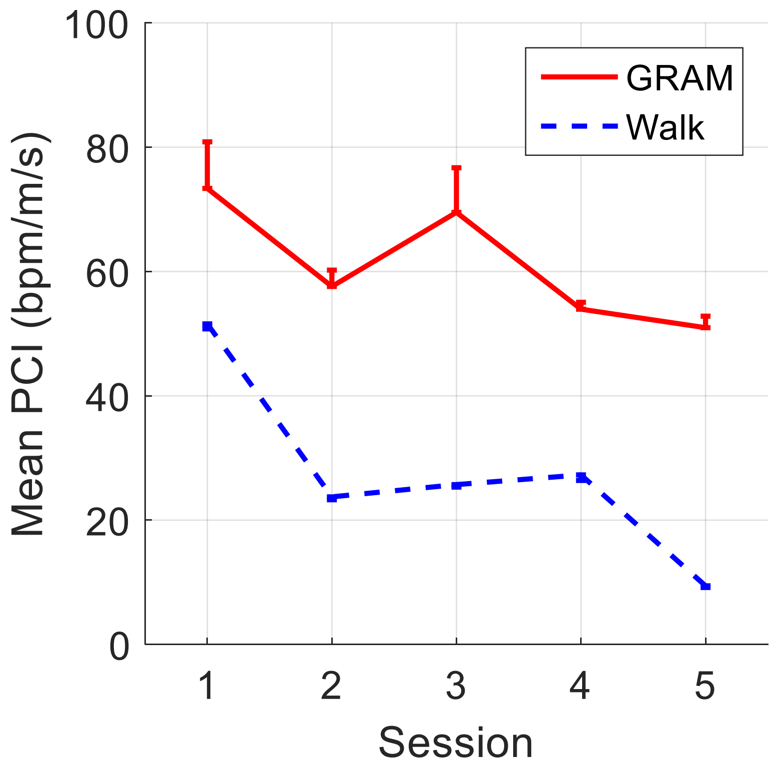

FIGURE 6.13: PHYSIOLOGICAL COST INDEX. THERE WAS A SIGNIFICANT DIFFERENCE BETWEEN GROUPS, AND OVER SESSIONS AND FOR THE REPETITION TERM. THE INTERACTION TERM WAS NOT SIGNIFICANT. ERROR BARS ARE +/- 1 STANDARD ERROR.

D. Changes in Heartrate and Physiological Cost Index

As demonstrated by the linear model, there was not a significant difference for the group or day term, but the interaction term trended to significance for the change in heart rate for both groups (p=0.4, p=0.8, p=0.09) The change in heart rate for the last session was 31 bpm for using GRAM for this subject (Figure 6.12).

PCI was significantly different between groups and over days but was not significantly different for the interaction term (Figure 6.13, linear model p <0.001, p< 0.001, p =0.34). PCI was significantly different for the repetition term (p=0.02). On the first session and last session, GRAM had a PCI that was 143.7 bpm/m/s and 99.8 bpm/m/s. The walking task had a PCI of 101 bpm/m/s on the first day and on the last day a PCI of versus the walking task a PCI of 18.5 bpm/m/s ± 9.9 SD.

6.4 DISCUSSION

We demonstrated the feasibility of walking linkage-driven wheelchair to be used for assistance in a single case study with a young nonimpaired subject. Using regular walking as a control experimental condition, we studied how difficult it was to learn GRAM in a task analogous to the 10-meter walk test used in physical therapy circles. The main results were that 1) the design approach to GRAM is feasible for wheelchair propulsion, and while slower than walking GRAM can move at speeds comparable to the range reported in the literature for manual wheelchair use 2) GRAM had a higher physiological metrics in terms of repetition frequency and projected reps then the control. GRAM also had a higher PCI metric despite slower speeds or was less efficient than the walking task and for this subject, and 3) required practice within the session to improve velocities, versus resting between sessions.

A. Overground Speeds of GRAM and Physiological Efficiency

GRAM was slower than the walking task, and by the end of the study still lagged a significant 17.5 % behind in over ground velocities. However, the speeds reported on the third day, 0.57 m/s, fall in the range of values reported for over ground manual wheelchair speeds of 0.48-0.8 m/s meaning its performance is comparable to other traditional manual wheelchairs, at least for this subject.

Similarly, GRAM had a significantly larger change in PCI. The differences in PCI decreased as time went on in the study, suggesting part of the exertion came from learning how to coordinate GRAM. Another factor could be also the cardiovascular fitness of the subject in question and we predict PCI and delta heart rate would be a function of both the experience of the subject in the device as well as their prior fitness levels.

The subject was also able to significantly increase their repetition count over time for both groups, leading to a 1.7 increase of projected repetitions from the first day with GRAM, compared to a 1.25 increase of projected repetitions from the walking task.

B. Learning How to Operate GRAM

The learning rate of GRAM was not significantly different from the mean of learning rates reported for unimpaired subjects from [13]. Similarly, the subject was able to increase their over ground speed in GRAM about 36% over the course of 5 days, compared to their increase of walking speeds of about 15% over the study. This suggests that GRAM, like walking, is a motor skill requiring time to learn. Considering the novelty of performing a walking motion from a seated position, versus an able-bodied individual learning to master performance on a walking task, it is not surprising GRAM had a greater rate of improvement, suggesting there was more to learn.

Interestingly, most of the improvement for GRAM happened within the session for this subject which would suggest the motor task was not complex since sleep was not required to consolidate learning [104]. Possible explanations for this are the current mechanical setup of GRAM could require excess force before the linkage is in motion, masking the speed improvements one may observe with a system that could provide assistance. Another related possibility is that GRAM could have over fatigued the subject, and an experimental protocol that allowed a day off between sessions for the leg muscles to completely recover may have unmasked between session learning. Based on these preliminary results however prolonged training sessions are recommended to master GRAM at least for this subject.

C. Study Limitations and Further Research

The main limitation of this work is the small sample size. More participants of various age and fitness level are needed to fully assess the difficulty in learning GRAM and these results are strictly generalizable currently to only this subject. Similarly, the effects of coordinating a wheelchair equipped with dual GRAM units was not investigated in this first proof of concept; future studies should observe how coordinating both legs to learn GRAM could affect overground speeds and repetitions. Furthermore, our study would need to be extended to individuals with leg impairments like stroke survivors to investigate the potential benefits of this type of wheelchair-based leg exercise. Future studies should also look into using V02 measurements to assess physical effects since heart rate has limitations and is affected by factors like subject fitness and medications.

GRAM should also be compared to other types of leg-propelled wheelchair exercise devices to ascertain if differences in the metrics we studied were due to our propulsion mechanism. For example, some wheelchairs made for individuals with stroke utilize a pedaling mechanism; and practicing pedaling has been shown to be correlated with increased walking speeds in stroke patients. However these devices do promote greater practice in dysfunctional muscle synergies instead of relearning the synergies of walking [139]. It would be interesting to assess then if a walking motion from a wheelchair could give a similar or greater benefit then these other mechanisms with less dysfunctional synergies.

A next direction for future research would be examining the use of GRAM with more individuals with walking impairment, like stroke. Past work from our group has already demonstrated the potential of wheelchair-based therapy in the upper extremity with a lever-drive wheelchair for individuals with stroke [85], [118]. Increasing exercise opportunity is especially important for individuals with stroke since typically they are taught to drive wheelchairs using only their unimpaired side, contributing to further disuse. In general, it has been shown that individuals with stroke do not receive optimal amounts of exercise repetitions for their upper or lower limbs [105].

In the context of stroke rehabilitation specifically, GRAM could potentially serve as a way to promote greater use of the impaired limb, as by the last day the subject nearly doubled their projected repetitions for their brief bout of exercise. Similarly, lower PCI is desirable for rehabilitation since cardiopulmonary exercise elicits motor recovery after stroke [106]. For example, this subject had a mean change in heart rate of 29.4 bpm, which demonstrates the feasibility of GRAM as an exercise device since it put the subject in their target zone for cardiovascular exercise. Further, GRAM was a task that could be learned and improved upon within the session, providing a possible way to motivate the individual and minimize frustration. This could possibly increase exercise compliance, a core issue facing stroke rehabilitation. Further research studies should investigate the potential of GRAM to play a key role in increasing walking therapy accessibility and mobility to disabled individuals like those with stroke.

CHAPTER 7: SUMMARY OF THE PRIMARY FINDINGS OF THIS DISSERTATION

7.1 REVIEW OF DISSERTATION ACHIEVEMENTS

The primary contribution of this research was to elucidate the efficacy of a new class of device for assistance and therapy after stroke called exochairs. First, I experimentally validated a previously developed upper extremity exochair called LARA in a cohort of young unimpaired individuals to assess motor learning of two versions of the device: one that used a simple one-way bearing lever design and the other that used yoked hand-clutching to add the ability to back up. I found that the yoked hand-clutching providing greater maneuverability and physiological effect. I also found that yoked hand-clutching was a complex motor task since most performance improvements occurred overnight for subjects. These factors made the yoked hand clutch version of LARA more appropriate for application in patients with stroke.

I validated this hypothesis by conducting a study involving 5 chronic stroke survivors. I found that all subjects were able to learn how to drive the chair with person-specific strategies, and experienced a 3-fold wheelchair speed increase over the course of study. I found that the rate of learning of the stroke cohort was 2 times greater then what was seen previously in the unimpaired control group, and from a sampling of classic motor learning experiments from the literature. I also found that despite being a more mechanically simple device, LARA was able to provide clinical improvements that were comparable to more complex robotic devices.

Motivated by three subjects from the previous study who had difficulty learning how to coordinate clutching, I then investigated the use of machine learning algorithms to provide a non-heuristic way to add assistance to these drivers in a feasibility study. I explored a breath of paradigms settling on a Weighted KNN algorithm that I validated on an unimpaired subject to 85%

online accuracy, with 90% offline accuracy. I compared this approach to using a simple switch to relieve the complexity of clutching and discussed implications regarding balancing device complexity and cost with improved physical accessibility for rehabilitation robotics.

Next, I focused on the design and testing on a low-cost hand exoskeleton to be used in the home to improve hand extension after stroke. I showed preliminary feasibility in an unimpaired subject to produce necessary hand forces to promote greater hand use after stroke.

Finally, the culmination of this work was in Chapter 6, where I detailed the design and evaluation of a lower-body exochair called GRAM for walking impairment after stroke. I discussed the rationale behind using a novel single DOF six-bar linkage mechanism to lower device complexity and cost while at the same time providing a salient walking motion to aid recovery. I demonstrated feasibility with a single unimpaired subject using normal walking as a control over a test analogous to the 10-meter walking test used in physical therapy. I presented data that showed GRAM was a learnable motor skill for this subject. It had similar overground speeds to traditional manual wheelchairs, and it provided a physiological effect similar to moderate cardiovascular exercise. This provided evidence of its feasibility as a potential therapeutic and assistive tool.

After injury stroke patients are often taught to operate wheelchairs using their unimpaired side, contributing to disuse of their impaired limb and limiting recovery. We developed LARA as a way to involve the impaired limb during a time where individuals otherwise would not have such an opportunity – wheelchair use. With design iterations we were faced with a decision between utilizing a simpler one-way bearing operated LARA versus one that used yoked hand clutching to drive. With increased mobility from the yoked hand clutching, but seemingly increased difficulty in operation, the primary contribution of Chapter 2 was elucidating that yoked hand-clutching was a more suitable fit for stroke recovery. The yoked hand-clutch version of LARA did not move at significantly slower speeds than the one-way bearing version of LARA. Similarly, results showed yoked hand clutching promoted increased physiological effect and was a complex motor skill. These factors further evidenced its suitability for use in stroke therapy as a way to increase therapy opportunity via an engaging physically taxing task that also provided mobility.

In our follow-up study using the yoked hand clutch version of LARA with 5 chronic stroke subjects, the primary contribution was that given an appropriately designed technology like LARA, even chronic patients could “unmask” a latent ability to improve function. Traditionally, clinical scores like FM have been used to assess the efficacy of devices or treatments. However, recent work has suggested that these scores are limited and do not capture the full potential of patients to recover functionality and that there may be differences in the ability for patients with stroke to recover based on their severity [15]. Using motor learning rates as a paradigm, we found a metric that revealed individuals with stroke had a far greater rate of learning, 2 times greater than the average rate measured in unimpaired subjects performing the same task and the mean seen in a sampling of classic motor learning experiments. This never-before-seen result holds heavy implications for future design in robotic therapy – that is, a shift in paradigm towards devices that improve motor learning versus clinical measures may lead to more real-world functionality. Consequently, as a second contribution, we demonstrated that our device had a comparable increase in clinical metrics to more complicated robotic devices, despite short training times and being mechanically much simpler than these devices.

We used a corollary from these results to inform our next study, which was the application of machine learning to create an automated clutching system to increase accessibility of LARA to more impaired patients. We found in our previous study that 3 out of 5 of our subjects had difficulty with the coordination required with yoked hand clutching, and thus a grip-free clutching could provide the increased mobility of yoked hand clutching while regressing the skill to an appropriate difficulty level. Because we observed a breath of driver-specific strategies to operating LARA we sought a data-driven method to provide greater robustness for the array of different ways to drive LARA. The primary contribution of this work was as follows: first, we demonstrated that an unimpaired subject could successfully operate a lever-drive wheelchair with a machine learning based automated clutching system, a first application of machine learning in this way as far as we know. Second, we found that the subject experienced a significant increase in algorithm performance over their brief practice each day as they learned the system, and this rate of learning was comparable to healthy subjects learning other tasks. Lastly, we found that the automated LARA had an offline accuracy 90%, and online accuracy of 85%, evidencing its potential as an appropriate regression for more highly disabled drivers.

The primary contribution for RHED was the novel approach to its design and control. By utilizing the fact that stroke survivors can easily clasp their impaired hand but not extend, we were able to minimize our design requirements to a device that just focused on assisting hand extension. This translated to a tool that was more lightweight and therefore cheaper, so it would be more usable in a home environment. The use of a magnetometer to control the extension of the impaired hand also is a novel contribution since it avoids the issues seen in other control approaches used in previous hand exoskeleton design. For example, EMG control has been used in the past, and has been shown to possibly reinforce dysfunctional muscle synergies. RHED demonstrates the central theme of this dissertation, which is maximizing function and accessibility through innovative design.

Finally, the culmination of the work was GRAM which took the lessons learned from

LARA and applied them to a novel lower body exochair. Previous work in gait rehabilitation has consistently reiterated that movement recovery requires a consistent practice of a walking motion to reactivate spinal circuits with naturalistic loading and feedback [137]. Whereas the feasibility of pedal-driven wheelchairs has been shown for stroke lower body rehabilitation, our approach utilized a unique walking mechanism to provide a more task-specific approach to therapy [103], [143]. The primary contributions of this work were 1) demonstrating GRAM's feasibility in an unimpaired subject; the subject could ambulate in GRAM with overground velocities comparable to typical manual wheelchair propulsion 2) demonstrate that GRAM can provide cardiovascular exercise needed to promote motor rehabilitation and 3) elucidate GRAM is a motor skill that can be improved upon with practice within the training session, evidencing its potential as a tool that can improve patient compliance in therapy.

Although the study on GRAM was a single case report, the results taken holistically in the context of the rest of the dissertation are encouraging evidence and suggest further development of exochairs as a new paradigm in therapy and mobility.

7.2 SUMMARY OF THE CONTRIBUTIONS

Previous work with LARA showed the therapeutic potential of a predecessor stationary version of the device, called RAE, in stroke subjects. For example, one study showed after only eight 45-minute sessions with the device, there were significant improvements in FM score (8.5 ± 4 pts increase, $p = 0.009$, prior FM score: 17 ± 8 out of 66) and in active-range-of-motion (AROM) for the arm ($66\% \pm 20\%$ increase, $p = 0.003$) which were sustained at a 3-week follow-up [82]. Another study also showed a 1.7 factor increase in AROM with a small cohort of stroke subjects, indicating the potential of RAE as a device to promote therapy [84]. Similarly, previous literature showed lever drive chairs lower physical strain and improve efficiency compared to push-rim propulsion, making them attractive for use in stroke, but lack maneuverability [63], [66], [68], [69], [100]. For the first time, this dissertation showed a solution to this problem, the clutch LARA, was most suited for application in stroke therapy in a comparative performance study of two LARA models in healthy young adults. The yoked hand clutch version of LARA was more maneuverable (able to back up, turn in place), had a significantly greater change in heart rate (Last session: yoked hand clutch had a reading of 23.9 beats per minute (bpm) ± SD 12.8 bpm, vs one-way bearing, 15.5 bpm ± SD 7.1 bpm, p=0.007), and physiological effect (Last session: yoked hand clutch had a PCI of 31.7 bpm/m/s ± 25.0 SD, vs one-way bearing, PCI of 18.8 bpm/m/s ± 9.9 SD, p=0.02). The yoked hand clutch version of LARA was also a complex motor skill (improvements occurred overnight as opposed to within the session as seen in the one-way bearing). As a comparison to the brief literature on exercise wheelchairs, a study using 10 unimpaired males driving a wheelchair operated by one hand and leg compared to a traditional wheelchair found the unilateral wheelchair produced significantly lower heart rate values, oxygen consumption and carbon dioxide levels. Specific numbers were not reported however, so a direct comparison to LARA cannot be made [81]. Overall, this contribution meant the yoked hand clutch version of LARA was more suitable for potential application in stroke patients.

In addition to the previous work with RAE in stroke subjects, the literature also showed the one-way bearing LARA could be driven by stroke subjects in a small pilot study using four stroke subjects. This study showed drivers could operate LARA over 10 trials of a 3.3-meter distance, and on the last trial the average over ground speed was 0.1 m/s [113]. This dissertation showed stroke subjects could also drive the clutch LARA over a 14-meter figure-8 track, and on their last day they reached an average speed of 0.15 m/s. This dissertation also found subjects experienced a change in heart rate that put them in the zone for cardiovascular exercise (mean change 36.4 bpm on session 6, $p = 0.02$). In comparison to the unimpaired clutch group from Chapter 2, this group had a change in heart rate of 23.9 beats per minute (bpm) ± SD 12.8 bpm on their last day. The stroke drivers had a mean learning rate twice that of healthy subjects taken from the literature (α=0.62 versus α= 0.28 ± 0.16 SD), and also had clinical outcomes comparable to more complicated robotic devices (2.2 points ± 1.1 SD change in Fugl-Meyer and 2.2 blocks ± 2.5 SD change in Box and Blocks score) [13]. Additionally, it is often assumed people with a stroke who have a severely paretic are unable to learn novel, skilled behaviors that incorporate use of that arm. This dissertation showed that a group of people with chronic stroke learned to use their impaired arm to propel the yoked-clutch lever drive wheelchair. Over six daily training sessions, each involving about 134 training movements with their "useless" arm, the users gradually achieved a 3-fold increase in wheelchair speed on average, with a 4-6 fold increase for three of the participants. These results suggest that appropriately-designed assistive technologies can unleash a powerful, latent ability for motor learning even for severely paretic arms. These results also validated the potential of the yoked hand clutch version of LARA as seen in Chapter 2.

Previous literature had developed limited ways for people with hand impairment to propel wheelchairs, such as power-assisted push rim wheelchairs or lever drive chairs with simplified hand controls, like the yoked clutch LARA in Chapter 3 [66], [64], [118]. This dissertation showed for the first time that a person could learn to manually propel a lever drive chair when aided by a clutching control system that identified desired clutch states from sensors mounted on the levers, and this was seen in the following results. 1) The driver was able to successfully navigate around an environment representative of a daily wheelchair use. 2) The performance of the system was over 90% accuracy offline, and 85% accuracy online. Typically, it is not useful to compare accuracy rates across applications, since what qualifies as appropriate accuracy is dependent on the use case [120]. However, to give intuitive understanding of this performance as compared to the literature, a group from Carnegie Mellon used the KNN algorithm with three accelerometer sensors on a wheelchair for an analogous purpose (classifying what propulsion style a healthy person operating a manual wheelchair was using to later provide cueing, similar to how this algorithm classified when to brake in the cases of going forward, backing up, etc.) and found an overall accuracy rate ranging from 60-90% depending on to amount of sensors used and environment the chair was driven in [145] . This indicates the LARA machine learning control system had a reasonable performance rate, and could be a potential tool to promote therapy.

Previous work in this dissertation showed the potential of the LARA exochair to promote exercise and mobility in healthy and impaired subjects. This dissertation showed the concept of LARA could be extended to a novel device called GRAM for the lower limb with a prototype and its associated performance in a healthy subject. This dissertation showed a healthy subject was able to learn the GRAM exochair and ambulate at speeds within the range of overground velocities for a traditional manual wheelchair (starting on session 3 with 0.57 m/s, and reaching 0.62 m/s on last session, compared to a range of 0.48-0.8 m/s from the literature) [144]. In comparison to the healthy drivers learning LARA, the yoked clutch drivers had a mean overground velocity of 0.71 m/s on their last day, and the one-way bearing drivers had a mean velocity of 0.74 m/s. This subject also experienced a significant increase of repetition frequency, reaching about 1Hz by the end of the study. In context of gait therapy for stroke patients, it was found that in a typical 36-minute therapy session, the average amount of steps taken was roughly 300 [138]. Therefore, it is not unreasonable to suggest that GRAM could potentially achieve a similar amount of exercise repetitions within that time frame, with the additional safety that the patient cannot fall over in GRAM due to its design. These results indicate the potential of extending the exochair concept via GRAM to the lower body to promote therapy after stroke.■

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APPENDIX: A

Select Clinical Measures Taken As Is From Therapy Handbook

Subject ID: ___________________________

Date: ___________________________

Exam#: ___________________________

<u>Box & Blocks</u>

Right	Left

<u>Box and Blocks Test</u>

- Open test box and place the divider between the two compartments
- Position the box lengthwise along the edge of a standard height table with compartment with cubes on same side as the subject's dominant hand
- Seat the subject in a standard height chair facing the box
- The examiner sits facing the subject to view blocks being transported and note discrepancies in technique
- "I want to see how quickly you can pick up one block at a time with your right/left hand (point to hand), carry it to the other side of the box and drop it. Make sure your fingertips cross the partition. Watch me while I show you how."
- Examiner transports three cubes over partition

- "If you pick up two blocks at a time, they will count as one. If you drop one on the floor or table after you have carried it across, it will still be counted, so do not waste time picking it up. If you toss the blocks without your fingers crossing the partition, they will not be counted. Before you start, you will have a chance to practice for 15 seconds. Do you have any questions? Place your hands on the sides of the box. When it is time to start, I will say 'READY' and then 'GO.'"
- Perform a 15 second practice. If mistakes are made, correct them.
- "This will be the actual test. The instructions are the same. Work as quickly as you can. READY…(wait 3 seconds) GO."
- After one minute STOP
- Count the number of blocks transported and record. Subtract blocks transported more than one at a time or if the fingertips did not cross over
- Turn the box so all the blocks are on the same side as the next hand to be tested
- "Now you are to do the same thing with your left hand. First you can practice. Put your hands on the sides of the box as before. Pick up one block at a time with your hand and drop it on the other side of the box. READY…(wait 3 seconds) GO."

Subject ID: ___________________________

Date: ___________________________

Exam#: ___________________________

Modified Ashworth Scale of Spasticity

After Katz et al, Arch PMR 73:339-347, 1992

Shoulder	Elbow	Wrist	Finger

0 No increase in muscle tone.

1 Slight increase in tone, manifested by a catch and release or by minimal resistance at the end of the range of motion when the affected part(s) is moved in flexion or extension.

2 Slight increase in muscle tone, manifested by a catch, followed by minimal resistance throughout the remainder (less than half) of the ROM, but affected part(s) easily moved.

3 More marked increase in muscle tone through most of the ROM, but affected part(s) easily moved.

4 Considerable increase in muscle tone, passive movement difficult.

5 Affected part(s) rigid in flexion or extension

For elbow, go from full flexion to full extension in 1 second with patient supine and arm stop a pad (per Bohannon and Smith, Phys Ther 1987).

MOTOR ACTIVITY LOG

Task	Use?	AS	HW	If no, why?
1.Turn on light switch				
2.Open drawer				
3.Remove clothing from drawer				
4.Pick up phone				
5.Wipe off counter				
6.Get out of car				
7.Open refrigerator				
8.Open door with doorknob				
9.Use TV remote				
10. Wash your hands				
11. Turn on/off water faucet				
12. Dry your hands				
13. Put on socks				
14. Take off socks				
15. Put on shoes				
16. Take off shoes				
17. Get up from chair with armrests				
18. Pull chair away from table				
19. Pull chair towards table				
20. Pick up glass/bottle/cup/can				
21. Brush your teeth				

22. Put makeup/lotion/cream on face				
23. Use key to unlock door				
24. Write on paper				
25. Carry object in hand				
26. Use fork or spoon				
27. Comb your hair				
28. Pick up cup by handle				
29. Button a shirt				
30. Eat half sandwich or finger food				
<u>MEAN SCORE</u>				

1. Did you perform this activity during the past week? Yes or No (if yes, continue)

2. How much did your affected arm participate in this activity (AOU-Amount of Use)

   0 (Never/ Not at All) 5 (Always During/All the Time)

3. How well did your affected arm help during this activity? (QOM – Quality of Movement)

4. 0 (Inability to use the affected arm for this activity)

   5 (Ability to use the affected arm just as well as before the stroke)

Subject ID: _______________
Date: ____________________
Exam #: _________________

Intrinsic Motivation Inventory

For each of the following statements, please indicate how true it is for you, using the following scale:

Not at all True			Somewhat True			Very True
1	2	3	4	5	6	7

1. I believe doing this activity could be beneficial to me.
2. I put a lot of effort into this.
3. I think I am pretty good at this activity.
4. I was anxious while working on this task.
5. After working at this activity for a while, I felt pretty competent.
6. This activity did not hold my attention at all.
7. I did this activity because I wanted to.
8. I am satisfied with my performance at this task.
9. I think this is an important activity.
10. This activity was fun to do.