Manual Wheelchair Propulsion Training

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Manual wheelchair users are at high risk of developing upper limb pain and injury. While much has been published identifying the prevalence of upper limb pain, very little has been published on its treatment and prevention. Consequently, a propulsion training system was developed based on biomechanical, ergonomic, and motor learning theory principles. Three groups were compared: a control group (CG) that received no training, an instruction only group (IO) that reviewed a multi media instructional presentation (MMP), and a feedback group (FB) that reviewed the MMP and received additional real time feedback (RTF). The purpose of this study was to 1) Develop propulsion-training programs that minimized injurious biomechanics; 2) Test if the training programs can cause lasting changes; 3) Investigate if resultant forces and moments at the shoulder can be reduced and 4) To determine if one treatment (MMP) was superior to the other (RTF) in achieving these goals. First, the RTF systems' design was completed and tested on a pilot subject (chapter 2). Next the training systems were tested over ground (chapter 3) and on a dynamometer where shoulder forces were modeled (chapter 4) (N=27). Results showed baseline pain measures to be extremely low and did not increase significantly ($p>.2$). In addition, the effects of training were not influenced by surface type or speed condition (presence or absence of a target speed). In chapter 2, the FB group who received RTF and MMP displayed larger increases in contact angle(CA)(angle along the arc of the hand rim) and greater decreases in rate of rise of peak resultant force (rorFr) than the IO group who received the MMP alone ($p<.05$). While both training groups decreased stroke frequency (SF), the IO group displayed a larger

reduction than the FB group ($p<.05$). Furthermore, both treatment groups showed a short term increase in peak resultant force (maxFr) however their long term values were not significantly greater than baseline and their shoulder forces did not increase significantly ($p>.05$). Finally, the CG showed a long term increase in maxFr at the hand rim ($p<.05$), however their shoulder forces did not increase.

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1.0 INTRODUCTION

For many full time manual wheelchair users, independence depends on the integrity and preservation of their upper limbs. Activities like wheelchair propulsion and transfers can place great demands on the arms. These activities, relied on for independence and community integration can also accelerate the aging process and can lead to the development of pain and injury .[1-5] For example, Lundqvist et al. found that pain was the only factor correlated with lower quality-of-life scores.[6] The impact of upper limb pain can range from limiting one's activities to near total dependence on others. Dalyan et al.[5] determined that of individuals experiencing upper limb pain, 26% needed additional help with activities of daily living and 28% reported limitations on independence. Gerhart et al.[1] found that upper limb pain was a major reason for functional decline in individuals with SCI who required more physical assistance since their injury.

While much has been published identifying the prevalence of upper limb pain in individuals with SCI, very little has been published on its treatment and even less on its prevention. Subbarao et al.[7] found that individuals with SCI and upper limb pain did not get relief from the majority of treatments. They believed that treatment ineffectiveness could be explained, in part, by the fact that primary contributing factors to upper limb pain, wheelchair propulsion and transfers, could not be avoided. The authors concluded that, "future research should be focused upon new methods of wheelchair propulsion and transfer techniques that lessen stress and cumulative trauma on the wrist and shoulders." Until recently it was unclear what recommendations to pass onto users of manual wheelchairs in terms of preventive strategies. However, recent wheelchair research combined with ergonomic principles have lead to the creation of specific recommendations.[8] These recommendations appear as part of a Clinical Practice Guideline development by the Consortium for Spinal Cord Injury which underwent extensive review. The training programs developed as part of this study closely follow the clinical practice guideline recommendations.

Task performance modification based on ergonomic analysis has been proven to reduce the incidence of pain and cumulative trauma disorders of the upper limb in various work settings.[9-14] It is our belief that the pain and injury associated with wheelchair use could be reduced if individuals were taught how to propel their wheelchairs based on sound scientific evidence. The realization of these specific aims allowed us to determine how effectively scientifically-based manual wheelchair propulsion training impacts the stroke biomechanics of wheelchair users. It was our hope that prevention of pain in wheelchair users will have profound impact on people living with disability, increasing quality of life and decreasing healthcare costs associated with secondary injury.

1.1 ERGONOMICS LITERATURE

Although the number of studies linking propulsion mechanics of manual wheelchair users (MWU) may be small, the ergonomics literature provides a strong basis for our training intervention. There have been three large evidence-based reviews describing the link between repetitive tasks and upper limb injury. In 1997 the National Institute of Occupational Safety and Health (NIOSH) reviewed the epidemiologic evidence of this link.[15] In 1999, the National Research Council (NRC) published *Work-related musculoskeletal disorders: a review of the evidence*.[16] In 2001, the NRC, together with the Institute of Medicine, completed a review entitled *Musculoskeletal Disorders and the Workplace*.[17] These comprehensive reviews have found strong links between work activities, physiologic and psychophysical condition at work, and injury.

One conclusion from these comprehensive reviews and from the ergonomics literature is that limiting the frequency of repetitive tasks should reduce the risk of injury. In fact, a number of studies have strongly implicated the frequency of task completion as a risk factor for repetitive strain injury and/or pain at the shoulder.[18,19,20] Although the majority of studies are correlative and do not prove cause-and-effect relationships, longitudinal work has found similar results.[21] These longitudinal studies provide stronger evidence of causation. It should be noted that frequency of a task has been defined differently in each study. However, wheelchair propulsion, with a stroke occurring approximately once per second, would exceed what the majority of studies consider to be a frequent task.

Another conclusion that can be drawn from the reviews and the ergonomics literature is that forces should be minimized during wheelchair propulsion. Higher forces have been correlated with injuries and/or pain at both the wrist[22,23,24] and shoulder.[25,26] Longitudinal studies have also found that higher loads or high-force work predicts risk of development of pain or injury.[27,28,29] It is important to note however that the forces considered to be "high" in the majority of these studies are regularly exceeded during wheelchair propulsion, on a daily basis.[30] For example, one study defined high force as 39 Newtons (N),[24] while another study related high force to lifting a tool that weighed only 9.8 Newtons.[22] Wheelchair users propel with forces routinely over 70 Newtons.[31] Yet another study in the able bodied population noted that pulling or pushing a mass over 50 Kg was related to shoulder pain.[32] Clearly, the average wheelchair user weighs more than 50 Kg (110 pounds).

Based on the conclusions drawn from ergonomics literature as well as motor learning literature (see section below) we decided to focus our training and visual feedback system on reducing stroke frequency and increasing contact angle. It was our hope that increasing contact angle while maintaining a constant velocity would cause a reduction in forces at the pushrim, as work remains constant. Contact angle has been included so that while reducing stroke frequency, subjects do not impart large forces over a shorter or more constrained stroke. By increasing stroke length while holding velocity constant it was our hope that subjects would be performing the same work equivalent, however doing so over a longer period of time. This would ultimately reduce the force exerted on the hand rim at a given speed. Although providing visual force feedback to the user was considered as an alternative option, the literature has suggested that encouraging focus on forces can cause unintended negative consequences in stroke mechanics. Specifically, when MWU attempt to reduce resultant forces through real time force feedback, they tend to direct all forces tangentially to the wheel. De Groot et al. has shown that this can result in decreased mechanical efficiency,[33,34] and Rozendaal et al. has shown that redirecting the forces tangentially can increase moments at the shoulder.[35] It was our hope that training MWU to use an increased contact angle would lead to reduced forces without causing other unintended harmful changes.

1.2 WHEELCHAIR PROPULSION BIOMECHANICS

The literature backing up force reduction as a general principle and the specific recommendation to complete a low frequency, long propulsive stroke was recently evaluated as part of a clinical practice guideline. The recommendations were evaluated by leading ergonomics experts, and a contract methodologist separately graded each supporting study. The ergonomics grade received for the recommendations we followed in this study was a “1” meaning “strongly agrees with ergonomics principles”. The overall grade combining the clinical, epidemiological, and ergonomic literature was a “B” meaning “strong support for the recommendation.”[36] Furthermore the guideline, a group effort involving an expert panel of 10, was peer-reviewed by over 30 professional from various disciplines. None of the reviewer disagreed with this recommendation and most were in support of it. Thus the basis for the training to be completed in this study underwent extensive outside expert evaluation. Consequently there was ample reason to believe that there was room to improve wheelchair propulsion technique.

For example, a propulsion study completed by Boninger et al. found that subjects with paraplegia fell into four groups based on contrasting propulsion technique observed and subjects average stroke frequency varied between groups from 1.1 to 1.6 Hz and the contact angle varied from 102 to 134 degrees.[37] Newsam et al. found variations in average stroke frequency between groups based on injury level from 0.9 to 1.1 Hz.[38] Importantly the standard deviation in Newsam’s study was 0.2 Hz in the low paraplegic group. de Groot et al. also found large variations in contact angle and frequency between groups trained to push with a particular propulsion technique.[39] It was our belief that variation between groups based on technique, level, and training indicate that there is ample room for change with wheelchair propulsion training.

Another aspect of wheelchair propulsion relevant to training involves the propulsive stroke and recovery pattern. The recovery pattern is referred to as the path that the hand follows when not on the pushrim. Two separate studies completed at the University of Pittsburgh have investigated the relationship between the recovery pattern and biomechanics.[37,40] These findings showed that a single looping over propulsion pattern (SLOP) was most common. However a particular style referred to as a semicircular (SC) stroke pattern was found to be associated with a lower stroke frequency and the greatest time spent in propulsion relative to recovery. In essence, wheelchair users who followed a SC pattern hit the pushrim less frequently and used more of the pushrim to go the same speed, which is in fact an emphasis or specific aim of this training study. Veeger et al. found that individuals who used a circular propulsion technique (equivalent to the semicircular pattern) were significantly more efficient as well.[41] However recent work by the same group found that when subjects without disability were trained in both the semicircular and the arc style of propulsion, the arc or pumping style was more mechanically efficient (arc = 7.1% vs. semicircular = 6.7%).[42] This same study also found increased stroke frequency with the arc style (range 61 to 70 strokes per minute) compared to the semicircular style (53 to 56 strokes per minute). Given the conflicting findings related to propulsion pattern we felt it premature to classify or label a specific propulsive style with one term. For the purposes of training we decided to refer to a biomechanically correct technique using terms like "taking a long smooth stroke.", comfortably grabbing the push rim, matching the speed of the push rim, and using as much of the push rim as possible to smoothly apply force to the push rim.

It is clear that in addition to wheelchair propulsion, transfers likely impact the risk of injury. It is also true that in addition to propulsion technique, other factors that can be changed are the type of wheelchair and the fit of the wheelchair to the individual, or wheelchair set up. However, it is unclear what transfer techniques are better as work in the area is extremely limited at this point. In addition, wheelchair users are often hesitant to change the set up of their wheelchair which can require special tools and skills. De Groot et al. have shown that wheelchair propulsion technique can be taught to new users through visual feedback.[33,34] However, their study included only individuals without disabilities. It is unclear if and how wheelchair propulsion technique can be taught to long term wheelchair users. Ultimately, if a less injurious propulsion technique can be taught to wheelchair users, it will represent a low-cost, easy to apply intervention that can reduce pain. In the present study, participants used their own wheelchair and set up was not changed in order to isolate the effectiveness of wheelchair propulsion training alone.

1.3 LEVEL OF SPINAL CORD INJURY AND PROPULSION

The design and implementation of an effective propulsion training intervention must take into a account the functional abilities of its participants. For example, literature has shown that level of spinal cord injury can impact propulsion biomechanics, consequently the proposed training intervention was designed to include individuals who we believed could safely and effectively benefit from our specific propulsion technique and training recommendations[43-50]

It has been suggested that an optimized propulsive stroke results from a balance tradeoff. More specifically, the interaction between the mechanical constraints of a wheelchair and the biomechanical abilities of an individual ultimately contribute to the formation of a propulsive stroke pattern [51,52] Therefore, when training or encouraging individuals to use a particular stroke style, it is critical to recognize that individuals with paraplegia and tetraplegia have been shown to display differences in their propulsion biomechanics. [43-45,48,49,53 54,55] Furthermore, additional variations in propulsion biomechanics have been seen amongst individuals with tetraplegia, specifically those above and below cervical level 6. [43-45,48,49,53 56] For example, the functional potential of the wheelchair user often determines the orientation of the push angle on the hand rim during propulsion. Individuals with tetraplegia have been found to position their hands more backward, relative to top-dead-centre, as compared to individuals with paraplegia.[43,49] It has been explained that applying force more backward on the hand rims could be a compensatory strategy for triceps brachialis paralysis which occurs above cervical level 7.[49] Newsam also found that during propulsion individuals with C6 tetraplegia in particular demonstrated greater wrist extension and less forearm pronation then MWU with lower level SCIs.[57] Considerable differences in force application during steady-state wheelchair propulsion have also been demonstrated between individuals with tetraplegia(ITP) and individuals with paraplegia(IWP).[53]Fraction of effective force (FEF) a measure of propulsion efficiency has been found to be lower in individuals with tetraplegia, than in individuals with paraplegia, largely as a consequence of significantly larger inwards directed lateromedial force (Fy) into the pushrim. [53] Friction at the hand rim is necessary to produce the tangential component of FEF and can be generated through hand grasping, wrist moment generation and/or directing the resultant force away from the tangential direction. Individuals with tetraplegia without hand function, tend to rely heavily on lateralmedial force to create friction which can change the biomechanics and efficiency of the propulsive stroke particularly in comparison to individuals with lower level injuries.[43] Therefore, if triceps function is limited or absent(above C7), one's ability to generate friction in a downward or outward direction may be disrupted.[43]

Other studies have indicated biomechanical contrasts in propulsion related to level of SCI as well. Finley et al found that a group of MWU with upper-limb impairment(cervical injuries) propelled with a higher stroke frequency and reduced hand-rim contact time, smaller peak joint angles, joint excursion of the wrist, elbow, and shoulder during the contact phase then those without upper limb involvement.[44] They also propelled with a reduced power output and reduced hand-rim propulsive and resultant forces, moments, and joint compressive forces. [44] It was concluded that these kinematic and kinetic strategies may have been compensatory strategies allowing MWCUs with upper-limb impairment to propel with as much independence as possible. In addition it was suggested that, taking strokes more frequently while applying lower magnitude forces to the push rim may serve to protect from the development of secondary upper-limb pathologies. [44] It is worth noting that these authors found that, participants with C-6 tetraplegia in particular were significantly slower than all other groups for the majority of test conditions.

Newsam et al. found that individuals with C6 level SCI, using manual wheelchairs, at their fastest self selected propulsion velocities were slower than typical community demands, and suggested that their ability to function independently in a manual chair outside the hospital setting should be further explored. [56] Furthermore, Van der Woude et al also found considerable differences with respect to force application during wheelchair propulsion between IWT and IWP. It was suggested that higher medially directed force in IWT may result from a loss of elbow extension strength. [58] The need to apply extra hand rim friction to compensate for the lack of hand grip strength in IWT may also be responsible for the application of larger medially directed force. It was also noticed that IWT tended to decrease stroke angle with a higher load, whereas IWP tended to increase stroke angle. Van der Woude et al explained that the low effectiveness of force application, as well as the different pattern of force application in persons with tetraplegia should be taken into account when developing other wheelchair propelling mechanisms and training programs.[43]

The proposed propulsion training programs have been designed with the aforementioned literature in mind. If a MWU had a spinal cord injury, we decided to exclude participants above the level of cervical 7 for fear of emphasizing biomechanics that were not practical or safe given the functional constraints of their injury levels. More specifically, we felt it was not prudent to encourage individuals lacking adequate triceps innervation to use a larger contact angle with reduced stroke frequency, at velocities representative of those needed on daily basis. Consequently, as we felt that the presence of sufficient triceps strength was one of the critical determining factors for inclusion into this study all participants with tetraplegia received manual muscle testing to confirm normal triceps function.

1.4 MOTOR LEARNING THEORY

Another question critical to the design of an effective training intervention relates to the way in which people learn and synthesize information. Numerous theories exist within the realm of motor learning that must not be overlooked when developing such an intervention. Careful thought must be put into the design of practice schedules if they are to promote learning, acquisition, retention, focus of attention and transfer of motor skills. Shea and Morgan demonstrated that ordering of motor skills during practice affects immediate performance and retention while the manipulation of practice schedules creates an empirical phenomenon termed contextual interference (CI). [59] The CI effect is however often considered a performance paradox because the increase in interference caused by random practice schedule deteriorates acquisition performance while enhancing retention and/or transfer [60-63] [64]An explanation for the contextual interference effect is that random practice encourages a learner to compare and contrast the methods and strategies used for performing different tasks. Switching between tasks during practice provides the learner with better contrastive knowledge than the repetitive practice that occurs under a blocked or drill like order. This contrast between tasks makes learning each task more distinctive and memorable, resulting in retention. Thus, if the main goal is to maximize long term learning effects, which is one aim of this study, one would conclude random practice condition is preferable over blocked practice.[60,65-68]

Magill and Hall pointed out that various factors tend to interact with CI.[67] Some of these factors include the ecological validity of an experiment (which refers to the extent to which the findings can be generalised beyond a present learning activity or scenario), age, gender, experience level of the learner, the type of skill, task difficulty, and the absence or presence of knowledge of results (KR) during practice trials. The classical view of KR holds that it is an essential source of information that directs a learner towards a more accurate performance of a goal-directed action. In contrast, practice without KR allows performance to drift away from the goal, weakening the representation of an action in memory.[69-74] For the purposes of this study, knowing push angle during wheelchair propulsion, provides essential information to the learner enabling pursuit of forward motion of the chair which is ultimately a goal.

A final component of motor learning relevant to the design of this training program centers on the concept of focus of attention. It has been proposed that the effectiveness of instructions in motor skill learning depend largely on the focus of attention they induce.[74-77] The advantages of an external focus have been attributed to performers' use of more automatic control processes when attending to the movement effect than when attending to the actual movements. [77,78] Literature has suggested that when individuals engaged in a motor learning task concentrating on movements themselves, performers tend to actively intervene in the control processes, resulting in degraded performance and learning. The advantages of focusing on the outcome of one's movements might not only be important with respect to the instructions provided but might also have implications for the feedback given to the learner. In fact, the results of a study by Shea & Wulf (1999) suggested that feedback can be more effective if it directs the performer's attention away from his or her own movements and to the effects of those movements.[79] Therefore, a subject seeing their wheelchair propulsion arm movement pattern in real time may induce a more internal focus where as viewing push angle and cadence assigns attention away from one's own movements and more to the effects of those movements which may intern, facilitate a more automatic control process and enhance retention of optimal propulsion biomechanics.

1.5 DOSING AND TIMING

The dosing and timing parameters incorporated into our propulsion training intervention have been constructed to be practical while maximizing effect and minimizing fatigue and subject drop out. This practice schedule has been designed deliberately to provide less practice frequency because if one is to isolate changes in propulsion technique, physiological adaptations as a consequence of training need to be excluded. [34] Therefore, the learning protocol has to be performed at low intensity and duration, and with a limited frequency while still offering sufficient practice time to promote motor learning. [34] Motor learning literature tells us that in the early stages of learning (initial practice sessions), rapid improvements are seen, followed by consolidation and then weeks to months later, learning plateaus which results in few declines even without continued practice.[80-84] Our training schedule will concentrate three practice sessions into a three week period followed by a final testing session three months after the first visit. Within each of the three training session's dynamometer training will last approximately 25 minutes with 12, 90 second practice blocks separated by 40 second rest periods. Rest periods providing spaced practice intervals have been shown to significantly improve performance learning and consolidation , compared to training without rest periods.[85] The 25 minute practice time and 3 weeks of training is also consistent with successful wheelchair training protocols completed by DeGroot.[33,34,86]

De Groot, et al. used a 3 week training protocol (three 4-min exercise blocks at a low intensity with 2 min rest, three times a week,) in two studies and found statistically significant improvements in subjects, cycle time, push time and work per cycle, which were similar to findings by van der Woude, L.H.V, 1999, who use a seven week training intervention(30 min exercise at 50–70% heart rate reserve, three times a week).[34,87] Furthermore, de Groot 2002, found that the twelve minutes of manual wheelchair practice in novice able-bodied subjects in the above mentioned study induced a significant decrease in push frequency, however the 12 minutes of practice appeared to be too short to show any significant practice effects on the mechanical efficiency. It was suggested that Fraction of Effective Force (FEF), a measure of propulsion efficiency, may be a variable modifiable only with a more long term training strategy.[34] Rodgers, however found an increase in propulsive moment with a decreased stroke frequency indicating a more mechanically economical propulsive stroke following the training regime of 6 weeks.[45] Rodgers protocol was built on a different methodology however, in that it combined strengthening, stretching, and aerobic exercise to improve biomechanics, rather than a feedback training/practice methodology like de Groot's intervention or our proposed plan. In addition, while Rodgers used actual MWU it is important to note de Groot's subjects were all novice able bodied wheelchair users which may have resulted in some study limitations.[34]

1.6 PRELIMINARY STUDIES

The studies that support this work have been in progress for over 15 years. The Human Engineering Research Laboratories (HERL) at the university of Pittsburgh has developed instrumentation to measure kinetics during wheelchair propulsion through a device known as the Smart^{Wheel}.[88,89] The Smart^{Wheel} is now sold as a commercial device and is being used in over 25 rehabilitation and research centers around the world. We have developed methods for analyzing pushrim forces critical to assessing injury mechanisms.[90-92] We have found stable pushrim force and moment measures that change with speed, and are statistically valid metrics.[93] We have published a large series of MRI and x-ray imaging results for people with paraplegia.[94] This study found a high prevalence of osteolysis of the distal clavicle, another repetitive strain type injury. Interestingly, our study found a much lower prevalence of rotator cuff tears than reported by Escobedo et al.[95] The main difference between the two studies was the age of the populations. The study by Escobedo had an older group with more years since injury. These combined results point to the need for prevention, so that the younger population studied in Pittsburgh does not develop the injuries seen in the Escobedo study.

The work that directly supports this study involves relating injury to wheelchair propulsion variables.[96,96,31,97] We have found that wheelchair pushrim forces are related to nerve conduction

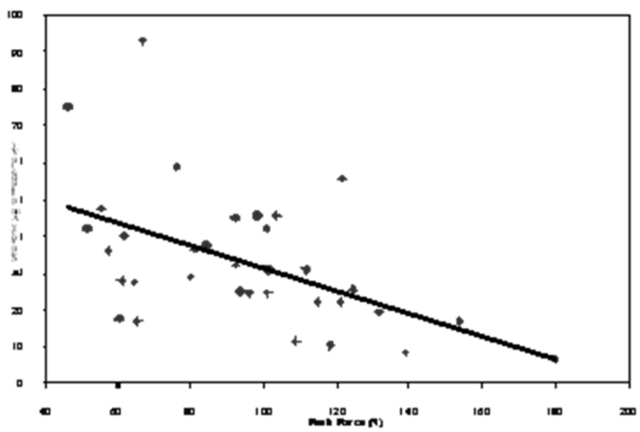


Figure 1. Median nerve function and pushrim forces

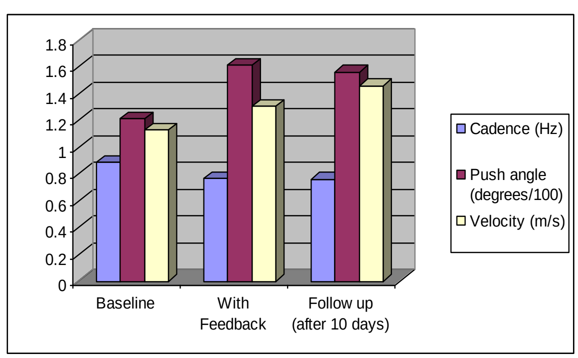


Figure 2. Preliminary data

study variables (see figure 1).[31] Median nerve conduction studies are used to diagnose carpal tunnel syndrome. This study found that, when controlling for weight, there were correlations between median nerve function and the cadence of propulsion and rate and rise of the resultant force. As a follow up to this study, we completed analysis of longitudinal data. The longitudinal analysis showed that risk of injury to the median nerve could be predicted by wheelchair propulsion biomechanics.[98] Individuals who used greater force and cadence at their initial visit had greater progression in median nerve damage approximately three years later at time 2. Once again, peak resultant force was a predictor of progression of nerve conduction study abnormalities. It is important to note that rate of rise of the resultant force is highly correlated with the weight normalized peak resultant force. Our most recent publication found an inverse relationship between median nerve health and range of motions at the wrist.[96] Greater range of motion was associated with better median nerve function. Further analysis found that greater wrist range of motion was associated with greater push angles, lower forces and cadence. By taking long strokes, wheelchair users are able to generate work without high peak forces.

We recently collected preliminary data using real time feedback on a single subject (see figure 2). The subject propelled on our dynamometer while Smart$^{\text{Wheel}}$ data were collected. He then received real time feedback as described below in the methods section. As indicated in the figure with real-time feedback at a self selected speed, the subject decreased cadence and increased push angle. It is interesting to note that, with these changes, velocity increased as well. All of these changes were maintained when we retested the subject 10 days later. This preliminary data supports that a feedback program can have an effect on propulsion and supports the need to collect data at both a fixed velocity and at a self selected speed.

1.7 OVERVIEW, SPECIFIC AIMS AND HYPOTHESIS

Overview

This study followed a group of 27 full time manual wheelchair users over the course of a three month period to determine the effectiveness of a manual wheelchair propulsion training program and or multimedia intervention(video with text overlay). The training protocol taught participants propulsion techniques based on sound biomechanical and motor learning theory principles and assessed changes in biomechanics and any unintended changes in upper extremity pain and shoulder forces that may have emerged or change during the course of the study.

Specific Aim 1): (Chapter one) Develop wheelchair propulsion-training programs based on research that links injury to specific propulsion biomechanics. We developed two interventions, a multimedia instructional program (MMP), and a program that provides real time biomechanical feedback (RTF) to the users. These focused interventions trained users to minimize stroke frequency and maximize contact angle (the angle along the arc of the pushrim from the start of propulsion to the end of propulsion)

Specific Aim 2): (Chapter two) Test which training program causes the greatest short-term changes in propulsion biomechanics. Three groups were compared: a control group (CG) that received no training, an instruction only group (IO) that reviewed the MMP, and a feedback group (FB) that reviewed the MMP and received RTF.

Hypothesis 2a) The IO group will have a slower stroke frequency and larger contact angle when compared to the CG group.

2b) The FB group will have a slower stroke frequency and larger contact angle when compared to IO group.

Specific Aim 3): (Chapter two) Test which training program causes the greatest persistent or long-term changes in propulsion biomechanics.

Hypothesis 3a) The IO group will have a slower stroke frequency and larger contact angle when compared to the CG group 3 months after training.

3b) The FB group will have a slower stroke frequency and larger contact angle when compared to IO group 3 months after training.

Specific Aim 4): (Chapter three) Investigate if resultant forces and moments at the shoulder can be reduced as a result of training.

Hypothesis 4a) The IO group will have reduced resultant forces and moments at the shoulder when compared to the CG group 3 months after training.

4b) The FB group will have reduced resultant forces and moments at the shoulder when compared to IO group 3 months after training

2.0 HAND RIM WHEELCHAIR PROPULSION TRAINING USING BIOMECHANICAL REALTIME VISUAL FEEDBACK BASED ON MOTOR LEARNING THEORY PRINCIPLES

2.1 INTRODUCTION

Due to lower limb paralysis, individuals with spinal cord injury (SCI) rely extensively on their upper limbs for mobility and activities of daily living (ADL). Thus, any loss of upper limb function significantly affects mobility and independence[3,4,99]. Some have gone so far as to suggest that damage to the upper limbs may be functionally and economically equivalent to an SCI of a higher neurological level[100]. Unfortunately upper limb pain is very common in manual wheelchair users, with carpal tunnel syndrome occurring in between 49% and 73% of individuals [100-106] and rotator cuff tendinopathy and shoulder pain present in between 31% and 73%[1,100,104,107-109]. Substantial ergonomics and propulsion biomechanics literature have identified specific biomechanical parameters associated with risk of injury to the upper limb[15-17]. It is possible that appropriately training individuals to propel a wheelchair could result in a significant reduction in upper limb pain and injury. In an effort to reduce secondary injuries, the Consortium for Spinal Cord Medicine recently recommended that individuals minimize the frequency of propulsive strokes as well as the propulsive forces required to manually propel a wheelchair [110]. More specifically, wheelchair users should be encouraged to use low frequency, long and smooth strokes during the propulsive phase while allowing the hand to drift down and back below the pushrim during the recovery phase [110]. Unfortunately, many wheelchair users often receive little to no information from the rehabilitation professionals on how to safely propel a wheelchair and no evidence-based training programs have yet been introduced into clinical practice.

Numerous studies have explored methods in which to improve manual wheelchair propulsion biomechanics. [33,34,43,45,111,112] Two studies have proposed programs focusing primarily on upper limb strength training [45,113], while others have investigated simulated manual wheelchair propulsion training protocols completed on stationary ergometers at low intensities and durations with no feedback or with only visual velocity feedback [111,114,115]. These studies have produced subtle but desirable changes on able bodied subjects like increased mechanical efficiency (ME), push time, contact angle, and decreased stroke frequency accompanied by little to no improvements in force application. To the best of our knowledge, only two research groups have implemented real time visual feedback during wheelchair propulsion training at this time. De Groot *et al* presented able body subjects with real time velocity and Fraction of Effective Force (FEF) feedback and found trained subjects to exhibit higher FEF accompanied by significantly lower mechanical efficiency [34,111]. Kotajarvi BR *et al* presented FEF, velocity, and power output feedback to experienced wheelchair users and again found no improvements in force effectiveness but did observe increased contact angle and decreased stroke frequency [112]. FEF is the proportion of force at the hand rim that contributes to forward motion defined as Ft/F, where Ft is the tangential force obtained by dividing the measured wheel torque by the radius of the pushrim, and F is the resultant force. FEF is also considered a mechanical outcome measure that does not relate to ME consistently therefore higher FEF is not necessarily viewed as better from a physiological or mechanical perspective [116]. It has also been suggested that FEF does not change drastically with exercise or propulsion training because it is controlled largely by the geometry of the wheelchair-user interface which is a closed chain from the shoulder down to where the hand grips the push rim. ME is however an outcome measure shown to express improved performance during sub maximal exercise and is sensitive to both changes in propulsion technique and wheelchair-interface.

While these studies have contributed substantially to the understanding of propulsion training, it is likely that further inspection is warranted, particularly in the area of visual feedback software design and presentation. It is possible that these studies may have had limited success because their visual feedback components were not necessarily designed according to, or supported by, the principles of motor learning theory. Consequently the question of how to best train an individual remains unclear. Motor learning theory indicates that the way in which visual feedback variables are selected and presented are most critical to skill acquisition, performance and retention. [59,75,117,118]

The purpose of this study is to describe the development of a sub maximal training protocol that not only reflects propulsion biomechanics literature and clinical practice guidelines but also attempts to optimize the effectiveness of visual feedback by incorporating elements of motor learning theory into its design. The proposed training protocol incorporates a Biomechanical Feedback-Based Learning Software, with discontinuous real-time viewing of key spatio-temporal and kinetic parameters presented randomly while a participant propel his/her own W/C. The rational of the training protocol parameters along with the technical characteristics of the software, specifically developed for manual wheelchair propulsion training, are described in the first chapter. Preliminary results of one individual with SCI who completed the W/C propulsion training program are presented and discussed.

2.2 METHODS

2.2.1 Biomechanical Feedback-Based Learning Software Development

The training program feedback screen presented 1) contact angle, 2) stroke frequency and 3) velocity. A determination was also made to present these variables randomly and discontinuously (variables ordered randomly and appear and disappear during a trial). Variables would be presented one at a time (contact angle alone or stroke frequency alone) and in combination (contact angle with stroke frequency). A target velocity was also provided with these variables however no velocity was given during the self selected speed condition. It is important to note that training and testing conditions requiring set target velocities were selected to be challenging, manageable and distinct from one another. For example, the target velocity during training on the dynamometer was 2 m/s while over ground testing occurred at 1.5 m/s to promote generalizability and also to accommodate a greater range of wheelchair users. Furthermore, these targets were close to normal adult walking speed, reportedly greater than the self selected velocities of many active manual wheelchair users.

The proposed training software was originally programmed with flexibility in mind to support presentation of a number of continuous streaming variables. These variables include velocity (mps), contact angle (degrees), stroke smoothness (peak/average force ratio), stroke frequency (strokes/sec), peak force (N) and average force (N). These variables were to be presented together, in real time, and continuously during propulsion. However, revisions were made after a review of the literature on motor learning theory and based on suggestions from an expert in the field of motor learning theory and training methodologies. It was determined that the feedback portion of the training program presented an overwhelming number of continuous streaming variables which would be detrimental to learning. Motor learning literature indicates that too many interactive elements presented continuously can quickly exceed the capacity of a person's working memory, increasing cognitive load thus making learning more difficult. [119] Consequently, the number of feedback variables was reduced from six to three and their presentation was to occur discontinuously in random order and combination. The most challenging training scenario involved three variables at once; however the majority of trials were limited to no more than two variables at a time in this exploratory study.

2.2.2 Motor Learning Theory Key Terminology

The items presented in 1-6 below is a brief outline of the motor learning theory concepts applied to the design of the feedback training software:

1. External focus of attention- Shifts a performer's attention away from his or her own movements and toward the effects of those movements which involves development of more automatic control processes. [77,118]
2. Discontinuous variable presentation- variables presented intermittently has been shown to improve learning. [59,75,120]
3. Random practice - Can enhance long-term retention and skill transfer.[59,60,120]
4. Contextual interference- variable practice schedules occur in a random order and combination which improves learning.[59-61]

5. Knowledge of Performance (KP) - KP provides extrinsic, post-response information about movement committing an action to memory.[70]
6. Number of variables presented - Too many variables presented in real time at once can degrade learning. [119,120]

2.2.3 Variable selection

Contact angle (CA) (degrees pushed during each propulsive phase), velocity (m/s), and stroke frequency(SF) (strokes per second) were selected because they have been shown to have a strong association with the development of upper limb impairments.[15-17,31,110,121] For example, studies have found a link between median nerve damage (the pathology behind carpal tunnel syndrome), forces applied to the handrim, and cadence [33,110,112]. A study by Boninger et al. provides longitudinal data that shows how a person propels his/her wheelchair can predict future changes in median nerve integrity. [31,96] In addition, there is substantial ergonomics literature documenting the association between the frequency of a task and force exerted and risk for injury at the shoulder and wrist. [15-17] Contact angle has been included so that while reducing stroke frequency, subjects do not impart large forces over a shorter stroke. By increasing stroke length, while assuming velocity is held constant, subjects will be doing an equivalent amount of work over a longer period of time. This may reduce the force exerted on the pushrim at a given speed. In addition, by focusing on SF and CA, and not directly on the movement pattern of propulsion itself, promotes an external focus of attention, a motor learning technique shown to be beneficial in learning motor tasks. The advantage of focusing on the outcome of one’s movements (external focus) is that the performer’s attention is shifted away from his or her own movements and toward the effects of those movements. This type of learning involves development of more automatic control processes. [75,77]

In addition, a decision was made to not use force feedback because our training program was sub maximal and FEF tends to increase with higher workloads. It has also been suggested that increasing FEF may cause subjects to push with a higher percentage of force tangential to the pushrim which could lead to unintended changes in biomechanics. [33,34,112] Hence, the goal of this training program was to use SF and CA feedback to encourage subjects to take longer, less frequent strokes, to decrease force exerted at a given velocity, without causing unintended changes in force direction.

2.2.4 Presentation of Feedback Variables

Another question critical to the design of an effective training intervention relates to the way in which people learn and synthesize information. Careful thought had to be put into the design of practice schedules as they aimed to promote learning acquisition, retention, focus of attention and transfer of motor skills. Given these considerations, this training protocol has been programmed to provide discontinuous feedback with random ordered repeated training and rest periods. [59-61] Motor learning literature suggests that continuous real time feedback tends to decrease learning and retention of motor skills because the person can become dependent upon that feedback as a substitute for his or her own error-detection and error-correction capabilities. [70,117,120,122] Switching between tasks during practice provides the learner with better contrastive knowledge than the repetitive practice that occurs under a blocked or drill like order [62]. Blocked practice sessions include only one aspect of a task, practicing it over and over until it is performed correctly where random practices employ multiple varied aspects of a task within a session. This contrast between tasks makes learning each task more distinctive and memorable, resulting in improved retention.

Random practice at different propulsion speeds can encourage the learner to compare and contrast the methods and strategies used when propelling at a given speed. [59] Shea & Morgan demonstrated that ordering of motor skills during practice affects immediate performance and retention while the manipulation of practice schedules creates an empirical phenomenon termed contextual interference (CI)[59]. Contextual interference is a term that relates to the quality of learning experience that occurs during random versus blocked practice. [59,60] CI is elevated during random practice because an individual must form an action plan prior to executing the next motion or sequence. When CI is low as is the case with blocked practice, an action plan suitable for an initial task remains in ones working memory ready for the next identical task requiring less effort and thought. The CI effect has often been considered a performance paradox because while the increase in interference caused by random practice schedule diminishes initial acquisition, long-term retention and/or transfer performance is enhanced [59,60,120]. Because the goal of this work was to maximize the long-term learning effects of training, practice schedules were administered randomly as this method has been shown to increase long term skill retention better than blocked practice. [59-62,120]

This training program has also been designed to provide a combination of reinforcement and feedback through knowledge of performance (KP) [69,70]. KP provides extrinsic, post-response kinematic or kinetic information regarding aspects of movement otherwise difficult to perceive. [70] In this context KP, for example knowing contact angle during wheelchair propulsion, directs a learner towards better performance of a goal-directed action like forward motion of a wheelchair.

In contrast, practice without KP allows performance to drift away from the goal, weakening the representation of an action in memory.

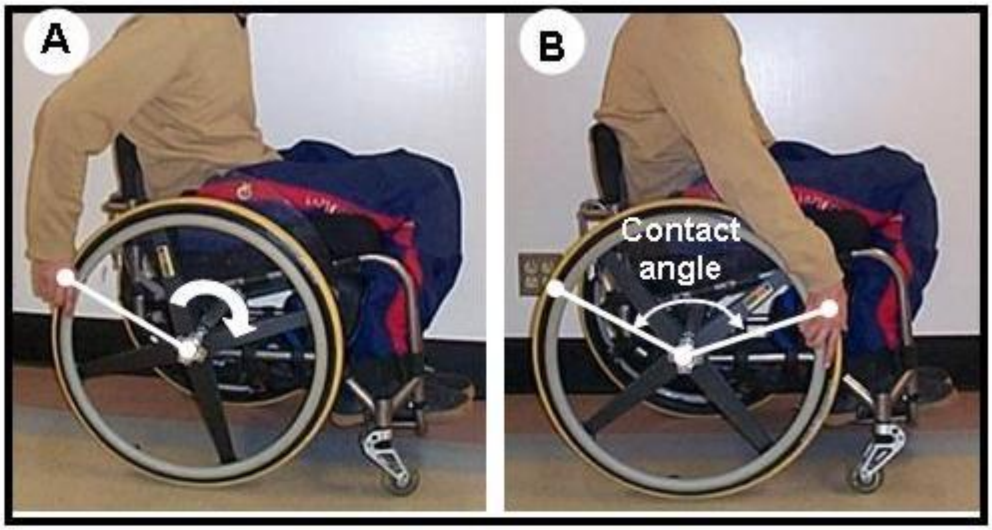


Figure 3. Schematic representation of the contact angle which includes the initial(A) and final(B) hand contact with the push rim during the propulsive phase of manual wheelchair propulsion

2.2.5 Case Study

One long-term manual wheelchair user (gender=male; age=45.6 year-old; weight = 65.7 kg; height = 1.80 m) who sustained a complete T4 SCI more than 11 years ago. Subjective assessment and objective clinical examinations confirmed that this subject was not experiencing any active signs or symptoms of U/E impairments or any other condition that might alter his ability to manually propel his wheelchair during testing and training. The subject provided written informed consent prior to participation in this study as approved by the institutional review board.

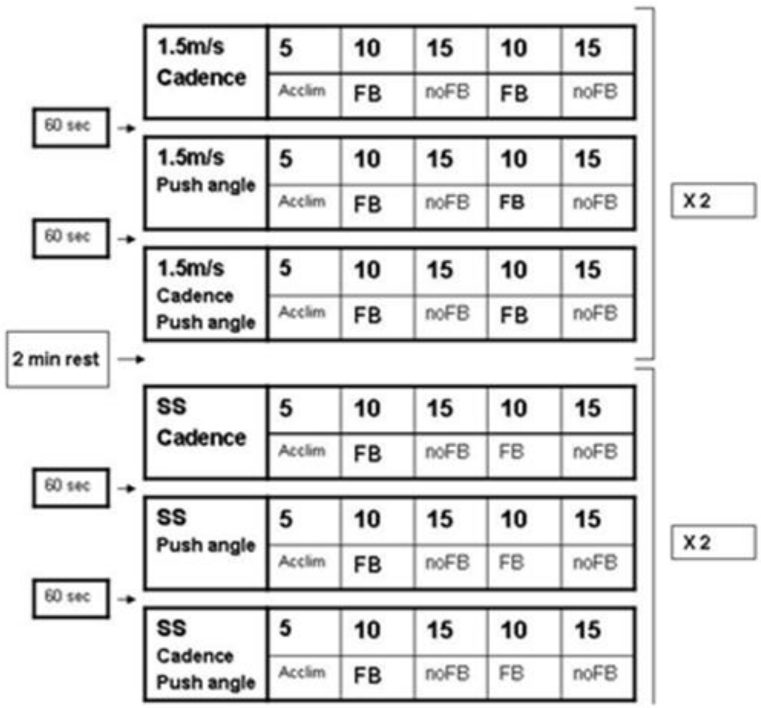


Figure 4. Schematic representation of real-time feedback variable presentation during training. Sixty second rest periods followed each fifty five sec propulsion period and two minutes of rest occurred after each block of training. Within a propulsion period a subject was given 5 sec to acclimate then received visual feedback for 10 sec, no feedback for 15 sec, feedback for 10 sec, and no feedback for the remaining 15 sec. A training block was considered presentation of all combinations of variables with a target velocity or at a selected speed

2.2.6 Instrumentation

The participant's wheelchair was fitted bilaterally with two instrumented wheels (SMART^{wheel}®; Three Rivers Holdings, Mesa, Az) while data were collected only from the side of the non-dominant U/E. [123] This instrumented wheel allows one to accurately record the three orthogonal components (x, y and z) of the forces and moments applied directly to the wheelchair pushrim during propulsion within a local coordinate frame at a sampling frequency of 240 Hz (Figure 1).

The instrumented wheel does not alter the feel or set up of a participants own wheelchair, as it closely replicates usual wheel's size, position and orientation when mounted. While propelling, key propulsion variables collected were streamed as real time visual feedback and presented on a large monitor facing the participant. There was no perceivable delay between the actual action on the hand rim and the feedback received visually on the screen. In addition, the participant's wheelchair configuration was noted and maintained throughout the entire study and the Smart wheels were equipped with solid tires eliminating the need to monitor tire pressure.

During training, the participant's wheelchair was positioned over a custom-built computer-controlled wheelchair dynamometer anchored to the floor using a four-point tie-down system. The dynamometer used for training was comprised of an independent double drum system. The target velocity presented was 2m/s ± .25 m/s in the value of a bar range reportedly close to normal adult walking speed. [124] The study was designed to occur at a low intensity while maintaining speeds and rolling resistance typically encountered during daily propulsion. The rolling resistance of the dynamometer used was fixed at 14.2 N, reportedly just under that of rolling on low pile carpeting.[125,126]

2.2.7 Wheelchair Propulsion Assessment

Biomechanical assessment of over-ground manual wheelchair propulsion was performed upon enrollment and three months post. During these assessments, the subject was instructed to manually propel his wheelchair at a self-selected speed (task #1) and at a pre-determined speed of 1.5m/s (task#2), respectively over a 50-meter distance on an unobstructed, in-door, level tile surface (width=3 m). To ensure that the pre-determined speed of 1.5m/s was maintained during task#2, the subject was instructed to follow a power wheelchair traveling at this preset speed. [127] Kinetic data were recorded during these over ground assessments. Over-ground wheelchair propulsion was favored over dynamometer testing as it is thought to be more representative of routine wheelchair propulsion observed in daily life.

2.2.8 Wheelchair Propulsion Training Protocol

The subject visited the lab four times over a three month period. The first three visits included real time feedback training on the dynamometer followed by testing on the dynamometer and on an over ground course. Training occurred on V1, V2 (10 days after V1), and V3 (10 days after V2). The fourth visit occurred three months after V1 and involved only testing on the dynamometer and overground. All training occurred at low intensities thus isolating effects of technique learning and to prevent physiological adaptation. The longest period of time spent in propulsion without a break was 55 seconds. In that 55 sec, the participant would propel continuously while receiving visual feedback (contact angle, stroke frequency, velocity) which would appear and disappear during the trial (Figure 2). All combinations of variables presented equated to 12 minutes of active propulsion with a total rest time of14 minutes. A target velocity (self-elected speed and 2 m/s) was highlighted in a range of bright colors against which real velocity was plotted. For the stroke frequency and contact angle, real values were directly displayed on the screen (Figure 3). Prior to training the subject was instructed to minimize cadence and maximize contact angle while maintaining velocity.

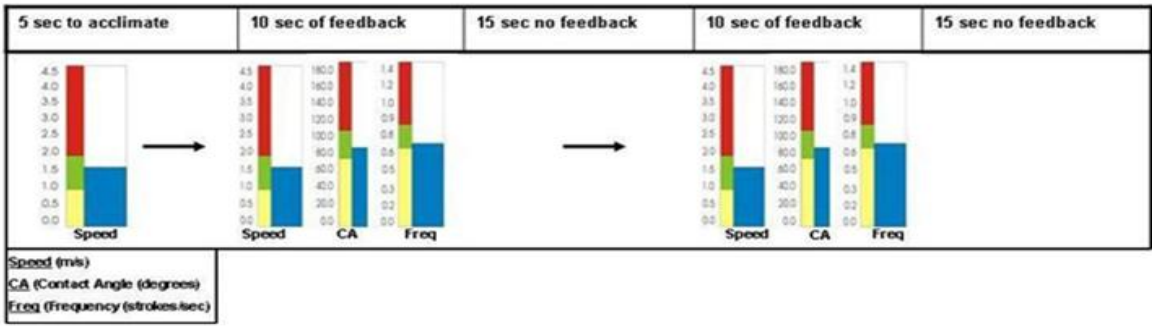


Figure 5. The following screen capture is an example of an actual real-time screen display where all feedback variables are presented. Speed is presented in the first five seconds to help the subject acclimate and reach the steady state target speed then all of the variables are visible for 10 sec and disappear for 15 sec while the subject continues propulsion. The green portions of the bars are target zones while the yellow and red zones indicate an under or overshoot of the green target. The blue bars are the wheelchair users' real time data streamed from the instrumented wheel at 240 Hz. A countdown clock is also provided in the bottom left corner (not illustrated).

2.2.9 Outcome Measures and Data Analysis

Key spatio-temporal and kinetic parameters were computed during the propulsive phase of five consecutive stroke cycles during manual wheelchair propulsion after a near-constant velocity (steady-state) was achieved. The start ($F_{resultant}$>5N) and end ($F_{resultant}$<5N) of the push phase of each stroke were automatically selected using a customized Matlab program (The Mathworks, Natick, MA). Spatio-temporal outcomes included the mean propulsion velocity (m/s), mean stroke cadence (number of stroke/s), mean absolute push time (s) and mean contact angle per stroke (°). Kinetic outcomes included the peak and mean resultant forces applied to the hand rim ($F_{resultant}$), the peak and mean moments out of the plane of the wheel (M_p) applied by the hand and the rate of rise of force (ROR_{Force}). For these outcome measures, changes were observed between the first and final visit 3 months later to document the effects of the training program on one individual.

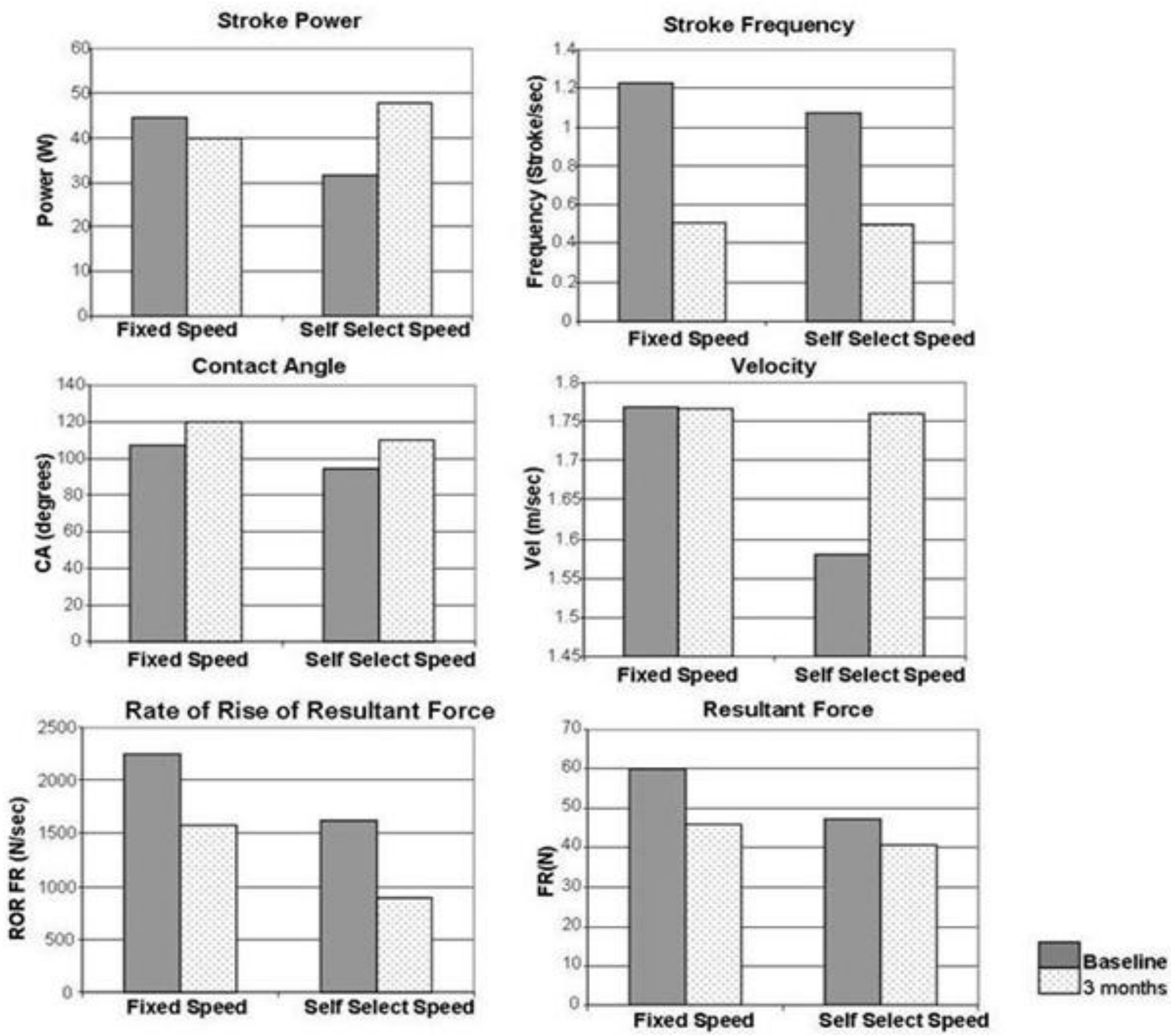


Figure 6. Summary of the kinetic and spatio-temporal outcomes measured pre- and post-training during manual wheelchair propulsion at a pre-determined (1.5m/s) and at a self-selected speeds. Solid bars correspond to mean values measured at baseline, pre-training whereas dotted bars represent mean values reached three months post-baseline after 3 training session.

2.3 RESULTS

As seen in Figure 4, the subject increased contact angle while simultaneously reducing stroke frequency, mean resultant force, and peak rate of rise of force applied to the pushrim at both self selected and pre determined velocities. At a target velocity, contact angle increased from 107.3 ±10.5 to 120.2±15.3 (º), while decreases were seen in stroke frequency 1.2±0.5 to 0.5±0.05 (strokes/ s), mean resultant force 59.9±4.5 to 45.6±7.1 (N), and peak rate of rise of force 2250.1±315.0 to 1584.4±1012.3 (N/s). At a self selected speed contact angle increased from 94.3± 18.9 to 109.7± 12.6 (º), while decreases were seen in stroke frequency 1.0± 0.6 to 0.49± 0.04(strokes/ s), mean resultant force 47.1± 12.3 to 41.6±5.4 (N), and peak rate of rise of force 1623.4± 974.4 to 895.6± 227.5(N/s). A small change in self selected speed was observed pre to post training (1.52mps to 1.45mps). With increased velocity a higher power was seen in force directed tangentially to the hand rim (31.5 watts to 47.9watts) (Figure 4).

2.4 DISCUSSION

Manual wheelchair propulsion can be an intense activity requiring the application of large forces to the push rim repetitively over time. The risk of developing secondary upper limb impairments justifies the need for interventions that can minimize the progression of these potentially debilitating conditions. This paper presents the rational supporting the proposed training protocol along with an overview of its technical characteristics.

The results of this exploratory study confirm that favorable changes can be observed after completing a sub maximal propulsion training protocol based on contact angle and stroke frequency visual feedback. In fact, the case subject studied in this paper was a long term wheelchair user presenting with well-defined propulsion biomechanics; however, training still produced substantial changes. The subject's mean stroke frequency decreased with an increase in mean contact angle. In addition, the mean resultant force, mean moments out of the plane of the wheel, and rate of rise of resultant force were also reduced. All of these changes occurred while velocity remained constant and all stroke improvements occurred and persisted three months after baseline assessment. In the self selected speed condition a small drop in velocity was observed however the subject reduced total average force while generating more power, taking fewer strokes and increasing contact angle. We believe that these values may indicate improved technique.

It was an aim of this project to establish a clinically useful tool geared towards injury prevention rather than purely maximizing gross mechanical efficiency. Direct force feedback like FEF was eliminated as a training variable because literature indicates that it does not always relate to ME consistently and can cause individuals to radically alter their propulsion technique unsafely.[112] In addition, it is evident that propulsion on a dynamometer does not always translate to over ground where an individual must incorporate chair handling skills and utilize visual and environmental cues during propulsion. However the present study showed that the participant's propulsion biomechanics could be improved over ground after training on a dynamometer alone. As more subjects complete this training protocol, the findings will help verify the program's effectiveness, generalizability and the extent to which it can serve as a safe and practical clinical tool.

While careful thought was put into the programs design to illicit improved propulsion biomechanics, inherent limitations exist. It is evident that shifts in mechanical efficiency can take place due to physiological adaptations or as a consequence of improved propulsion technique. [34] As a result of training both physiological adaptations and learning responses (i.e., an improved propulsion technique) can take place as well. If one is to isolate changes in propulsion technique and ME, physiological adaptations as a consequence of training should be excluded. Therefore, a learning protocol needs to occur at a low intensity, duration, and frequency. [34] For example the dosing and timing of practice schedules in this protocol were intended to be sub maximal and spaced out to minimize physiological adaptations. It is apparent though, that some degree of unintended physiological adaption and motor learning may have occurred. It is likely that even a control subject propelling without training or receiving only velocity feedback could exhibit natural learning and some degree of physiological adaptation.

The current protocol requires at least one instrumented wheel, a dynamometer and a computer to support the software which could be perceived as disadvantageous. However, the cost savings resulting from a potential reduction in secondary upper limb impairments could easily off-set initial expenses and should not be overlooked. It is also important to consider the possibility that a successful training program could be carried out based on the principles of the current work using less equipment and technology. For example, with only verbal feedback provided by a therapist and one instrumented wheel, it may be possible to effectively measure and train an individual's stroke technique over time. In fact, clinicians without access to an instrumented wheel could still teach propulsion using verbal or auditory feedback based on the motor learning principles and clinical practice guidelines presented in the current paper. A client could be told to take low cadence, use long smooth strokes, with intermittent verbal instruction, over random ordered practice surfaces like carpet, ramp and tile. Again, consistent with motor learning theory, a client could then benefit from additional feedback which places emphasis on the effects of their movements like number of strokes taken, speed, and stroke length rather than movement pattern. The use of a low-cost video camera could also be used to record propulsion technique which could assist both the instructor and client throughout the learning process.

How far subjects are from propelling with ideal technique will vary from individual to individual. However, the literature has reported that a large range of cadence and push angle often occurs within representative populations of wheelchair users. [37,56,128]More specifically, a study completed comparing the propulsion techniques of long term wheelchair users found that at a given speed the average group cadence varied from 1.1 to 1.6 stroke per second and the push angle varied from 102 to 134 degrees.[37] The significant group variability indicates that there is considerable room for improvement. In addition, gross ME during propulsion has been shown to rarely surpass 11% which also suggests improvements in technique may be attainable. [129] These findings may have meaningful and clinically significant implications as propulsion occurring inefficiently may place a person at significant risk for developing upper extremity pain and injury. Because wheelchair propulsion involves impacting the pushrim thousands of times per day and clearly exceeds what the ergonomics literature considers to be a high force, high repetition task, any improvements could have an impact on the development of upper extremity pain and injury.

This case study was ultimately a starting point and could eventually serve as a new teaching approach for rehabilitation practitioners. Additional research needs to be conducted with a greater number of long-term manual wheelchair users in a randomized design with a control group before the beneficial effects of training can be ascertained. Then, the next logical step may be to develop a randomized clinical trial to verify if this program prevents, or limits, the development of secondary upper limb neuro-musculo-skeletal impairments over time.

2.5 CONCLUSION

This study translated key principles of motor learning theory into visual feedback- learning software presenting customized spatio-temporal and kinetic variables known to be critical to the development of efficient propulsion techniques. Preliminary results indicate that clinically-relevant changes can be expected three months from baseline after only three low intensity wheelchair propulsion training sessions completed over a twenty day period by a long-term manual wheelchair.

3.0 HAND RIM WHEELCHAIR PROPULSION TRAINING EFFECT ON OVERGROUND PROPULSION USING BIOMECHANICAL REAL TIME VISUAL FEEDBACK

3.1 INTRODUCTION

The first chapter described the theory behind the development of our training program that was tested on a pilot subject. The current study describes how training impacted the propulsion biomechanics of a larger group of subjects on a diverse over ground course.

Due to lower limb paralysis, individuals with spinal cord injury (SCI) depend on their upper extremities for mobility and to perform activities of daily living (ADL). The occurrence of upper limb pain and injury can therefore impact functional mobility and lead to decreased independence and quality of life.[2,4,99] Some have gone so far as to suggest that damage to the upper limbs may be functionally and economically equivalent to an SCI of a higher neurological level[100] . Unfortunately upper limb pain is very common in manual wheelchair users, with carpal tunnel syndrome occurring in between 49% and 73% of individuals [100-106] and rotator cuff tendinopathy and shoulder pain present in between 31% and 73%.[100,104,107,108,130,131] Substantial ergonomics and propulsion biomechanics literature have identified specific biomechanical parameters associated with risk of injury to the upper limbs.[15-17] These studies indicate that both forces and task repetition should be kept to a minimum in order to reduce the risk of injury. It is possible that appropriately training individuals to propel a wheelchair could result in a significant reduction in upper limb pain and injury as well.

In an effort to reduce secondary injuries, the Consortium for Spinal Cord Medicine has recommended that individuals minimize the frequency of propulsive strokes as well as the propulsive forces required to propel a manual wheelchair.[110] In essence, wheelchair users should be encouraged to use low frequency, long and smooth strokes during the propulsive phase while allowing the hand to drift down and back below the pushrim during the recovery[110]. Unfortunately, many wheelchair users receive little to no information from rehabilitation practitioners on how to safely propel a wheelchair and evidence-based training programs do not yet exist in clinical practice.

Current literature exploring methods to improve manual wheelchair propulsion biomechanics is scarce.[33,34,43,45,111,112] Two studies have proposed programs focusing primarily on upper limb strength training[45,113] while others have investigated low intensity training protocols on stationary ergometers using very limited forms of real time visual feedback.[111,114,115] These studies have produced subtle but desirable changes on able bodied subjects like increased mechanical efficiency (ME), push time, contact angle, and decreased stroke frequency accompanied by little to no improvements in force application. To this point, only two research groups have implemented real time visual feedback into a propulsion training system. De Groot et al. presented able body subjects with real time velocity and Fraction of Effective Force (FEF) feedback and found that increased FEF caused significantly lower mechanical efficiency.[33,111] Kotajarvi BR et al. presented real time FEF, velocity, and power output feedback to experienced wheelchair users and found no improvements in force effectiveness however subjects did increase contact angle and reduce stroke frequency.[112] FEF (Ft^2 /F^2 x100) is a variable that has been considered for its training value because it is the only force acting perpendicular to the hand rim that contributes to forward motion of the wheel.[132,133] Ultimately, FEF has not been found to be an overly effective training tool because it has not consistently related to Mechanical Efficiency ME which is ratio between power output and oxygen cost.[116] In addition, FEF does not appear to change drastically with exercise or propulsion training because it is controlled largely by the geometry of the wheelchair-user interface which is a closed chain from the shoulder down to where the hand grips the push rim. [132,133] Thus far, ME has been shown to be a reliable measure of propulsion performance because it is sensitive to both changes in propulsion technique and wheelchair-interface.

In addition to selecting outcome and training measures that best reflect propulsion performance, the physical environment in which propulsion occurs must be considered. For example, a majority of the aforementioned training studies have used only a treadmill or dynamometer rather than overground or real life propulsion scenarios and have not allowed participants to use their own wheelchairs. [33,34,43,45,111,112] In addition many propulsion training studies have used only able bodied subjects propelling at pre determined velocities. Although simulated environments allow for more rigorous experimental control of training and analysis, their findings may have less practical application and limited generalizability.

The present study was designed to allow researchers sufficient experimental control of training and testing parameters while viewing propulsion under the most realistic conditions possible. All participants were full time, experienced wheelchair users propelling in their own personal wheelchairs and propulsion testing occurred only on a real life course at pre determined and self selected speeds.

Two training systems were developed, both based on established motor learning theory, biomechanical and ergonomic principles as described in Rice et al. 2010.[134] The first system was a multi media instructional presentation (MMP) which defined key learning terms like contact angle (CA) and stroke frequency (SF) and described how to improve propulsion through correct application of these terms. A real time feedback (RTF) system was also developed to serve as a high tech supplement to the MMP which provided additional focused reinforcement based on substantial motor learning theory. The purpose of the MMP was to convey the same training concepts as the RTF system but in an easy to access, low tech package that if proven effective has the potential to be easily disseminated.

The purpose of this study was to determine if it was possible to train subjects to increase contact angle (CA) and decrease stroke frequency(SF) though these training systems and then to determine if one system was more effective than the other. The studies was also designed to test which training program caused the greatest short-term(same day) and long term changes (three months from baseline) in propulsion biomechanics over a real life course consisting of carpet, ramp and tile. Three groups were compared: a control group (CG) that received no training, an instruction only group (IO) that reviewed the (MMP), and a feedback group (FB) that reviewed the MMP and received additional (RTF). It was hypothesized that the FB group would have a larger contact angle (CA) and reduced stroke frequency (SF) in the short and long term compared to the IO group after training who would show the same improvements compared to the CG.

Although not included in the hypothesis, data was collected on pushrim forces and on subjects upper extremity pain to determine if CA and SF modification caused any detrimental or unintended changes in stroke characteristics.

3.2 METHODS

3.2.1 Subjects

All subjects provided written informed consent prior to participation in this study as approved by the institutional review board. The study population included men and women who use a wheelchair as their primary means of mobility (>80% their ambulation.), aged 18-65 who independently self-propel. Subjects were excluded if they had a spinal cord injury (SCI) above cervical 7, had a history of non-dominant traumatic upper extremity injury at the wrist or shoulder, were less than one year post SCI or had a disability that was progressive or degenerative. In addition all subjects had to use the same wheelchair throughout the entire study without any alterations in configuration.

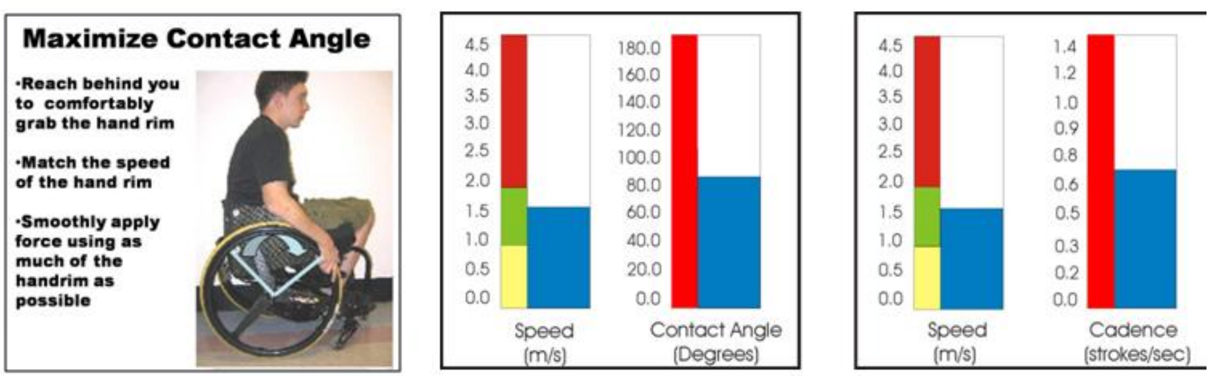


Figure 7. (Left) Sample Multimedia Instructional Presentation (MMP) screen. (Right) Sample Real Time Feedback (RTF) display

3.2.2 Description of Training Protocol

Subjects were randomized into one of three groups (FB, IO or CG) using a random permuted block method.[135] The FB and IO groups received different training regimes as described in the study hypothesis above. The training intervention groups received one (IO group) or two training elements (FB group) depending on group assignment and general verbal instruction prior to practicing techniques overground. The training elements consisted of:

<u>Multimedia Instructional Presentation (MMP)</u> Given to both FB and IO groups, the MMP was designed to be a free, easy to access intervention that allows for independent learning. The system was created in Microsoft Power Point to be an automated instructional video and slide presentation. First, the presentation emphasized how the repetitive nature and physical demands associated with propulsion can lead to the development of pain and injury. Then, the MMP stressed that use of specific propulsion techniques may help to minimize the development of pain and injury thus maximizing quality of life. The MMP defined key learning terms like contact angle (CA) and stroke frequency (SF) and encouraged subjects to maximize CA while minimizing SF (Figure 7). Left shows a sample frame from the MMP discussing contact angle.

<u>Real Time Feedback (RTF)</u> Given only to the FB group during dynamometer propulsion, the RTF training protocol was designed to reinforce the principles presented in the MMP. Real time SF and CA feedback was provided to encourage FB subjects to take longer, less frequent strokes, we believed that this would decrease force exerted at a given velocity, without causing unintended changes in force direction. The programs feedback screen presented 1) CA, 2) SF and 3) velocity (Figure 1 Right). All variables were shown randomly and discontinuously

(variables ordered randomly and appear and disappear during a trial), one at a time (CA alone or SF alone) and in combination (CA with SF)(Figure 8 Left). All training was intended to occur at a low intensity to isolate technique learning and to minimize physiological adaptation as described by De Groot et al.[34] For example, the longest period of time spent in propulsion without a break was 55 seconds (Figures 8 Left& Right). Training was also grouped into two speed categories, where a subject propelled at a freely chosen speed (self selected condition) or was given a target to hit and maintain (target speed condition). In the self selected condition a subject propelled naturally, while reacting to CA and SF feedback without any knowledge of their real time speed. In contrast, the target speed condition allowed subjects to see their real time velocity with a target goal. The target provided was 2.0 meters per second (mps) ± .25 mps in the value of a bar range reportedly close to normal adult walking speed.[124] This target speed was different from the testing target (1.5mps) to investigate the extent to which the propulsion techniques learned could be reproduced at different speeds.

Although IO and CG groups did not receive RTF training they spent equivalent amounts of time on the dynamometer as the FB group did practicing and testing (Figure 8 Left& Right). The IO and CG groups were also presented with a different screen than the RTF screen shown to the FB group. The IO and CG screen included a velocity bar that would appear for the target speed condition and disappear for the self selected condition. In addition, they were provided slightly different general verbal instructions after practice on the dynamometer in preparation for practice on the HERL over ground course. The FB and IO groups were given general verbal instruction consistent with the MMP (described below), while the CG was simply told to propel naturally.

General Verbal MMP Instruction

Prior to over ground practice the IO and FB subjects were reminded of the concepts they had been taught while dynamometer training. General verbal instructions were provided which encouraged them to:

- Maximize contact angle = “keep a long smooth stroke, get on the rim early (solid contact) and hang on as long as possible” (Figure 7 left)
- “At the end of the push phase when hands release, relax arms and let them swing back in preparation for the next stroke”
- The importance of the recovery phase was emphasized. “As propulsion speed increases it may be necessary to swing arms backwards more quickly and deliberately”
- “Keep head neutral look forward or straight ahead”
- “Don’t feel as though you must stay on the back rest”
- “Resist grabbing the tire for propulsion”

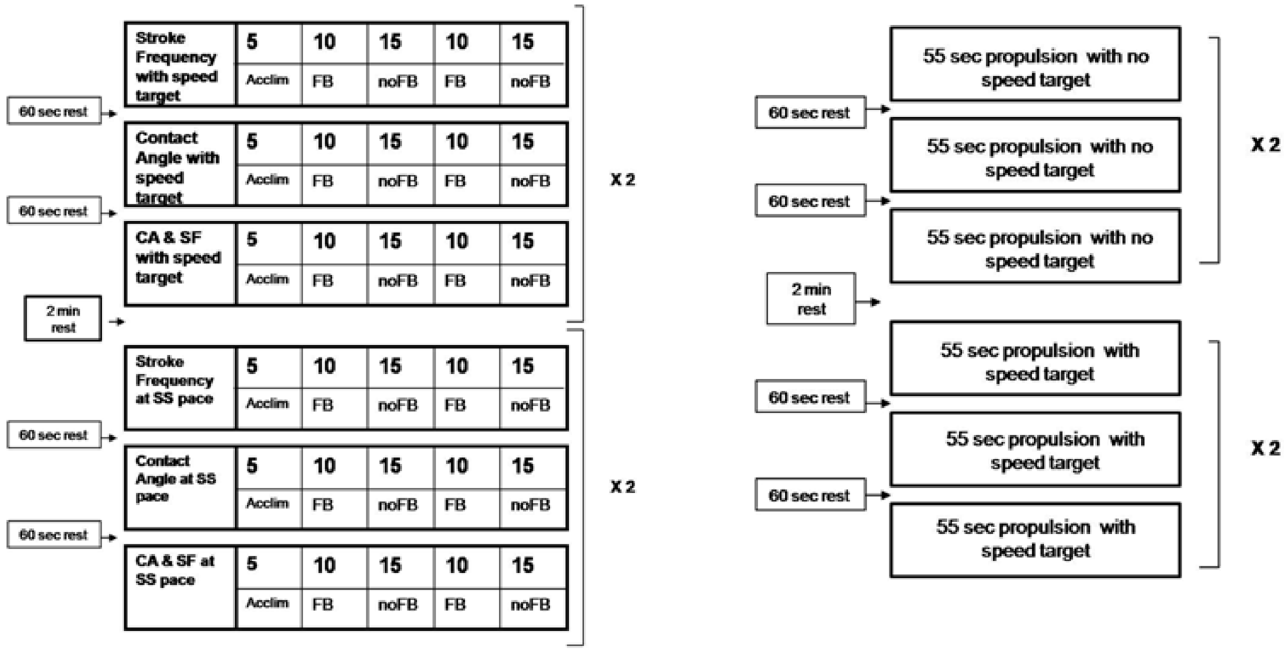


Figure 8. (Left) Feedback group, total rest and propulsion time on dynamometer (Right) IO and CG group, total rest and propulsion time on dynamometer which was equivalent to the FB group

Study Design		
Randomize Subjects		
Control Group	Instruction Group	Feedback Group
(1) Initial Biomechanical Testing Overground		
(2) Training		
Rest for 10 minutes	View MMP and rest 10 minutes total	View MMP and rest 10 minutes total
Propel on Dynamometer	Practice on Dynamometer	RTF on Dynamometer
Propel on course at HERL	GVI given then Practice on course at HERL	GVI given then Practice on course at HERL
(3) Repeat Biomechanical Testing Overground		
Repeat steps (2) ten days after initial testing		
Repeat steps(2)) twenty days after initial testing		
Repeat Step (1) three months later		

Figure 9. Study design overview. General Verbal Instruction (GVI), Human Engineering Research Lab (HERL), Multi Media Presentation (MMP)

3.2.3 Protocol Time Frame

FB and IO subjects visited the lab four times over a three month period while control group (CG) subjects visited the lab 3 times over a three month period. Therefore, the time from baseline to final testing was equivalent for all subjects. Training, which occurred on a dynamometer and over ground occurred on V1, V2 (10 days after V1), and V3 (10 days after V2). The fourth visit occurred three months after V1 and involved only over ground testing. To minimize drop out, the CG group was required to visit the lab 3 times however their rest and activity times matched the treatment groups (Figure 9). The protocol was carefully designed to ensure that all subjects' propulsion and rest times were equivalent regardless of group membership (Figure 9). For example while the FB and IO groups watched the MMP, CG members were instructed to rest for the same amount of time it would have taken to view the MMP (Figure 9).

3.2.4 Over ground Course

Subjects propelled over two different over ground courses during testing and training. Both courses incorporated: 1) traveling up a 1.2 º ramp, 2) traveling over level tile, and 3) traveling over level medium-pile carpet. Each condition was 15 meters long to allow sufficient space for the subjects to complete at least 5 strokes at both self selected and target velocities. The Smart^{Wheel} was used as a data collection tool during this activity. The real life course was completed twice, once at a self selected speed and once at 1.5 mps with the exception of the ramp which was completed only at a self selected speed. For the 1.5 mps trial subjects followed a power wheelchair traveling at that preset speed. Power wheelchairs have been used to pace a manual wheelchair successfully in previous studies.[136] The control of velocity was of particular importance because the easiest way to reduce cadence and reduce peak push rim force would be to slow velocity which would not result in fewer cycles for a given distance.

3.2.5 Instrumentation

The participant's wheelchair was fitted bilaterally with two instrumented wheels (SMARTwheel®; Three Rivers Holdings, Mesa, Az) while data was collected from the side of the non-dominant U/E as it may be less affected by pathology not related to wheelchair propulsion.[123] The instrumented wheel allows one to accurately record the three orthogonal components (x, y and z) of the forces and moments applied directly to the wheelchair pushrim during propulsion within a local coordinate frame at a sampling frequency of 240 Hz.[89] The instrumented wheel does not alter the feel or set up of participants own wheelchair and it closely replicates usual wheel's position and orientation when mounted. The SMARTwheels® do weigh close to twice the weight of a typical light weight everyday wheel. While propelling, key propulsion variables collected were streamed as real time visual feedback and presented on a 17 inch monitor facing the participant. There was no perceivable delay between the actual action on the hand rim and the feedback received visually on the screen. In addition, the participant's wheelchair configuration was maintained throughout the entire study and the Smart wheels were equipped with solid tires eliminating the need to monitor tire pressure.

The dynamometer used in the study was custom-built and computer-controlled and each participant's wheelchair was anchored to the floor using a four-point tie-down system. The dynamometer was comprised of an independent double drum system. The rolling resistance of the dynamometer was fixed at 14.2 N, reportedly just under that of rolling on low pile carpeting.[125,126]

3.2.6 Statistical Analysis

A mixed analysis of covariance (ANCOVA) was performed using a general linear mixed model (GLMM). A 3 x 3 x 2 x 3 GLMM was used with an unspecified covariance matrix. All of the main effects and the 2-way interactions with group were included in the model. The between-subject variables were group, having three levels (FB, IO and CG), surface having 3 levels (carpet, ramp, and tile) and speed having two levels (self selected and target). Since subjects were measured repeatedly, observations were nested within individuals for the 3 time points (baseline, short term and long term). Weight, time since injury, and level of injury were added to the model as covariates while velocity was included as a time dependent covariate for analysis of contact angle and stroke frequency. All forces have been normalized to velocity [F (Newton)/Velocity (mps)]. Follow up comparisons consisted of a priori contrasts and post hoc pair wise comparisons which were performed using Bonferroni adjustments to correct for inflation of Type I error. Alpha for all analysis was set at .05

3.2.7 Data Analysis

Contact angle (degrees), stroke frequency (strokes per second), resultant force (Newton/mps), and rate of rise of force (N/meter) adjusted means are presented in table 5. Baseline measures were recorded prior to training on visit one, where short term change has been defined as the period of time from baseline to immediately after training the same day on visit one. Long term change is defined as the change from baseline to three months.

3.3 RESULTS

3.3.1 Demographics

Short term comparisons were based on analysis of 27 subjects enrolled in the study. Subjects included 3 female and 24 males randomized into Feedback (FB), Instruction only (IO) and Control group (CG) categories. Subject characteristics are shown in Table 1. Due to subject drop out, long term observations were based on 22 subjects. Drop out, or missing data was due to scheduling conflicts and not complications associated with the training interventions. However, an independent sample T test was performed comparing the baseline demographic characteristics of those who stayed in to those who dropped out and no significant differences were found (Table 2).

Table 1. Subject Demographic Characteristics

Group	Short Term(same day)				Long term(3 month)			
	FB	IO	CG	Total	FB	IO	CG	Total
n	9	9	9	27	6	7	9	22
Age (yrs)	36.9(13.7)	37.5(10.3)	46.7(15.2)	40.0(13.4)	41.5(14.4)	39.2(11.2)	46.7(15.2)	42.3(13.6)
Years Since Injury (yrs)	19.9(11.0)	15.0(11.9)	19.1(10.1)	18.0(10.9)	21.4(10.4)	13.1(11.6)	19.1(10.1)	17.9(10.8)
Height (m)	1.75(0.07)	1.67(0.14)	1.80(0.08)	1.74(0.11)	1.77(0.04)	1.67(0.16)	1.80(0.08)	1.75(0.12)
Weight (kg)	76.1(19.7)	82.9(23.7)	84.3(8.3)	80.4(18.8)	82.7(19.8)	89.8(22.0)	84.3(8.3)	85.6(16.8)
Injury Level (median)	T2	T8	T10	T7	T2	T4/5	T10	T7
Gender (Male/Female)	8/1	8/1	8/1	24/3	7/0	6/1	7/1	20/2

Age, years since injury, height and weight reported as mean (SD), level of injury reported as median

Table 2. Subject Drop Out Comparison

Drop Out Change (T -Test)		Characteristics of Subject Who Did not Drop Out				Characteristics of Subject Who Droppe Out			
	Group	FB	IO	CG	Total	FB	IO	CG	Total
	n	7	7	8	22	3	2	0	5
p=0.77	Age (yrs)	41.0(14.4)	38.0(6.9)	49.0(17.1)	42.2(13.5)	33.7(12.6)	45.4(18.7)		33.2(12.4
p=0.2	Years Since Injury (yrs)	21.4(10.4)	12.8(11.4)	21.2(9.7)	18.8(10.6)	20.0(13.6)	17.0(21.2)		17.1(12.6
p=0.67	Height (m)	1.75(0.04)	1.65(0.14)	1.76(0.09)	1.72(0.10)	1.70(0.1)	1.75(0.14)		1.6(0.1)
p=0.56	Weight (kg)	82.7(19.8)	83.5(22.9)	82.4(4.9)	82.4(4.9)	62.8(12.1)	89.4(33.6)		71.6(24.3
p=0.73	Injury Level	T2	T4	T10	T6	T2	T5		T4

Age, years since injury, height and weight reported as mean (SD), level of injury reported as median
T -test p values represent statistical differences between drop out and non drop subject characteristics

3.3.2 Demographics and Biomechanics

An estimate of fixed effects analysis was performed to predict contact angle, stroke frequency, resultant force and rate of rise of resultant force based on demographic characteristics. Based on the model, older subjects tended to use a smaller contact angle, F (1, 159.0) = *36.7,* p < .001,

β = -.42; and used more strokes while lower level injured subjects used fewer strokes, F(1,168.3)=10.15,p=.002, β =-.003; F(1,164.6)=24.25, p=.001, β= -0.072. In addition, older and heavier subjects tended to use greater peak Fr, while lower level injured subjects used less peak Fr, F(1,169.8)=4.02,p=.04, β=*0.24*; F(1,170.9)=160.03, p=.001, β=*0.41*; F(1,166.9)=28.0,p.001, β=-9.2. Weight was found to be the only predictor of rate of rise of resultant force (rorFr) with heavier subjects producing greater peak rorFr F(1,169.8)= 12.65, p=.001, β=3.96.

3.3.3 Interaction Effects

3.3.3.1 Group by Time

Groups displayed statistically significant group by time changes in contact angle(CA),stroke frequency(SF), resultant force (Fr) and peak rate of rise of resultant force(rorFr) after training p<.05(Table 4). This meant that these dependant variables changed differently depending on group membership from baseline to short term and long term time points.

3.3.3.2 Group by Speed by Time

The 3-way interaction of group (FB, IO and CG) x speed (self select and target) x time (baseline, short term, and long term), was not found to be statistically significant for contact angle(CA),stroke frequency(SF), peak resultant force (Fr) or peak rate of rise of resultant force (rorFr) (p>.05)(Table 3). This meant that dependant variables did not change differently based on the self selected or target velocity conditions (presence or absence of a target). Because the two conditions were not significantly different statistical analysis was performed based on the average of the self selected and target speed conditions.

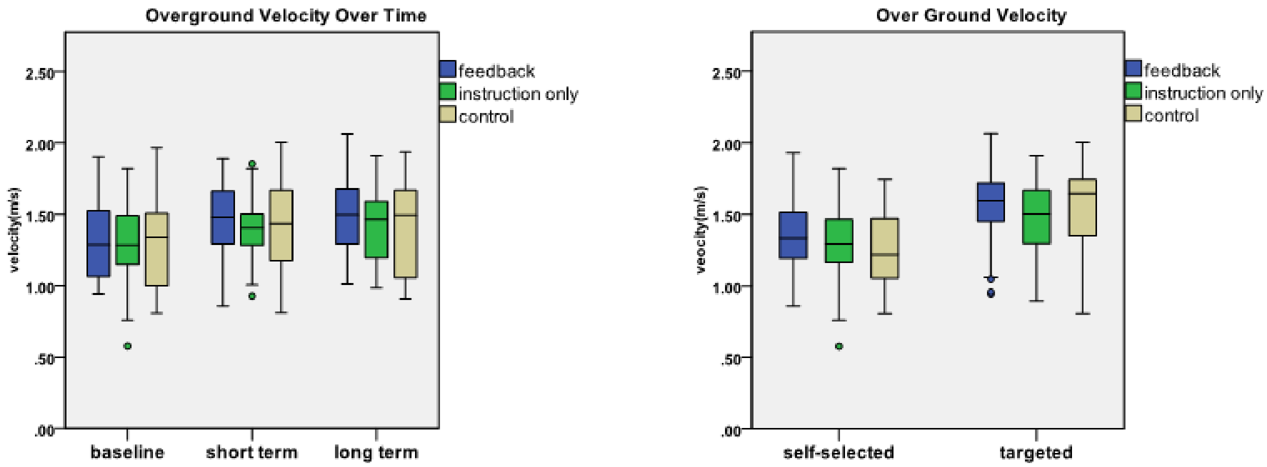


Figure 10. (Left) The following descriptive box plot represents the self selected and target speed conditions averaged together, separated by group, and displayed over time.(Right) Box plot of the self selected and target speed conditions averaged together, separated by group, and displayed over time.

3.3.3.3 Group by Surface

Significant interactions were not found between group membership (CG, IO or FB) and surface type (carpet, ramp or tile) ($p>.05$) Table 3. Therefore the affects of training were not influenced by surface type. Consequently analyses of the dependant variables have been performed based on the average of carpet, ramp, and tile surfaces.

Table 3. Interaction Effects

	CA (°)		SF(strokes/sec)		Fr(Newton/mps)		rorFr (Newton/meter)	
Interaction	**Sig**	**F score**	**Sig**	**F score**	**Sig**	**F score**	**Sig**	**F score**
*Group by Time	p=.001	*F*(4, 65.5)= 27.50	p=.001	F(4,164.8)=27.8	p=.002	F(4,166.4)=4.4	p=.0001	F(4,162.9)=8.2
Group by speed by time	p=.629	*F*(3,166.5) = 0.46	p=.811	F(3,169.8)=0.21	p=.86	F(3,177.3)=0.14	p=.97	F(3,176.8)=0.027
Group by Surface	p=.098	*F*(4,165.1) = 1.80	p=.568	F(4, 169.0)=0.80	p=.48	F(4,171.6)=.91	p=.325	F(4,169.3)=.032

(CA) contact angle; (SF) stroke frequency; (Fr) resultant force; (rorFr) rate of rise resultant force
F score= F(degrees freedom between, degrees freedom within) = f score statistic

Table 4. Adjusted Means Averaged Across all Surfaces

		CA (°)		SF(strokes/sec)		Fr(Newton/mps)		rorFr (N/meter)	
Group	**Time**	**Mean**	**Std. Error**	**Mean**	**Std. Error**	**Mean**	**Std. Error**	**Mean**	**Std. Error**
CG	**Baseline**	100.5	1.6	1.05	0.02	71.9	2.6	1082.6	70.1
	Short term	99.6	1.7	1.10	0.02	72.4	2.4	1119.4	65.7
	Long term	99.2	1.6	1.10	0.03	79.3*,B(IO)	3.1	1397.5 *,B(IO&FB)	100.4
IO	**Baseline**	95.0	1.5	1.20	0.02	73.2	2.4	1266.6	64.9
	Short term	102.6*,A	1.5	0.94*A,	0.02	78.9 *,A,B	2.2	1091.3*,A	60.6
	Long term	104.6*,A,B	1.4	0.93*,A,B	0.03	73.1	2.7	1060.3*,A	88.6
FB	**Baseline**	97.9	1.5	1.04	0.02	71.9	2.4	1183.0	64.8
	Short term	107.7*,A	1.5	0.82*,A	0.02	81.0*,A,B	2.2	814.8*,A	61.0
	Long term	111.8*, A,B	1.5	0.87*,A,B	0.03	74.8	2.9	916.8*,A	94.3

*Significant difference (p<0.05) within group change over time from baseline using pairwise comparison with Bonferroni correction.
A Significant difference from control group (p<0.05)
B Significant difference from treatment groups (FB from IO) (p<0.05)
Covariates included weight, time with injury, and level of injury.
Adjusted means are average of carpet, ramp and tile and average of self selected and target speed conditions

3.3.3.4 Contact Angle and Stroke Frequency

When controlling for velocity, weight, time with injury and level of injury the FB and IO groups showed a significant increase in CA and decreases in SF at both short and long term time points were the CG group did not change significantly (table 4, fig 10-12). The FB and IO groups were significantly different from the CG as well (table 4, Fig 10-12). Significance between the two treatment groups (IO and FB) also occurred in long term but not short term time points for both CA and SF(table 4). Furthermore, in both short and long term time points the FB group showed a significantly greater percent increase in CA than the IO group and the IO group from the CG(Figure 10-A). In terms of SF the FB and IO groups showed an identical short term decrease however in the long term the IO group showed a greater percent decrease than the FB group (Figure 10-B).

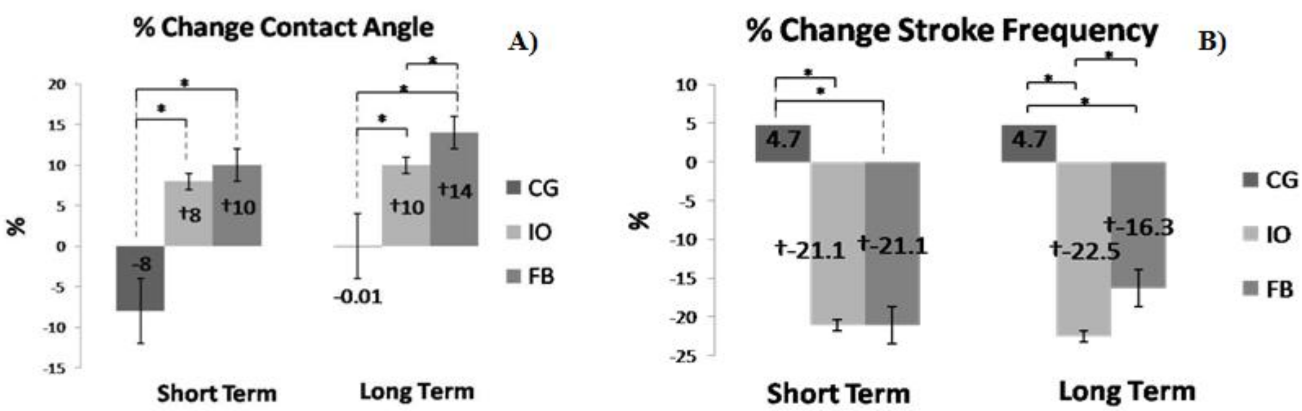


Figure 11. % change in contact angle and stroke frequency from baseline where baseline = 0%
*** Significant change between groups p<.05,**
† Significant difference within group change from baseline (p<0.05)

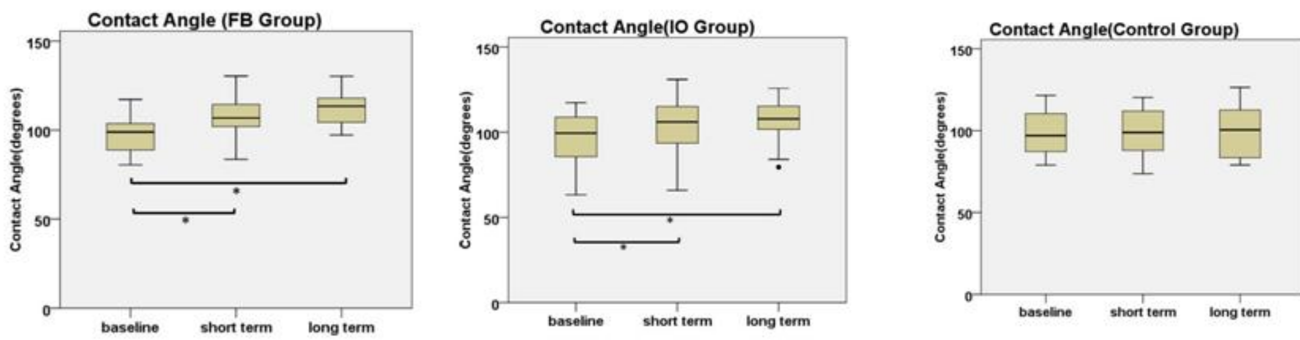


Figure 12. Box plots of the changes in mean contact angle by group, averaged across all surfaces, speeds, over time. All means have been adjusted for time with injury, level of injury, and weight. * represent a significant difference between conditions p<.05

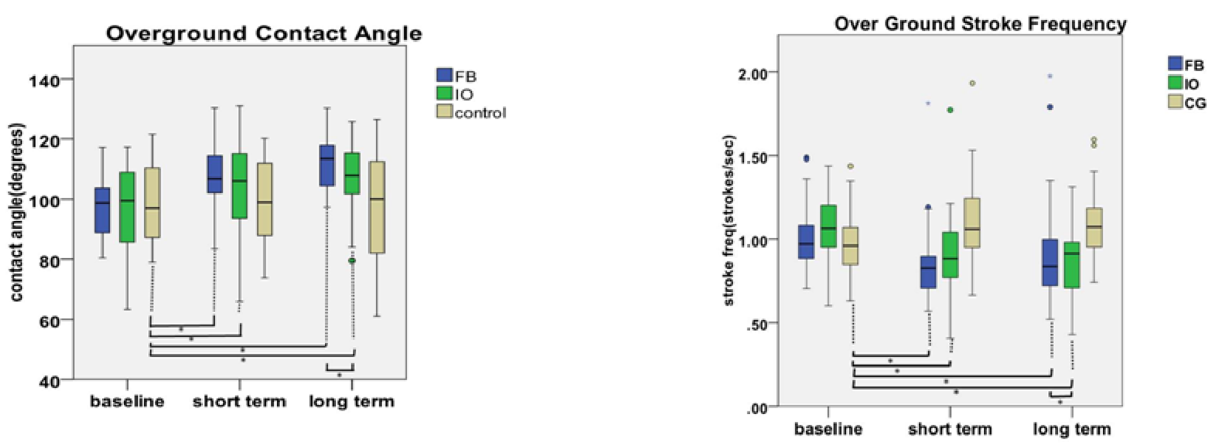


Figure 13. (Left)-Box plot of change in contact angle (degrees) over time. Means are presented by group, time, and averaged across all surfaces. * represent a significant difference between conditions p<.05(Right)- Box plot of change in stroke frequency (strokes/ second) over time. Means are presented by group, time, and averaged across all surfaces. * represent a significant difference between conditions p<.05.

3.3.3.5 Resultant Force and Rate of Rise of Resultant Force

When controlling for velocity, weight, time with injury and level of injury, both The FB and IO groups showed significant short term increases in peak FR at the hand rim that were also significantly different from each other($p<.05$)(Table 4). The FB group also showed a larger percent increase in peak Fr than the IO group in the short term (Figure 13 A). Their long term changes were not significantly larger than baseline however. In contrast the CG showed a significant increase in long term peak Fr that was also significantly different from the FB and IO groups (Figure 13-A).

The FB and IO groups showed significant short and long term reductions in peak rorFr that were significantly different than the CG but not from each other $p<.05$(Figure 13-B). The FB group showed a larger percent decrease in peak rorFR than the IO group at short term and long term time points (Figure 13-B). Figure 14 represents the absolute mean values of both peak Fr and peak rorFr over time.

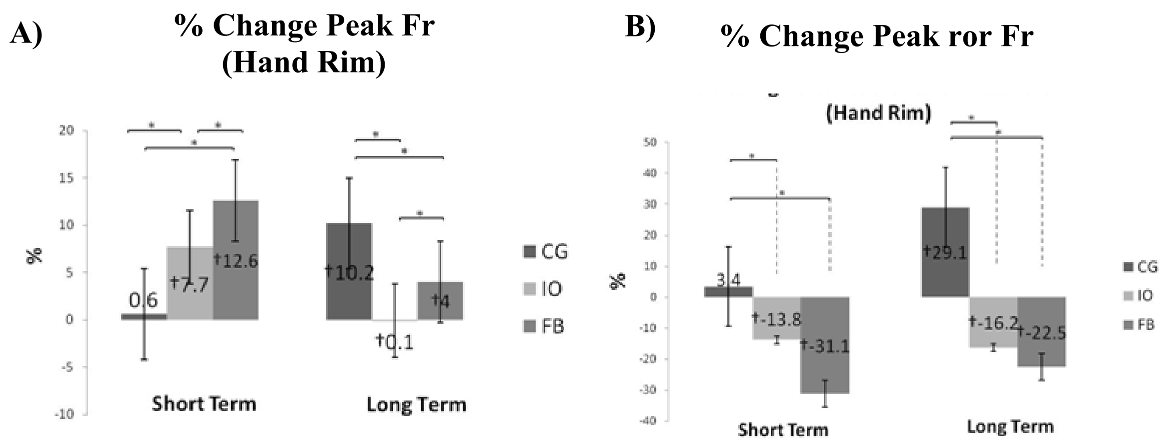


Figure 14. % change in peak ror FR and peak Fr from baseline where baseline = 0%
*** Significant change between groups p<.05,**
† Significant difference within group change from baseline (p<0.05)
% change is based off of forces normalized to velocity (F/V)

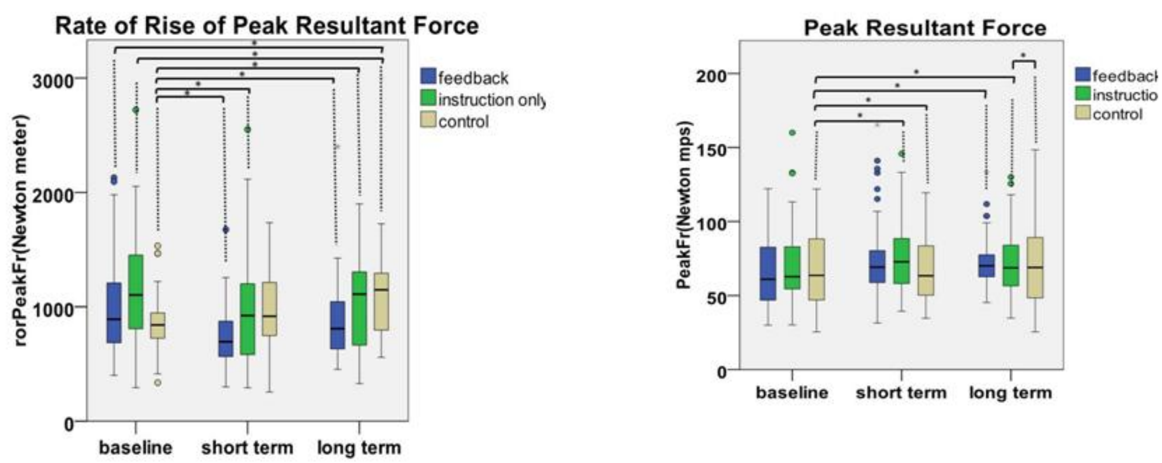


Figure 15. (Left) Box plot of change in rate of rise of peak resultant force (Newton meters) over time. Means are presented by group, time, speed condition, and averaged across all surfaces. (Right) Box plot of change in peak resultant force (Newton meters/second) over time. Forces have been normalized to velocity (F/V) and adjusted for weight, time with injury, and level of injury. * represent a significant difference between conditions p<.05.

3.4 DISCUSSION

Similar to previous training studies our treatment groups demonstrated improvements in CA and SF compared to a CG whose biomechanics remained nearly unchanged over time.[34,111] Our results stand out in that we found that the type of surface and presence or absence of a speed target did not impact individuals' ability to learn propulsion techniques. Consistent with our hypotheses the FB group displayed larger short and long term increases in CA indicating that they benefited from the addition of RFT training. While both treatment groups reduced stroke frequency considerably from baseline, contrary to our hypothesis, the IO group showed a larger long term reduction in SF than the FB group suggesting the addition of RTF may not have resulted in added benefit for that variable. It is also evident that subjects may have found transferring techniques from the dynamometer to over ground to be challenging in some instances. Another explanation is that FB subjects reported real time SF feedback was more difficult to react to than CA feedback. For example subjects felt that maximizing the contact angle feedback graph caused an automatic reduction in SF without the person having to focus on contact angle or maintaining velocity. In contrast, subjects reported the inverse of this scenario to be less natural. Primary focus on minimizing SF did not automatically lend to large CA unless subjects actively focused on maintaining velocity. In essence, the real time SF feedback portion may have required more cognitive load suggesting contact angle feedback alone may serve as a better all around training variable. Additional testing with greater number of subjects could help to confirm these points and to ensure that subjects groups baseline propulsion biomechanics are as close to equivalent as possible

Previous training studies have found resultant forces to be somewhat resistant to change. It has been suggested by other researchers that changes in force application during propulsion tend to occur either as a long-term or immediate adaptation. Specifically, Degroot et al. stated that a range of 3-7 weeks of training was either too short to illicit force related variables or that adaptations occur nearly instantaneously during the first seconds or minutes of practice where changes may escape detection. [33]

The present study found training to have varying effects on force related outcome variables. Both treatment groups showed short term increases in peak Fr that returned to near baseline levels in the long term. The IO and FB groups also showed reductions in peak rorFr while controls increased peak rorFR. The decreases in peak rorFR are likely a result of the concomitant increases seen in contact angle. In fact the FB group who displayed a larger percent increase in CA than the IO group also displayed a greater percent decrease in peak rorFR.

The short term increases in peak Fr seen in the FB group may have occurred because they needed time to adjust to the novelty of real time feedback learning. Motor learning theory has described this type of occurrence to be a natural learning process where the phenomenon of contextual interference occurs. Specifically, there is a learning strategy that occurs with random practice, initially causing receptive interference and short term performance deficits. These short term deficits then lead to better long term skill acquisition and consolidation however. [59] In essence, in the long term, treatment group members learned to apply force less rapidly and less frequently via increased contact angle and decreased stroke frequency. In contrast, control group members applied greater force more rapidly while using nearly the same stroke frequency and contact angle. These findings may suggest that even experienced wheelchair users consider technique training from time to time to refresh skills.

Interaction findings suggest that treatment subjects in the current study were able to learn and apply changes in CA and SF to all surfaces without having to rely on a target speed. This is encouraging given the diverse conditions MWU typically encounter on a daily basis. In addition, these findings support that long term wheelchair users can learn to improve skills with just video and general verbal instruction at self selected velocities. For example, the IO training method could be administered easily in a variety of settings requiring only a computer. It is also true that the motor learning theory applied to the design of the RTF software could be applied to training in a clinical setting. In fact, clinicians without access to an instrumented wheel or a dynamometer could still teach propulsion using verbal instruction based on the motor learning principles and clinical practice guidelines emphasized in the current study. A wheelchair user could be told to minimize stroke frequency, use long smooth strokes, with intermittent verbal instruction, over random ordered real life over ground practice conditions.

This study was designed to illicit improved propulsion biomechanics through a sub maximal training activity. This type of approach has also been used successfully in previous training studies to minimize training or physiological adaptation to isolate technique change. Although our study was designed to minimize work load no physiological measures were used to ensure subjects were working sub maximally which is a study limitation. It has been suggested by the American College of Sports Medicine (ACSM) guidelines that working at 30 % of heart rate reserve (HRR) can help to ensure a sub maximal effort.[111] Future studies could implement metabolic and EMG testing to more precisely monitor activity levels. In addition, EMG has been used to monitor muscle coactivity which tends to decrease as an individual's skills improve.[34]

Although subjects were able to improve their propulsion technique with video, verbal instruction, and real time visual feedback additional approaches have been used successfully.

Aspects of wheelchair set up like rear axle position can impact propulsion biomechanics however set up was held constant in the present study.[121] It was apparent that our participants presented with a range of wheelchair configurations, some conducive to propulsion and some clearly limiting. A future study should incorporate both technique training and chair set up to determine if one aspect is more critical technique than the other. In addition, strength and conditioning programs using moderate to high intensity protocols have been used to successfully improve propulsion technique.[45,111] The extent to which fitness or technique learning is more vital to propulsion remains unclear and may warrant further investigation. Certainly from a clinical point of view all of these elements should be considered when teaching wheelchair propulsion.

3.5 CONCLUSION

This study applied key principles of propulsion biomechanics, ergonomics and motor learning theory into the design of a low intensity training system teaching subjects to increase CA and decrease SF. The goal of training was to minimize potentially injurious biomechanics during over ground propulsion. Both FB and IO groups were able to improve their propulsion technique across all surfaces at both target and freely chosen speeds compared to a control group. Both treatment groups showed a short term increase in peak FR however their long term values were not significantly different from baseline. Given the large improvements seen by both training groups suggest that three weeks of simple video and verbal instruction alone may be sufficient to significantly improve many aspects of propulsion technique in long term manual wheelchair users. In addition results support that dynamometer training can translate to real life, over ground propulsion.

4.0 HAND RIM WHEELCHAIR PROPULSION TRAINING EFFECT ON INVERSE DYNAMICS PARAMETERS USING REAL TIME VISUAL BIOMECHANICAL FEEDBACK

4.1 INTRODUCTION

The previous chapters have described the theory supporting the development of our training system and the testing of the system on a group of wheelchair users during overground propulsion. The final chapter will describe the impact of training on upper extremity pain and shoulder and hand rim forces during dynamometer propulsion.

The study of forces occurring in the upper extremities during the push phase of propulsion is critical because propulsion has been identified as a primary contributor to upper extremity (UE) injury. Propulsion involves repetitive loading of the UE through an unsafe range of motion (ROM), all with relatively low gross mechanical efficiency. [7,137,138] Large forces and moments occur at the shoulder during the first half of the push phase while the humerus is extended, abducted and internally rotated, reportedly double the size of the opposing moments about the same axis(flexion, abduction, and external rotation).[139] Those experiencing higher posterior forces and internal rotation moments have been found to be more likely to have shoulder pathology as measured by MRI.[140] In addition the muscles surrounding the glenohumeral joint must generate forces to offset the moments that send the humerus upward towards the acromion during the push phase of propulsion. Uneven loading to the surrounding shoulder musculature may lead to the development of a rotator cuff muscle imbalance and then joint degeneration which can cause pain and injury.[141]

While propulsion training may be an effective preventative strategy, inherent complexities exist. For example, researchers have reported that force variables related to propulsion can be difficult to improve because of the constraints of human geometry and wheelchair user interface. For example, once the hand grasps the push rim in preparation for a stroke, arm posture becomes fixed which offers little freedom to optimize force application during a stroke.[116] It has also been suggested by multiple researchers that propelling a wheelchair with more effective force direction at the pushrim places greater demand on the shoulder musculature. [33,116] These studies indicate that to maintain high force effectiveness through the contact phase of a stroke the shoulder must handle large forces in many directions which could lead to the development of pain and injury. [33]

In an effort to reduce secondary injuries, the Consortium for Spinal Cord Medicine has approached the topic of propulsion from an ergonomics and pain and injury prevention frame of reference. Substantial ergonomics and propulsion biomechanics literature have identified specific biomechanical parameters associated with risk of injury to the upper limb. [15-17] It has been suggested that during propulsion both forces and task repetition should be kept to a minimum in order to reduce the risk of injury.[110] Wheelchair users should be encouraged to use low frequency, long and smooth strokes during the propulsive phase while allowing the hand to drift down and back below the pushrim during the recovery phase.[110]

The purpose of this study was to determine if it would be possible to minimize forces and moments at the shoulder and handrim through contact angle and stroke frequency instruction.

Two sub maximal dynamometer based wheelchair propulsion training programs were implemented. Three groups were compared: a control group (CG) that received no training, an instruction only group (IO) that reviewed a multi media instructional presentation (MMP), and a feedback group (FB) that reviewed the MMP and received additional real time feedback (RTF).

It was hypothesized that

1. The IO group would have reduced resultant forces and moments at the hand rim and shoulder when compared to the CG group immediately and 3 months after training.

2. The FB group would have reduced resultant forces and moments at the hand rim and shoulder when compared to IO group immediately and 3 months after training

3. The IO group would have a lower stroke frequency and larger contact angle when compared to the CG group immediately and 3 months after training.

4. The FB group would have a lower stroke frequency and larger contact angle when compared to the IO group immediately and 3 months after training.

4.2 METHODS

4.2.1 Subjects

All subjects provided written informed consent prior to participation in this study as approved by the institutional review board. The study population included men and women who use a wheelchair as their primary means of mobility (>80% their ambulation.), aged 18-65 who independently self-propel. Subjects were excluded if they had a spinal cord injury (SCI) above cervical 7, had a history of non-dominant traumatic upper extremity injury at the wrist or shoulder, were less than one year post SCI or had a disability that was progressive or degenerative. In addition all subjects had to use the same wheelchair throughout the entire study without any alterations in configuration.

4.2.2 Design Overview

All training procedures in the current chapter were identical to those described in the previous chapter 2(figure 16). In the present study all data was collected during dynamometer propulsion where inverse dynamics analysis could be performed.

Subjects were randomized into one of three groups (FB, IO or CG) using a random permuted block method.[135] The FB and IO groups received different training regimes as described in the study hypothesis above. The training intervention groups received one (IO group) or two training elements (FB group) depending on group assignment. The training elements consisted of:

1. Multimedia Instructional Presentation (MMP) The MMP given to both FB and IO groups was an automated instructional video and slide presentation highlighting common injuries that occur as a result of propulsion and how to use specific propulsion techniques to minimize their development. The MMP defined key learning terms like contact angle (CA) and stroke frequency (SF), described how to improve propulsion through correct application of these principles, and then emphasized the importance of using proper technique to prevent injury.
2. Real Time Feedback (RTF) Given only to the FB group, the RTF training protocol was designed to use real time SF and CA feedback to encourage subjects to take longer, less frequent strokes, to decrease force exerted at a given velocity, without causing unintended changes in force direction. The programs feedback screen presented 1) CA, 2) SF and 3) velocity (figure 17). All variables were presented randomly and discontinuously (variables ordered randomly and appear and disappear during a trial), one at a time (CA alone or SF alone) and in combination (CA with SF) Rice et al. 2010. Training was grouped into two speed categories, where a subject propelled at a freely chosen speed (self selected condition) or was given a target to hit and maintain (target speed condition). In the self selected condition a subject would propel naturally, while reacting to CA and SF feedback without any knowledge of their real time speed. The target speed condition allowed subjects to see their real time velocity with a target goal. The target provided was 2 meters per second (mps) ± .25 mps in the value of a bar range reportedly close to normal adult walking speed.[124] A thorough description of the motor learning theory aspects of this study has been presented in Rice et al 2009 (chapter one).[134]

Study Design		
Randomize Subjects		
Control Group	Instruction Group	Instruction & Feedback Group
(1) Initial Biomechanical Testing a) Dynamometer baseline testing b) Complete pain questionnaires		
(2) Training		
Rest for 10 minutes	View MMP and rest 10 minutes total	View MMP and rest 10 minutes total
Propel on Dynamometer	Practice on Dynamometer	RTF on Dynamometer
Propel on course at HERL	Practice on course at HERL	Practice on course at HERL
(3) Repeat Biomechanical Testing		
Repeat step (2) ten days after initial testing		
Repeat steps (2) twenty days after initial testing		
Repeat step (1) three months later		

Figure 16. Description of study design

4.2.3 Training Time Frame

FB and IO subjects visited the lab four times over a three month period while control group (CG) subjects visited the lab 3 times over a three month period. Therefore, the time between baseline and final testing was equal for all subjects. To minimize drop out, we required only 3 lab visits for CG members who propelled the same amount of time as treatment subjects per visit but received no training. The protocol was carefully designed to ensure that all subjects' propulsion and rest times were equivalent regardless of group membership (Figure16). For treatment subjects (IO & FB) the first three visits included training and testing. Training occurred on V1, V2 (10 days after V1), and V3 (10 days after V2). The fourth visit occurred three months after V1 and involved only testing on the dynamometer. All training was intended to occur at low intensities to isolate technique learning and to minimize physiological adaptation as described by De Groot et al.[34] For example, the longest period of time spent in propulsion without a break was 55 seconds. In that 55 sec, a FB subject would propel continuously while receiving visual feedback (CA, SF, and velocity) which would appear and disappear during the trial. All combinations of variables presented equated to 12 minutes of active propulsion with a total rest time of 14 minutes.

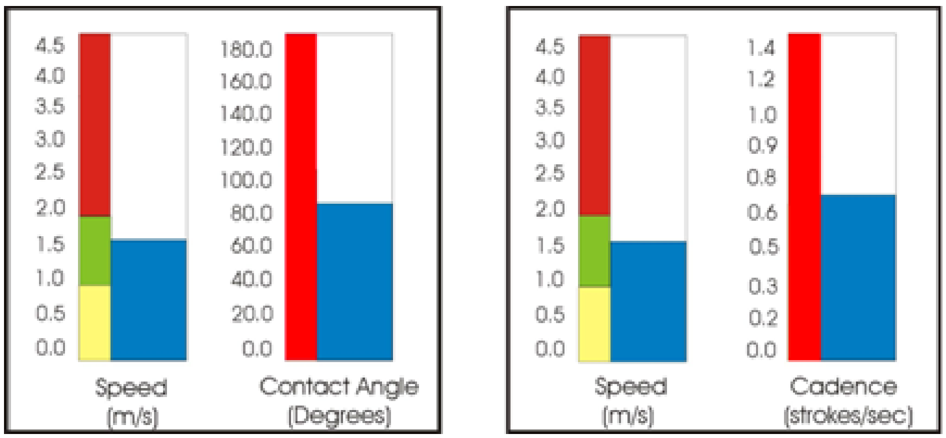


Figure 17. Sample real time feedback display

4.2.4 Testing Procedures

Biomechanical assessment of wheelchair propulsion was performed three times: upon enrollment (baseline), immediately after visit 1 training (short term) and three months post baseline (long term). During these assessments, subjects were tested separately at a self selected pace and with a target velocity. For the target speed condition a subject was instructed to manually propel his or her wheelchair at a target speed of 1.5m/s on a dynamometer for thirty seconds. The testing target was different from the training target speed to test generalizability. During the self selected condition subjects could see their real time velocity, however without a target. In both conditions, recording began after steady state velocity was achieved which included kinetic and motion analysis data collection.

4.2.5 Pain Assessment

Two self administered pain questionnaires were given to subjects prior to propulsion trials on each visit to the lab. These tests included the Wheelchair Users Shoulder Pain Index (WUSPI) and the Carpal Tunnel Syndrome (CTS) questionnaire Severity of Symptoms (SS) and Functional Status Scales(FS).[4,142] A WUSPI final score ranges from 0 (least pain) to 150 (worst pain possible). The test consists of 15 questions, answered on a scale from 0(no pain)-10(worst pain). The mean of the 15 questions is then multiplied by 15 to achieve a final score. The SS component of the CTS scale is scored by taking the mean of 11 questions rating wrist pain from 1(mild) to 5(most severe). The total FS score was the average of eight questions asking subjects to rate pain experienced during functional activities on a scale of 1(no difficulty performing activity) to 5(unable to perform an activity). These questionnaires were administered to investigate the extent to which study interventions may have had an impact on pain and were included as part of a data safety monitoring plan.

4.2.6 Instrumentation

The participant's wheelchair was fitted bilaterally with two instrumented wheels (SMARTwheel®; Three Rivers Holdings, Mesa, Az) while data was collected from the side of the non-dominant U/E as it may be less affected by pathology not related to wheelchair propulsion. [123] The instrumented wheel allows one to accurately record the three orthogonal components (x, y and z) of the forces and moments applied directly to the wheelchair pushrim during propulsion within a local coordinate frame at a sampling frequency of 240 Hz.[89] The instrumented wheel does not alter the feel or set up of participants own wheelchair and it closely replicates usual wheel's position and orientation when mounted. While propelling, key propulsion variables collected were streamed as real time visual feedback and presented on a large monitor facing the participant. There was no perceivable delay between the actual action on the hand rim and the feedback received visually on the screen. In addition, the participant's wheelchair configuration was noted and maintained throughout the entire study and the Smart wheels were equipped with solid tires eliminating the need to monitor tire pressure.

During training and testing, the participant's wheelchair was positioned over a custom-built computer-controlled wheelchair dynamometer and anchored to the floor using a four-point tie-down system. The dynamometer was comprised of an independent double drum system. The rolling resistance of the dynamometer was fixed at 14.2 N, reportedly just under that of rolling on low pile carpeting.[125,126]

4.2.7 Kinematic Data

Two OPTOTRAK 3020 motion capture systems were used for testing. This system is capable of outputting three-dimensional marker position data relative to a global origin located between the two rollers of the wheelchair dynamometer. The marker set included markers at the third metacarpophalangeal joint, radial styloid, ulnar styloid, lateral epicondyle, acromion, C7 vertebrae and greater trochanter. All kinematic data was down sampled at 60 Hz and digitally filtered with a 4th order zero-phase low pass Butterworth filter with a 7 Hz cutoff frequency.

4.2.8 Inverse Dynamics

Mercer et al. previously described the anthropometric model used for this study.[140] Segment lengths and upper extremity circumferences of all subjects were measured and input to Hanavan's mathematical model which calculates the inertial properties of each body segment.[143] Pushrim forces were transformed to the glenohumeral joint using the previously described inverse dynamics model. Calculations for the model were performed using Matlab where shoulder joint forces were transformed to the anatomical coordinate system of the proximal segment of the shoulder joint, the trunk, as follows: anterior(+x), posterior(-x), superior(+y), inferior(-y), medial(+z), and lateral(-z). Shoulder joints moments were calculated relative to the humeral local coordinate system described in previous work .[144] The humeral and trunk local coordinate systems are coincident when the arm is in a neutral posture. Abduction (+) and adduction (-) moments occurred about the x-axis, external (+) and internal (-) rotation produced moments about the y-axis and extension (+) and flexion (-) moments occurred about the z-axis.

4.2.9 Statistical Analysis

A 3 x 1 x 2 x 3 General Linear Mixed Model (GLMM)was used with an unspecified covariance matrix.[145] All of the main effects and the 2-way interactions within groups were included in the model. The between-subject variables were grouped having three levels (FB, IO and CG), surface having 1 level (dynamometer) and speed having two levels (self selected and target). Since subjects were measured repeatedly, observations were nested within individuals for the 3 time points (baseline, short term, and long term). Weight, time since injury, and level of injury were added to the model as covariates. Follow up comparisons consisted of a priori contrasts and post hoc pair wise comparisons which were performed using Bonferroni adjustments to correct for inflation of Type I error. Alpha for all analysis was set at .05. A Shapiro-Wilk test was used to test for normality and the Box Cox test was used to measure linearity.

4.2.10 Data Analysis

Data analysis included changes in contact angle, stroke frequency and the forces occurring during the push phase of propulsion. Peak resultant forces were recorded at the hand rim and shoulder. Shoulder component forces and moments were also modeled. Since propulsion forces were found to be highly correlated with velocity, all forces were normalized to velocity [F (Newton)/Velocity (mps)].

4.3 RESULTS

4.3.1 Demographics

Short term comparisons were based on analysis of 27 subjects enrolled in the study. Short term observations compared visit one baseline trial to the first propulsion trial post training the same day. Subjects included 3 female and 24 males randomized into Feedback (FB), Instruction only (IO) and Control group (CG) categories. Subject characteristics are shown in Table 5.

Due to subject drop out, long term observations were based on 17 subjects. Drop out, or missing data were due to scheduling conflicts and equipment malfunction and not complications associated with the training interventions. However, an independent sample T test was performed comparing the baseline demographic characteristics of those who stayed in to those who dropped out. Significant differences were not found between the demographic characteristics of those who stayed in and those who dropped out therefore data imputation was not warranted (Table 6).

4.3.2 Demographics and Biomechanics

An estimate of fixed effects analysis was performed to predict changes in peak resultant forces occurring at the handrim with demographic characteristics. Older and heavier subjects tended to use greater Fr, while lower level injured subjects used less Fr, $F(1,175.8)=7.9, p=.005, \beta=0.35$; $F(1,177.8)=93.8, p=.001, \beta=0.33$; $F(1,173.8)=20.7, p=.001, \beta=-0.82$.

Table 5. Participant Data

	Short Term(same day) Subject Characteristics				Long term(3 month) Subject Characteristics			
Group	**FB**	**IO**	**CG**	**Total**	**FB**	**IO**	**CG**	**Total**
N	9	9	9	27	7	5	5	17
Age (yrs)	36.9(13.7)	37.5(10.3)	46.7(15.2)	40.0(13.4)	41.0(14.4)	38.0(6.9)	49.0(17.1)	42.2(13.5)
Years Since Injury (yrs)	19.9(11.0)	15.0(11.9)	19.1(10.1)	18.0(10.9)	21.4(10.4)	12.8(11.4)	21.2(9.7)	18.8(10.6)
Height (m)	1.75(0.07)	1.67(0.14)	1.80(0.08)	1.74(0.11)	1.75(0.04)	1.65(0.14)	1.76(0.09)	1.72(0.10)
Weight (kg)	76.1(19.7)	82.9(23.7)	84.3(8.3)	80.4(18.8)	82.7(19.8)	83.5(22.9)	82.4(4.9)	82.4(4.9)
Injury Level	T2	T8	T10	T7	T2	T4	T10	T6
Gender(M/F)	8/1	8/1	8/1	24/3	7/0	4/1	4/1	15/2

Values represented as means and standard deviations, injury level is presented as median

Table 6. Drop Out Data

Drop Out Change (T -Test)		Characteristics of Subject Who Did not Drop Out				Characteristics of Subject Who Dropped Out			
	Group	**FB**	**IO**	**CG**	**Total**	**FB**	**IO**	**CG**	**Total**
	n	7	5	5	17	2	4	4	10
p=1.4	**Age (yrs)**	41.0(14.4)	38.0(6.9)	49.0(17.1)	42.2(13.5)	28.3(8.4)	36.8(14.8)	43.5(14.0)	33.2(12.4)
p=0.3	**Years Since Injury (yrs)**	21.4(10.4)	12.8(11.4)	21.2(9.7)	18.8(10.6)	16.3(14.0)	17.7(13.7)	15.6(11.9)	17.1(12.6)
p=0.8	**Height (m)**	1.75(0.04)	1.65(0.14)	1.76(0.09)	1.72(0.10)	1.6(0.09)	1.6(0.1)	1.8(0.07)	1.6(0.1)
p=1.4	**Weight (kg)**	82.7(19.8)	83.5(22.9)	82.4(4.9)	82.4(4.9)	57.3(6.5)	82.3(28.2)	87.5(13.0)	71.6(24.3)
p=1.9	**Injury Level**	T2	T4	T10	T6	T2	T5	T4	T4

Values are represented as means and standard deviations, injury level presented as median.
T-test results represent statistical differences between drop out and non drop subject characteristics.

4.3.3 Assessment of Pain

Subjects reported extremely low levels of pain in the upper extremities on the Wheelchair User Shoulder Pain Index (WUSPI) and the Carpal Tunnel Severity of Symptoms and Functional Status Scale. A paired t-test did not reveal a significant main effect change in pain status from baseline to long term (p=.2), (Table 7).

Table 7. Assessment of Pain

	Severity of Symptoms Scale [1(mildest pain) - 5(most severe pain)]				Functional Status Scale [1(no difficulty) - 5(can't perform)]				Wheelchair User Shoulder Pain Index [0(no pain) - 150(worst pain)]			
Group	**Baseline**	**n**	**Long Term**	**n**	**Baseline**	**n**	**Long Term**	**n**	**Baseline**	**n**	**Long Term**	**n**
CG	1.5(0.4)	9	1.2(0.4)	5	1.0(0.1)	9	1.0(.05)	5	8.4(10.4)	9	10.3(18.3)	5
IO	1.2(0.3)	9	1.2(0.4)	5	1.3(0.7)	9	1.1(0.2)	5	2.3(2.9)	9	7.2(12.4)	5
FB	1.6(0.5)	9	1.5(0.6)	7	1.2(0.5)	9	1.3(0.4)	7	7.8(10.1)	9	8.2(13.7)	7

All scores reported as mean (SD)

4.3.4 Self Selected Speed vs. Target Speed Conditions

An ANOVA revealed that subject's propelled faster with a target velocity (1.59mps) then during self selected propulsion (1.37 mps), p=.03 (figure 3). However the 3-way interaction of group (FB, IO and CG) x speed (self select and target) x time (baseline, short term, and long term), was not found to be statistically significant (p>.05). This meant that all changes in propulsion dependant variables occurring at a self selected velocity were highly correlated to those occurring at a target velocity. Consequently all statistical analysis is based on the average of the self selected and target speed conditions. Descriptive reports of data separated into self selected and target velocity conditions are provided in the appendix.

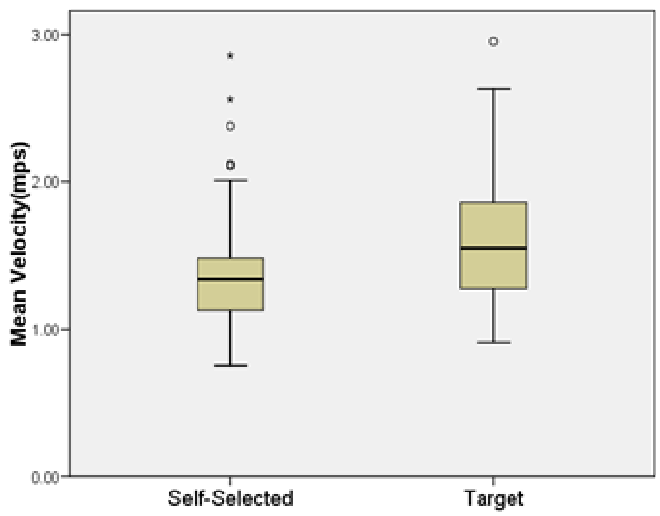
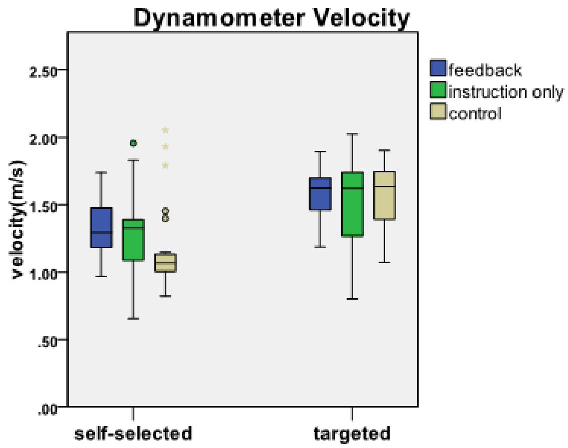


Figure 18. Self selected vs. target velocity



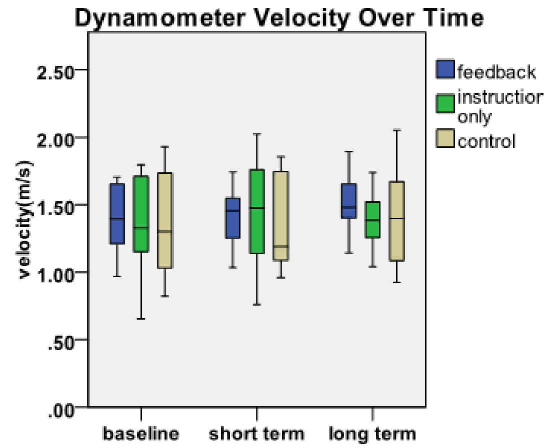


Figure 19. (Left) Descriptive box plot of velocity separated by speed condition (target & self select) and group averaged across all time points p>.05. (Right) Descriptive box plot of the self selected and target speed conditions averaged together, separated by group, and displayed over time, p>.5.

4.3.5 Resultant Force

A statistically significant group by time interaction was found in Fr at the hand rim only (table 9). Values in table 8 appear as adjusted means and standard errors and forces have been normalized to velocity [(Newton)/ Velocity (mps)].Table 8 shows mean changes in Fr while figure 20 shows the % of change. IO and FB members displayed significant short term increases from baseline p=.001, p=.008 that were significantly different from the change in the Control group but not from each other (table 8). The FB group also showed greater % change than the IO group (figure 20). In addition, control group members displayed a significant long term increase in Fr at the hand rim (p=. 001) that was significantly different from the long term change seen in the IO group p=.02 but not the FB group, p=.16. The CG's long term change was greater than the FB and IO groups changes (figure 20).

Table 8. Resultant Force Adjusted Means

Subject group	time	Fr Hand rim (N/mps)	Std. Error	Fr Shoulder (N/mps)	Std. Error
Control Group	baseline	69.5	2.8	56.5	5.6
	short term (same day)	69.8	2.6	50.4	4.9
	long term (3 months)	* B **77.6**	3.3	52.5	3.6
Instruction Only	baseline	71.7	2.6	63.3	5.8
	short term (same day)	* A **77.2**	2.4	58.9	5.4
	long term (3 months)	71.1	2.9	55.0	3.8
Feed Back	baseline	68.8	2.6	58.4	5.4
	short term (same day)	* A **77.5**	2.4	55.4	4.3
	long term (3 months)	71.7	3.1	56.9	3.3

*(within comparison) Indicates a statistically significant change from baseline $p<.05$,
[A] (between comparison) Indicates statistically significant difference from control group ($p<0.05$)
[B] (between comparison) Indicates significant difference from treatment groups (FB or IO)
Bonferroni Correction has been used to adjust for multiple comparisons
Covariates appearing in the model are weight, time with injury and level of injury
N/mps= Newton meters per second

Table 9. Resultant Force Interaction Effects

Group x Time Interaction effects	Significance	F score
Fr Handrim	***p=.002**	$F(4,162.2)=4.3$
Fr Shoulder	p=.94	$F(4,45.6)=0.1$

* indicates a significant difference between FB, IO and CG groups over time $p<.05$
F score= F ([df_between], [df_within]) = f score statistic

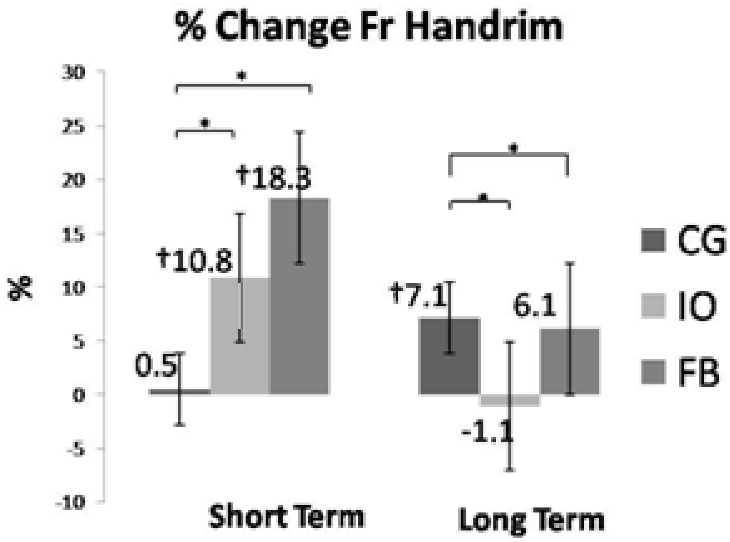


Figure 20. %change Fr at the handrim from baseline
*** Significant change between groups (p<.05)**
† Significant difference from baseline (p<0.05)

4.3.6 Contact Angle & Stroke Frequency

A significant group x time interaction was found for both contact angle and stroke frequency p=.0001, p=.0001(table 11). The FB and IO groups displayed significant short and long term changes in both contact angle and stroke frequency p<. 0001, (table 10). These changes were also significantly different from the changes seen in control group for both the short and long term time points p<.001. The FB and IO groups long term changes in contact angle and stroke frequency were significantly different from each other however their short term changes were not p=.01, p=.02, p=.32, p=.1(table 10). As seen in Figure 21-22, the FB group showed greater short and long term mean and percent changes in CA than the IO group whose changes were larger than the control group. The IO group however showed greater reductions in SF than the FB group (figure 21-22).

Table 10. Contact Angle & Stroke Frequency Adjusted Means

Group Dynamometer Target Speed (1.5mps)	Baseline	Short Term	Long Term(3month)
Contact Angle (º)			
Control (CG)	97.8SE2.3	95.4SE2.4	95.9SE 2.3
Instruction (IO)	93.3SE2.2	* A 100.6SE2.2	* A, B 103.7SE2.1
Feedback (FB)	96.6SE2.1	* A 105.5SE2.2	* A, B 108.5SE2.2
Stroke Frequency (Strokes/Sec)			
Control (CG)	1.0 SE 0.03	1.1 SE 0.04	1.1 SE0.05
Instruction (IO)	1.2 SE 0.03	* A 0.91 SE 0.03	* A,B 0.94 SE 0.04
Feedback (FB)	1.0 SE 0.03	* A 0.82 SE 0.03	* A,B 0.86 SE 0.04

*(within comparison) indicate statistically significant changes in contact angle and stroke frequency ($p<.05$) from baseline

A (between comparison) Indicates significant difference ($p<0.05$) from Control group

B (between comparison) Indicates significant difference between treatment groups (FB from IO)

Covariates appearing in the model include weight, time with injury and level of injury.

Table 11. Contact Angle and Stroke Frequency Interaction Effects

Group by Time Interaction effect	Significance	F score
Contact Angle	***p=.0001**	F(4,163.7)=27.5
Stroke Frequency	***p=.0001**	F(4,164.2)=23.9

* Significant difference between FB, IO and CG groups over time from baseline. p< .05
F score= F ([df_between], [df_within]) = f-score statistic

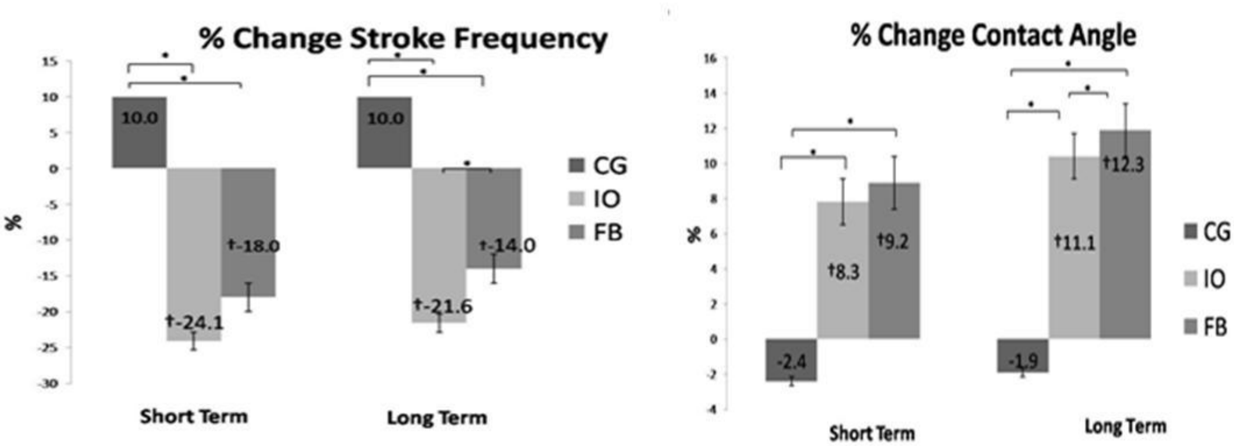


Figure 21. % change in contact angle and stroke frequency from baseline
*** Significant change between groups (p<.05)**
† Significant difference from baseline (p<0.05)

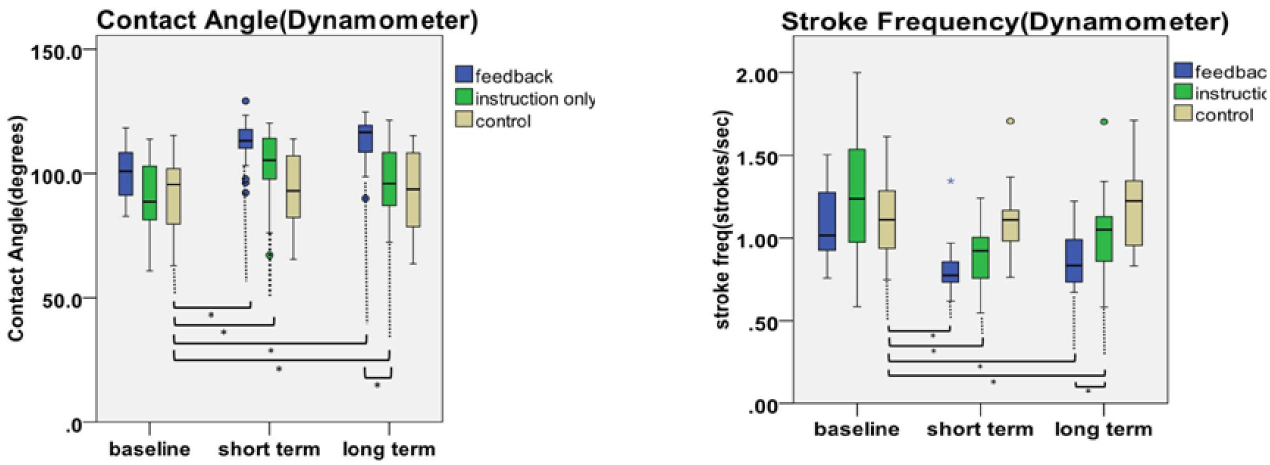


Figure 22. (Left) Box plot of change in contact angle (degrees) over time. Means are presented by group, time, and averaged across self selected and target speed conditions. (Right) Box plot of change in stroke frequency (stokes/second) over time. Means are presented by group, time, and averaged across self selected and target speed conditions. * represent a significant difference between conditions p<.05.

4.3.7 Component Forces and Moments at the Shoulder

Group by time interactions for the shoulder appear in table 8. All mean component forces and moments with p values are presented in table 9 while statistically significant findings are described below and appear in figures 6a-d.

4.3.7.1 Posterior Force (Min Fx) figure 6(a)

A significant group x time interaction was found in posterior force at the shoulder p=.01(table8). FB group members displayed significant short and long term increases from baseline p=.001, p=.03 that were not significantly different from the IO group, p=.26(table8). The short term change from baseline was significantly different from those seen in control group however the long term change was not p=.001, p=.12(table 9).

4.3.7.2 Inferior Force (Min Fy) figure 6(b)

A significant group x time interaction was found in inferior force at the shoulder p=.04, (table 8). The FB and IO groups displayed significant short term increases from baseline p=.003, p=.001, (table 9). Both groups (FB & IO) changes were not significantly different from the control group p=.1, p=.15, however they were significantly different from each other, p=.002 (table 9).

4.3.7.3 Maximum Medial Force (Max Fz) figure 6(c)

A significant group x time interaction was found in medial force at the shoulder p=.02, (table 8). The FB group displayed a significant short term increase from baseline p=.001, that was significantly different from both the control group p=.002 and the IO group p=.07(table9). These finding should be interpreted with caution because the IO and FB groups baseline measures were found to be significantly different from the CG (p<.05).

4.3.7.4 External Rotation Moment (Max My) figure 6(d)

A significant group x time interaction was found in external rotation moment at the shoulder p=.02, (table 8). The IO group displayed a significant short term increase from baseline, p=.01 that was significantly different from the control group p=.005, and approaching significance from the FB group p=.06(table 9).

Table 12. Shoulder Interaction Effects

Group x Time Interaction Effect	Significance	F score
Anterior Force (maxFx)	p=0.15	F(4,38.1)=0.15
Posterior Force (minFx)	***p=0.01**	F(4,35.5)=3.7
Superior Force (max Fy)	p=0.4	F(4,41.0)=0.93
Inferior Force (min Fy)	***p=0.04**	F(4,34.8)=2.7
Medial Force (maxFz)	***p=0.02**	F(4,19.4)=3.6
Lateral Force (minFz)	p=0.14	F(4,37.6)=1.8
Abduction Moment(maxMx)	p=0.16	F(4,40.9)=1.7
Adduction Moment (minMx)	p=0.49	F(4,40.8)=0.85
External Rotation Moment (maxMy)	***p=0.02**	F(4,38.5)=0.08
Internal Rotation moment (minMy)	p=0.3	F(4,33.5)=0.30
Flexion Moment(maxMz)	p=0.15	F(4,50.3)=1.7
Extension Moment(minMz)	p=0.45	F(4,38.4)=0.45

* Significant difference between FB, IO, and CG groups over time. $p < .05$
F score= F ([df between], [df within]) = f-score statistic

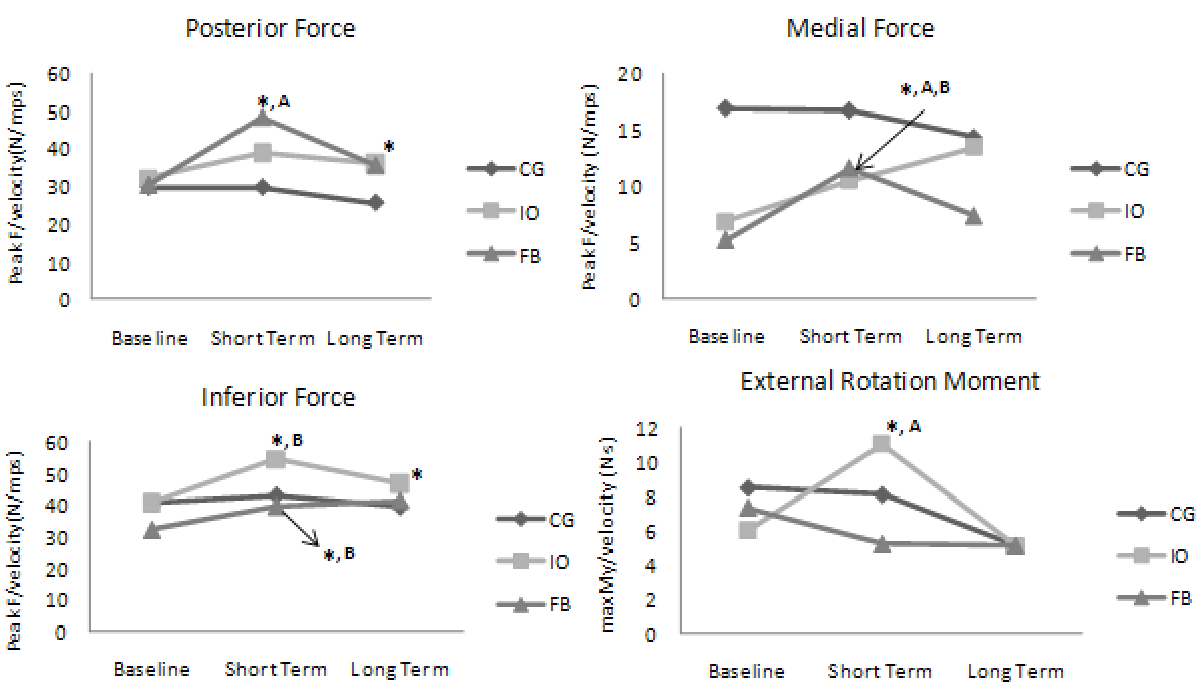


Figure 23. Statistically significant changes in forces and moments at the shoulder from baseline
*** Significant change from baseline p<.05,**
[A] significant difference from control group (p<0.05)
[B] significant difference from treatment groups (FB or IO)

Table 13. Shoulder Forces and Moments Adjusted Means

Forces	FB			IO			CG		
	Baseline	Short Term	Long Term	Baseline	Short Term	Long Term	Baseline	Short Term	Long Term
Anterior Force (N/mps)	7.7 SE 2.3	9.1.1 SE 2.5	6.3 SE 3.6	10.9 SE 2.5	15.1 SE 2.8	18.6 SE 4.2	13.0 SE 2.5	11.7 SE 2.8	18.3 SE 4.3
***Posterior Force (N/mps)**	30.4 SE 3.0	*** A 48.2 SE 3.8(p=.001)**	***35.5SE 3.6(p=.03)**	32.2 SE 3.3	39.0 SE 4.3	36.2 SE 4.1	29.8 SE 3.4	29.8 SE 4.2	25.6 SE 4.4
Superior Force (N/mps)	18.6 SE 4.1	30.3 SE 5.4	16.1 SE 4.8	14.1SE 4.5	23.7 SE 6.2	13.8 SE 5.4	10.2 SE 4.6	10.3 SE 6.1	15.2 SE 5.7
*** Inferior Force (N/mps)**	32.3 SE 3.6	*** B 39.7 SE 3.3(p=.003)**	41.5 SE 4.6	40.7 SE 4.0	*** B 54.4SE 3.7(p=.001)**	46.9 SE 5.4	40.7 SE 4.0	43.0 SE 3.7	39.6 SE 5.6
***Medial Force (N/mps)**	5.2SE 2.5	*** A, B 11.6 SE 2.9(p=.001)**	7.3 SE 1.9	6.8 SE 2.8	10.5 SE 3.2	13.5 SE 2.2	16.9 SE 2.7	16.7 SE 3.2	14.3 SE 2.4
Lateral Force (N/mps)	13.4SE 1.6	19.3 SE 1.6	36.2 SE 5.3	17.7 SE 1.8	20.7SE 1.8	8.0 SE 6.4	17.2 SE 1.8	19.7 SE 1.8	22.7 SE 6.3

Moments	FB			IO			CG		
	Baseline	Short Term	Long Term	Baseline	Short Term	Long Term	Baseline	Short Term	Long Term
Abduction Moment (N·s)	9.6 SE 1.9	6.6 SE 1.6	6.2 SE 1.7	4.7 SE 2.3	9.4 SE 1.8	4.7 SE 2.3	7.0 SE 2.2	7.0SE 1.7	5.2 SE 1.3
Adduction Moment (N·s)	5.1 SE 0.9	6.1 SE 0.9	3.2 SE 0.9	2.8 SE 1.1	5.8 SE 1.0	4.1 SE 1.1	2.3 SE 1.0	3.0 SE 1.0	2.7 SE 1.1
Ext. Rot. Moment (N·s)	7.3 SE 1.9	5.2 SE 1.8	5.1SE 1.0	6.0 SE 2.2	*** A 11.0 SE 2.0 (p=.01)**	5.1SE 1.1	8.5 SE 2.1	8.1 SE 2.0	5.1 SE 1.2
Int. Rot. Moment (N·s)	16.2SE 1.6	19.9 SE 1.4	15.4 SE 1.8	13.0 SE 1.9	16.9 SE 1.5	18.2 SE 2.1	9.2 SE 1.8	10.1 SE 1.5	9.8 SE 2.2
Flexion Moment (N·s)	9.6 SE 1.9	6.6 SE 1.6	6.2 SE 1.7	4.7 SE 2.3	9.4 SE 1.8	4.9 SE 1.3	7.0 SE 2.2	7.0 SE 1.7	5.2 SE 1.3
Extension Moment (N·s)	19.0 SE 2.3	22.3 SE 2.0	25.1 SE 1.7	13.4 SE 2.6	18.4 SE 2.3	14.7 SE 2.0	10.4 SE 2.6	11.4 SE 2.2	12.5 SE 2.1

* (within comparison) Indicates a statistically significant change from baseline p<.05,
A (between comparison) Indicates statistically significant difference from Control group, p<0.05
B (between comparison) Indicates significant difference from treatment groups (FB or IO)
Bonferroni Correction has been used to adjust for multiple comparisons
Covariates appearing in the model are weight, time with injury, and level of injury
(N·s) = Newton seconds
N/mps=Newton meters per second

4.4 DISCUSSION

This study was designed to train subjects' sub maximally in order to isolate technique changes in propulsion technique. Unlike a majority of propulsion training studies our participants were long term wheelchair users propelling in their own personal wheelchairs.[33,111,128,146,147] All data were collected during dynamometer propulsion to capture motion analysis and to control for the presentation real time feedback training variables. While over ground propulsion is typically favored to simulated propulsion environments, a previous training study comparing the two methods found that over ground propulsion did not result in improved learning over ergometer training.[146]

The present study found that intervention groups (FB & IO) demonstrated favorable changes in propulsion biomechanics with some short term increases in forces at the shoulder. While controlling for weight, time with injury, and level of injury both treatment groups demonstrated a long term increase in contact angle (CA) and reduced stroke frequency (SF), while peak resultant forces (Fr) at the hand rim, and shoulder remained nearly unchanged. These findings were consistent with a similarly designed training study where treatment subjects reduce SF without changes in resultant forces.[34] In addition, similar to the over ground study Ch 2, FB subjects showed greater percent change than IO subjects in CA but not SF suggesting real time feedback combined with video may be more effective means of training then video alone for that variable. In contrast control group subjects CA and SF stayed the same while there long term Fr at the hand rim increased significantly. The CG long term increase in handrim Fr was unexpected however their peak resultant shoulder forces remained low and unchanged from baseline to three months. Again this increase in Fr may suggest that propulsion biomechanics may not be stable over time in the absence of technique training.

While treatment subjects did not reduce forces after training, they applied nearly equivalent amounts of Fr to the handrim however using more of the push rim (contact angle) with reduced frequency. This finding may be of importance as previous studies have shown that individuals are more susceptible to injury when they propel with more frequent wheel contact of shorter duration per stroke.[148]

Contrary to our hypothesis, FB and IO group members displayed short term increases in shoulder forces and peak resultant forces at the handrim immediately after training however; these changes did not persist to the long term visit. For example both treatment groups displayed short term (same day) increases in peak Fr at the hand rim with the FB group producing a greater percent increase than the IO group. The FB group also produced a significant short term increase in posterior force at the shoulder which was worrisome because this specific force has been linked to injury in the past.[140] This may have been related to the fact that initially, FB subjects were engaged in a more complex learning task and needed time to adjust to the random presentation of real time feedback graphics. Motor learning theory has described this type of occurrence as a natural learning process where the phenomenon of contextual interference occurs. Specifically that there is a learning effectiveness associated with random practice, initially causing receptive interference that results in short term performance deficits but eventually leads to better long term skill acquisition and consolidation.[59] Our subjects followed this trend however it is important to note that pain levels remained low and unchanged throughout the entire study. To minimize the likelihood of injury, clinicians should be aware of this tendency towards short term performance deficits when an individual is learning a new skill. The literature has even suggested that in the early stages of learning subjects may benefit more from a low intensity, conservative training regime where a stationary ergometer is favored to over ground.[146]

An additional study component included the modeling of peak forces and moments at the shoulder. These measurements served as a safety precaution to ensure that training did not produce harmful or unintended changes that would have otherwise gone unnoticed. An additional safety check involved comparing peak forces produced over ground to those occurring on the dynamometer. Subjects did in fact display a similar trend over ground were treatment groups showed short term increases in handrim Fr. In addition, as seen in tables 4(chapter 2), over ground peak Fr at the handrim was slightly larger but comparable to forces produced during dynamometer propulsion. Consequently the forces modeled at the shoulder were representative of those actually occurring during over ground propulsion. This finding was unexpected because studies have shown that the forces produced during dynamometer propulsion tend to be greater than an equivalent task performed overground.[149] In addition the inertial forces acting on a wheelchair due to acceleration and deceleration of the trunk and arms are neglected when a wheelchair is strapped down.[149,150] These differences tend to be more pronounced during start up or acceleration periods, however all data in the present study was collected during steady state propulsion which could explain these findings.

Propulsion velocity was another critical aspect of this study impacting the studies design and data analysis. For example, from a technique stand point an easy way to reduce stroke frequency and peak hand rim forces would be to simply slow down which would not result in fewer cycles to reach a given location. A target speed condition was implemented to address this concern although subjects' velocities still tended to fluctuate from visit to visit even when a target was provided. Because propulsion biomechanics have been shown to vary with propulsion speed, it was necessary to normalize all forces to velocity [F (Newton)/Velocity (mps)]. [151,152] For example faster speeds have been associated with increases in shoulder forces and moments and maximum forces at the wrist have been shown to differ with changing speeds.[151,152] Normalized forces allowed for clearer inspection of peak forces occurring at different speeds. Self selected trials were also included because it is likely that the way an individual chooses to freely propel is most related to the development of pathology. In addition, the literature suggests that training at a specific velocity lends to skill acquisition at that trained speed only.[153] Thus combination velocity training can help an individual to generalize a skill (propulsion technique) thereby improving performance at different speeds.[153] This was confirmed because no differences in performance were observed based on the presence or absence of a velocity target. This was encouraging because subjects were able to successfully modify their propulsion technique while maintaining velocity with and without a pace target.

Although subjects were able to improve their propulsion technique with video and realtime visual feedback other approaches have been used successfully. Aspects of wheelchair set up like rear axle position can profoundly impact propulsion technique however wheelchair set up was held constant in this study.[121] It was apparent that study participants presented with a range of wheelchair configurations, some conducive to propulsion and some clearly limiting. Future studies should incorporate both technique training and chair set up to establish a clearer understanding. In addition, subject's baseline propulsion biomechanics varied considerably due to natural ability and experience. Although subjects were randomized into groups, their mean baseline levels were not always equivalent do to sample size which was an additional study limitation. In addition other forms of training that include strength and conditioning programs and moderate to high intensity protocols have been used to successfully improve propulsion technique .[147,154] The extent to which fitness or technique learning is more vital to propulsion remains unclear and warrants further investigation. Certainly from a clinical stand point all of these elements should be considered when teaching wheelchair propulsion.

4.5 CONCLUSION

Our study showed that through training long term manual wheelchair users were able to modify their propulsion biomechanics favorably over a three month period. FB subjects receiving real time feedback and video training displayed more pronounced changes in CA and SF than IO subjects receiving video training alone. Both treatment groups (FB and IO) displayed nearly equivalent long term peak Fr at the shoulder and handrim that did not change significantly from baseline. FB subjects also displayed some temporary increases in shoulder forces however they dropped back down to baseline levels upon completion of the study. Pain status remained low and unchanging for all subjects throughout the entire study. The fact that the IO treatment group benefited significantly from training suggests that a low cost easy to administer instructional presentation can still serve as an effective training tool. Future study should include a greater number of wheelchair users with a broader range of disabilities to confirm the effectiveness of training.

5.0 CONCLUSION

The goal of this dissertation was to develop and test a manual wheelchair propulsion training system based on literature related to propulsion biomechanics, ergonomics, real time visual feedback and motor learning theory. The first study described the development of the real time feedback (RTF) component of our training system. Because an underlying goal of this study was to create a system that could eventually be used clinically a great deal of research went into its design and development. First, with injury prevention in mind we decided to encourage a propulsion technique where both stroke frequency and propulsion forces were minimized which are concepts supported by Clinical Practice Guidelines (CPG), propulsion biomechanics and ergonomic literature. [15-17,31,37,94,96,110,121,155] Although many training methodologies exist we felt that real time visual feedback could be a particularly effective way to reinforce these concepts because RTF has been used successfully to train skills.[33]

Initially, we designed the system to be flexible because we did not know how many variables to present, which to select, or even the most effective way to present them. Many of these concepts have been overlooked or selected without a great deal of scientific basis in previous training studies which was a motivating factor behind the design of this study. Next we found through a review of motor learning literature that individuals learn best from RTF when the number of items presented is kept to a minimum, are viewed discontinuously, and are presented in random order.[59,62,77,118,120] Contact angle(degrees), velocity (m/s), and stroke frequency (strokes per second) were chosen as training variables because they are closely aligned with the CPG goals and have been shown to have a strong association with the development of upper limb pain and injury. [31,37,96,110,121,155] This system was then tested on one long term manual wheelchair user on a flat tile surface who showed marked improvements in CA, SF and other force related variables after three weeks of training.[134] The protocol was refined slightly based on these results and rigorously tested on a large group of experienced wheelchair users with a number of safety checks in place to guard against potentially harmful unintended changes.

In the second study, we tested the training systems ability to modify CA and SF over a diverse over ground course. Three groups were compared: a control group (CG) that received no training, an instruction only group (IO) that reviewed a multi media instructional presentation (MMP), and a feedback group (FB) that reviewed the MMP and received additional real time feedback (RTF). The purpose of the MMP was to convey the same training concepts as the RTF system but in an easy to access package that if proven effective had the potential to be easily disseminated. Both training systems allowed for independent learning as well.[134] In essence the MMP could be thought of as the primary training system with the RTF serving as a high tech supplement which provided additional focused reinforcement.

Both FB and IO groups were able to improve their propulsion technique across all surfaces at both target and freely chosen speeds compared to the control group. It was also evident that the FB group having received additional training with RTF and MMP displayed larger increases in CA than the IO group who received MMP alone. While both training groups decreased SF, the IO group displayed a greater reduction than the FB group. Both treatment groups showed a short term increase in FR however their long term values were not significantly different from baseline. Given the large improvements seen in both training groups suggest that three weeks of simple video and verbal instruction (MMP) may be enough to significantly improve many aspects of propulsion technique in long term manual wheelchair users.

The final chapter was designed to test the impact of training on upper extremity pain and shoulder forces and to also make comparisons between overground and dynamometer propulsion biomechanics. Peak resultant forces at the handrim, CA and SF were found to be very similar between dynamometer (ch3) and over ground (ch2) propulsion. In addition, the final chapter showed that shoulder forces did not change significantly from baseline to long term for all groups. Although an increase in peak hand rim Fr was observed in the CG, it was evident that their Fr shoulder forces did not increase. It was also apparent that the short term increases in propulsion forces seen in treatment subjects over ground also occurred on the dynamometer but again shoulder forces did not increase significantly. These findings may have gone unnoticed had dynamometer propulsion and shoulder modeling not been performed.

Overall, study results indicated that training based on CA and SF had many beneficial results with few unintended negative changes. Furthermore, it was clear that although a majority of training occurred during dynamometer propulsion, technique changes translated to over ground propulsion. The only unintended or unexpended changes occurring were those related to increases in Fr at the handrim. The FB and IO groups showed short term increases while the CG showed a long term increase.   Motor learning theory suggests that short term performance deficits are to be expected when learning from random stimuli which occurred in the FB group more so than the IO group.[59] It was interesting however that CA and SF did not worsen in the short term however their percent change was greater in the long term than short term.   Previous training studies have suggested that this time frame (three weeks of training) may be too short to notice force changes which also may have occurred in the present study be.[34,111] A longer period between baseline and follow up could help to confirm. Future studies should consider longer follow up testing time. Furthermore, the long term force increase seen in CG members may suggest that propulsion technique is not stable and that subjects could benefit from periodic training to refresh skills. In fact, a recent study found subjects presenting with poor transferring technique displayed worsening technique over time without training.[156] Again a longer follow up time combined with larger sample size could help to confirm these points. It was also evident that control group members visited the lab three times while the treatment groups required four visits. Although the baseline to long term time frame was equivalent for all subjects (3 months) this difference was a study limitation and may have altered the results.

Another study limitation was evident in that we did not report subjects chair configurations prior to training. Because rear wheel axle position has been shown to influence propulsion biomechanics a secondary analysis was performed to determine if training was effected by axle placement.[121,157-161] Axle position relative to the shoulder, in both horizontal (XPOS) and vertical (YPOS) directions were recorded using kinematic analysis by Optotrak. Both measurements were captured while subjects were sitting at rest with their arms adducted next to the thorax with elbows flexed at 90°. XPOS was a fore and aft position of the axle relative to the shoulder while YPOS represented relative seat height or the vertical position of the axle marker in relation to the shoulder (Figure 24). The XPOS value was larger if the rear axle position was posterior relative to the shoulder (biomechanically disadvantageous) and smaller or a negative value if anterior to the shoulder(biomechanically advantageous). The mean XPOS was 6.2 ± 5.1 cm while YPOS was 71.5± 5.8 cm. Figures 26 &27 show the differences in horizontal and vertical measurements by group. Significant differences were not found between groups (p>.05). Averaged across all time points, groups, surfaces and velocities, Pearson's correlation test revealed YPOS to be negatively correlated to contact angle (CA), r= -0.13, p= .02 (Fig 25) while XPOS was not significantly correlated to CA or stroke frequency (SF) p>.05. The relationship between SF and YPOS was similar to those found in previous studies. [121,158] To determine if axle placement influenced training results, a General Linear Mixed Model (GLMM) was performed where XPOS and YPOS were included in the model as covariates. Subject height, weight and time with injury were controlled for as well. Results were not significantly different from the model used previously in chapters 2 and 3 where XPOS and YPOS had not been controlled for. All group by time interactions were equivalent between the two models for dependant variables SF, CA, and max Fr. Therefore, a significant interaction between training and axle position was not found.

In looking more closely at our subject population, it was evident that mean XPOS was comparable (more rearward axle position) to previous studies and not within an range that could be viewed as advantageous.[121,161] For example, Mulroy et al. showed that a wheel axle placed 8 cm in front of the shoulder yielded a significant decrease in upward force around the shoulder.[161] Numerous researchers have reported a forward axle position to improve many aspects of propulsion including reduced stroke frequency, increased stroke time, increased contact angle, decreased rate of rise of force, lower electromyograph activity, lower oxygen cost and higher mechanical efficiency .[121,158,160,161] In essence wheelchair configuration can provide a user with a condition that fosters improved propulsion biomechanics however training can still effective regardless of set up. The present study found that the magnitude of change in propulsion technique was not influenced by rear axle position despite a majority of the subjects having a relatively forward axle placement. Future studies could incorporate varied axle positions with training to clarify this relationship.

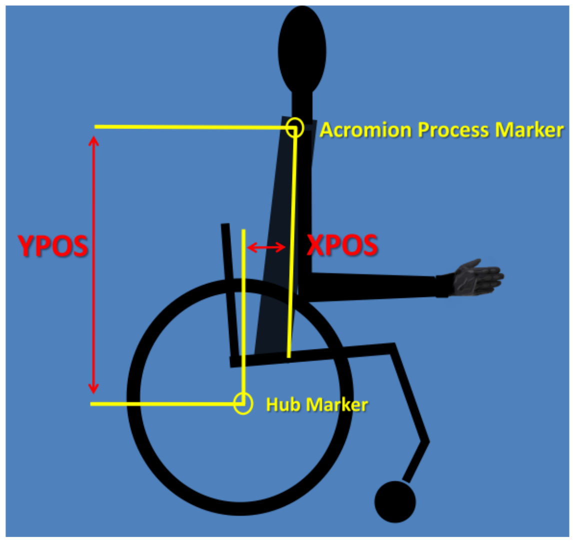


Figure 24. Axle position measurement. The markers needed for axle position measurements and orientations are shown. The XPOS is the fore-and-aft position of the axle with respect to the shoulder; YPOS is the height of the wheelchair with respect o the shoulder.

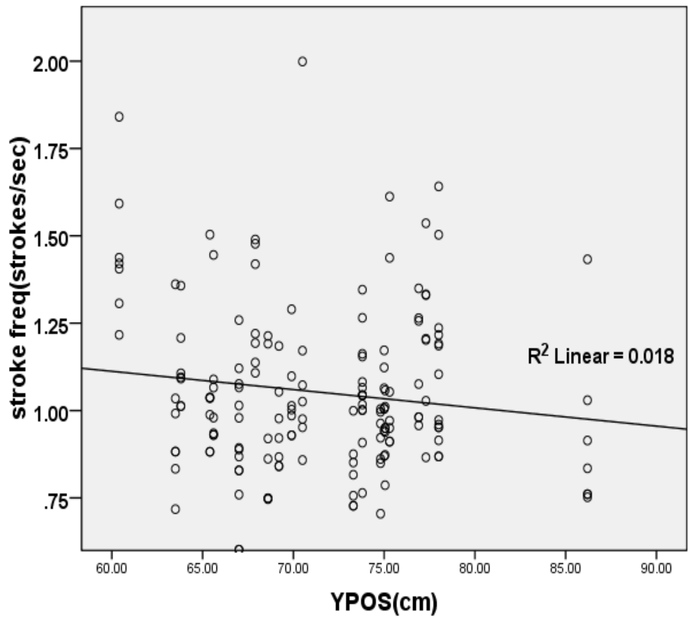


Figure 25. **YPOS vs. stroke frequency. The scatter plot shows the statistically significant relationship (p=.02) between the vertical position of the axle relative to the shoulder and stroke frequency across all groups, surfaces, and speeds. A regression line has also been provided(r= -0.13).**

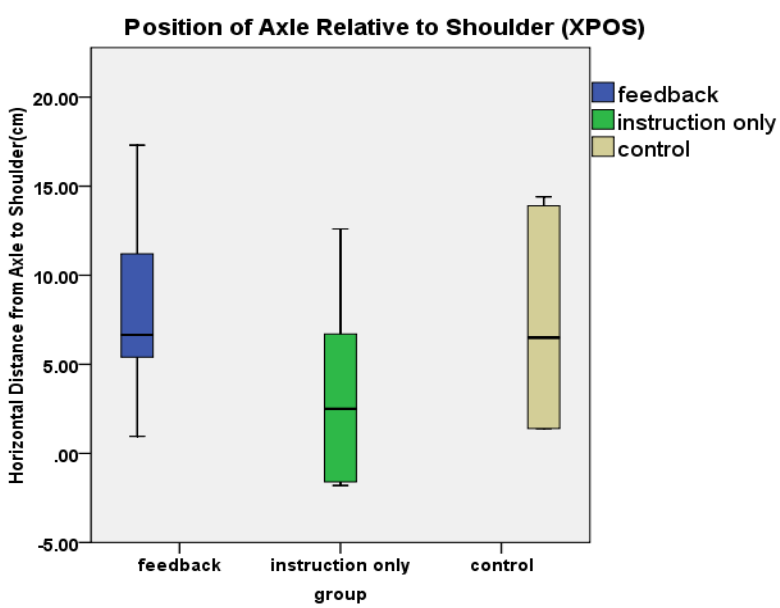


Figure 26. **Horizontal shoulder position relative to axle (XPOS). The box plot shows the XPOS mean value differences between groups FB, IO and CG.**

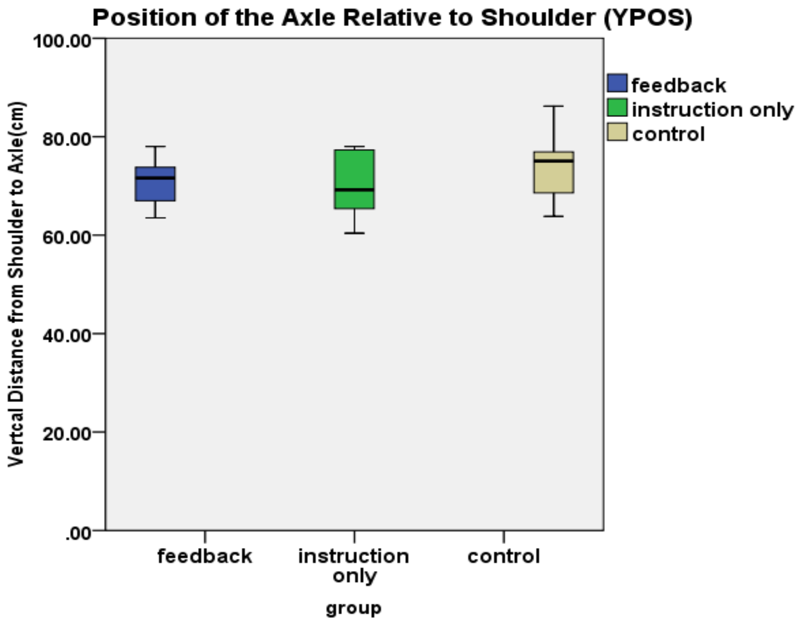


Figure 27. Vertical shoulder position relative to axle (YPOS). The box plot shows the YPOS mean value differences between groups (FB, IO and CG)

While the present studies time from baseline to long term follow up was three months, actual training time was limited to only three visits over the course of three weeks. Three weeks of training although comparable to previous propulsion training studies could be considered an insufficient amount of practice time for skill mastery. For example author Malcolm Gladwell has suggested the "10 year or 10,000 hour rule" is required for skill mastery. [162] The current study had to be designed with recruitment and retention in mind. Our training schedule was designed to be practical and cost effective while maximizing effect and minimizing fatigue and subject drop out. While it was not feasible to meet the learning criteria set by Gladwell and others, motor learning techniques described by Rice et al. 2010 were implemented to accelerate and maximize the learning process.[134] It is also important to note that other researchers in the area of propulsion training have indicated that long periods of practice may not lead to improved propulsion biomechanics because technique is largely dictated by the mechanical constraints of the task and the physical characteristics of the musculoskeletal system.[163] Despite the limited training time frame, intervention subjects displayed significant same day and three month changes in propulsion biomechanics thus supporting the dosing and timing parameters used.

While our findings have shown that simulated propulsion environments like the dynamometer can be valuable, future research could move away from artificial environments all together. A training study could be designed where subjects receive all training including real time feedback during over ground propulsion only. Projectors or screens could be mounted on the back of a chair or moving object in front of the participant who follows close by. In addition, 3D motion analysis systems like Vicon are fully capable of capturing the data necessary to model upper extremity forces during over ground propulsion.

While the present study has emphasized sub maximal training to illicit biomechanical changes related to technique improvements rather than training effects, there are additional advantages inherent to this approach. Keeping the work load low during skill training helps to minimize the risk of injury and may allow more individuals to benefit. For example, working at a level that is close to maximal can increase the likelihood of injury development. This combined with the fact that wheelchair users are already at an increased risk of developing injury supports use of a sub maximal approach. Furthermore, there is undoubtedly some minimum fitness threshold necessary for an individual to achieve a given contact angle and stroke frequency while maintaining velocity. If the target velocity condition was too challenging subjects' propulsion biomechanics may have deteriorated. The intention of this protocol was not to induce biomechanical failure or stress therefore a manageable target speed condition (1.5 m/s) was used in addition to a self selected condition. One of the goals of the target condition was to provide subjects with velocity assistance while learning to modify CA and SF in hopes that they could then apply the techniques to their freely chosen pace. It is likely that subjects would have found it more difficult to generalize propulsion technique learning to self selected velocities had their mechanics been overly challenged and disrupted during the target condition. The same rational was applied to the design of the studies overground course. This course was intended to be realistic but again not overly challenging. For example the ramp was 1.2 degrees, far less than the 4.8 degree minimum standard set forth by the Americans with Disability Act (ADA) guidelines. It was determined that a majority of individuals could learn and safely apply our propulsion techniques to this mild ramp condition. The extent to which individuals should apply the same techniques to more extreme environmental conditions is beyond the scope of this study and warrants further investigation. It was our hope that subjects could learn appropriate techniques sub maximally and then in the future if fitness and conditioning improves the techniques would translate to more varied and demanding surfaces.

In addition to intensity level, great detail was applied to selection of feedback variables. While FEF was excluded as a training variable because it may contribute to unintended changes in propulsion biomechanics, other types of force feedback should be considered as training variables. The present training system found that increased contact angle allowed subjects to apply force over a greater distance thus reducing rate of rise of peak resultant force which has also been linked to injury (Figure 1) $r=-0.45$, $p=.03$. [110] The other type of force related to pain and injury and not addressed directly as a training variable in this study is peak resultant force. Provided velocity remains constant, subjects could be instructed to minimize peak resultant forces streamed in real time. Being aware of peak Fr's during each stroke could help subjects to reduce the loads placed on the upper extremities. Minimizing both rate of rise and maximum resultant force would mean addressing the two most injurious forms of force as supported by ergonomics and wheelchair propulsion literature.[15,16,110] If this strategy was implemented however, it would be necessary for researchers to include simultaneous safety checks like monitoring component forces at the shoulder. For example, an individual could theoretically maintain velocity, reduce peak Fr however still produce elevated component forces known to be injurious. Therefore the inclusion of peak Fr as a real time training variable is feasible however other real time targets (velocity, CA &SF) would need to be presented at the same time to avoid other unintended changes. Finding the ideal stroke is ultimately a give and take if the total amount of energy remains constant in the system from trial to trial. Ultimately a wheelchair user must find the optimum balance in propulsion biomechanics that create a situation where the least amount of injuries movements and forces occur. A final perspective on propulsion proficiency could be to view it in terms of movement variability which should lessen as a function of practice and skill acquisition.[164]

The current protocol requires at least one instrumented wheel, a dynamometer and a computer to support the software which could be perceived as disadvantageous. However, the cost savings resulting from a potential reduction in secondary upper limb impairments could easily off-set initial expenses and should not be overlooked. It is also important to consider the possibility that a successful training program could be carried out based on the principles of the current work using less equipment and technology. For example, with only verbal feedback provided by a therapist and one instrumented wheel, it may be possible to effectively measure and train an individual's stroke technique over time. In fact, clinicians without access to an instrumented wheel could still teach propulsion using verbal or auditory feedback based on the motor learning principles and clinical practice guidelines presented in the current paper. A client could be told to use a low cadence and long smooth strokes, with intermittent verbal instruction, over random ordered practice surfaces like carpet, ramp and tile. Again, consistent with motor learning theory, a client could then benefit from additional feedback which places emphasis on the effects of their movements like number of strokes taken, speed, and stroke length rather than movement pattern. The use of a low-cost video camera could also be used to record propulsion technique which could assist both the instructor and client throughout the learning process.

Cowan et al. 2008 and others have described a training scenario for clinicians without access to a Smart Wheel or dynamometer. [134,165] A resource such as the Smart Wheel Users Group SWUG data base could be used as a reference for clinicians to compare and predict the performance of their clients during over ground wheelchair propulsion. The SWUG data is evidence-driven, clinically meaningful, and provides a practical method to objectively assess manual wheelchair propulsion.[165] Although the data base contains some information that can only be obtained through use of an instrumented wheel many variables can be collected with far less technology. For example a clinician could obtain speed and push frequency information through use of a stop watch and an over ground course of a known distance. A therapist could mark a path of a set distance on a tile, carpet, or ramped surface and record the time taken by their clients to complete the path. In the clinic, users could start from a stationary position and accelerate to a comfortable velocity, pushing through the finish line. The number of times the client pushes in that distance would equal stroke frequency and time taken to do so could be recorded and used to calculate speed (speed=distance/ time). [165] The values obtained would then be compared to data base values for the purposes of assessment and prediction. Although push rim forces cannot be obtained without an instrumented wheel, they could be addressed in a clinical setting indirectly through focus on maximizing contact angle. For example in the present study ror peak FR was correlated to size of contact angle r=-0.45, p=.03 (Figure 1). In essence, using a larger contact angle allowed treatment subjects to reduce rate of rise of peak resultant force. Therefore encouraging subjects to maintain velocity using long smooth strokes in which contact angle is maximized could help clients to minimize ror peakFr.[110]

6.0 APPENDIX

All Tables Correspond to Ch 4

Table 1. (Target Speed) Resultant force

Subject group	time	Fr Handrim (Fr/velocity)	Std. Error	Fr Shoulder (Fr/velocity)	Std. Error
Control Group	baseline	64.7	4.2	70.1	5.4
	short term (same day)	63.9	3.8	57.8	8.8
	long term (3 months)	73.0	4.7	55.1	8.3
Instruction Only	baseline	67.6	3.9	65.5	5.6
	short term (same day)	72.2	3.6	60.0	9.3
	long term (3 months)	67.0	4.4	54.5	8.5
Feed Back	baseline	67.1	3.8	55.6	5.4
	short term (same day)	72.6	3.5	47.6	8.6
	long term (3 months)	68.3	4.5	64.5	7.4

Table 2. (Self Selected Speed) Resultant force

Subject group	time	Fr Handrim (Fr/velocity)	Std. Error	Fr Shoulder (Fr/velocity)	Std. Error
Control Group	baseline	74.2	2.8	67.5	5.4
	short term (same day)	75.7	2.6	55.8	8.8
	long term (3 months)	82.3	3.3	54.1	7.4
Instruction Only	baseline	75.6	2.6	67.9	5.6
	short term (same day)	82.1	2.4	68.6	9.9
	long term (3 months)	76.3	2.9	54.9	8.4
Feed Back	baseline	70.5	2.6	57.8	5.4
	short term (same day)	82.3	2.4	50.4	8.7
	long term (3 months)	75.1	3.1	63.5	7.4

Table 3. (Target Speed) Shoulder Component Forces and Moments

Forces	FB			IO			CG		
	Baseline	Short Term	Long Term	Baseline	Short Term	Long Term	Baseline	Short Term	Long Term
Anterior Force (N/mps)	13.3 SE 3.1	11.1 SE 3.0	7.8 SE 5.9	11.8 SE 3.3	12.8 SE 3.2	24.3 SE 6.7	13.0 SE 2.5	11.7 SE 2.8	18.3 SE 4.3
Posterior Force (N/mps)	37.0 SE 3.7	43.0 SE 3.2	40.2SE 4.2	31.6 SE 3.8	33.3 SE 3.4	35.5 SE 4.9	29.8 SE 3.4	29.8 SE 4.2	25.6 SE 4.4
Superior Force (N/mps)	20.9 SE 6.1	27.9 SE 5.4	13.5 SE 5.1	17.7SE 6.5	22.1 SE 5.8	7.5 SE 5.4	10.2 SE 4.6	10.3 SE 6.1	15.2 SE 5.7
Inferior Force (N/mps)	45.5 SE 5.2	38.7 SE 4.4	53.1 SE 6.2	40.9 SE 5.7	45.2 SE 4.8	46.2 SE 7.0	40.7 SE 4.0	43.0 SE 3.7	39.6 SE 5.6
Medial Force (N/mps)	10.9 SE 3.5	11.6 SE 3.1	9.5 SE 2.1	7.0 SE 4.0	9.5 SE 3.5	13.9 SE 2.4	16.9 SE 2.7	16.7 SE 3.2	14.3 SE 2.4
Lateral Force (N/mps)	19.0 SE 2.8	19.3 SE 2.5	33.2 SE 3.5	18.3 SE 2.8	19.2 SE 2.6	11.8 SE 4.0	17.2 SE 1.8	19.7 SE 1.8	22.7 SE 6.3

Moments	FB			IO			CG		
	Baseline	Short Term	Long Term	Baseline	Short Term	Long Term	Baseline	Short Term	Long Term
Abduction Moment (N/mps)	13.0 SE 2.9	9.9 SE 2.3	11.7 SE 2.0	4.8 SE 3.2	6.2 SE 2.5	4.2 SE 2.2	5.9 SE 1.6	6.0 SE 1.4	4.9 SE 1.3
Adduction Moment (N/mps)	9.0 SE 2.0	5.7 SE 1.1	5.2 SE 1.4	2.8 SE 2.2	5.6 SE 1.2	3.8 SE 1.6	2.3 SE 1.0	3.0 SE 1.0	2.7 SE 1.1
Ext. Rot. Moment (N/mps)	13.9 SE 3.6	8.7 SE 2.6	7.4 SE 2.0	6.6 SE 4.0	9.4 SE 2.9	4.8 SE 2.2	8.5 SE 2.1	8.1 SE 2.0	5.1 SE 1.2
Int. Rot. Moment (N/mps)	25.0 SE 3.4	18.8 SE 1.9	19.7 SE 3.4	13.4 SE 3.8	15.6 SE 1.9	18.4 SE 3.8	9.2 SE 1.8	10.1 SE 1.5	9.8 SE 2.2
Flexion Moment (N/mps)	27.9 SE 5.2	19.8 SE 2.2	30.8 SE 1.7	14.0 SE 6.1	16.2 SE 2.3	15.0 SE 5.5	10.4 SE 2.6	11.4 SE 2.2	12.5 SE 2.1
Extension Moment (N/mps)	17.5 SE 4.4	8.8 SE 1.6	10.7 SE 2.9	5.1 SE 5.1	8.9 SE 2.4	4.7 SE 3.2	7.0 SE 2.2	7.0 SE 1.7	5.2 SE 1.3

Table 4. (Self Select Speed) Shoulder Forces and Moments

Forces	FB			IO			CG		
	Baseline	Short Term	Long Term	Baseline	Short Term	Long Term	Baseline	Short Term	Long Term
Anterior Force	6.1SE 3.1	9.2 SE 3.4	6.4 SE 5.1	8.8 SE 3.5	13.3 SE 3.9	14.4 SE 6.0	8.5 SE 3.4	8.1 SE 3.8	14.6 SE 6.0
*Posterior Force	32.0 SE 4.1	49.0SE 5.2	34.6SE 5.0	33.2 SE 4.7	43.2 SE 6.1	45.6 SE 5.8	34.2 SE 4.6	34.5 SE 5.8	30.2 SE 6.0
Superior Force	17.7 SE 5.5	30.5 SE 7.5	15.4 SE 6.5	13.4SE 6.3	24.3 SE 8.7	14.3 SE 7.6	6.4SE 6.2	5.3 SE 8.4	7.7 SE 5.7
Inferior Force	32.1 SE 4.8	40.7 SE 4.5	40.8 SE 6.5	41.1 SE 5.6	56.2 SE 5.1	49.0 SE 7.6	43.9 SE 5.4	46.4 SE 5.0	47.4 SE 7.7
Medial Force	5.2SE 3.4	11.1SE 4.0	7.0SE 2.7	76.1SE 3.9	10.5SE 4.6	12.2 SE 3.1	16.0 SE 3.8	15.2 SE 4.4	12.3 SE 3.2
Lateral Force	12.2 SE 2.2	17.6 SE 2.2	35.4 SE 7.5	18.7SE 2.6	19.3SE 2.6	7.5 SE 9.0	18.0SE 2.5	19.9 SE 2.5	22.2 SE 8.9

Moments	FB			IO			CG		
	Baseline	Short Term	Long Term	Baseline	Short Term	Long Term	Baseline	Short Term	Long Term
Abduction Moment	8.2SE 2.0	6.7 SE 1.8	8.0SE 1.4	4.9SE 2.3	7.6 SE 2.1	4.3SE 3.8	6.5 SE 2.2	7.1SE 2.0	5.2 SE 1.8
Adduction Moment	6.1 SE 1.3	5.9 SE 1.3	3.1SE 1.3	2.4 SE 1.5	7.1SE 1.5	4.2 SE 1.6	2.4 SE 1.4	3.3 SE 1.4	2.0 SE 1.6
Ext. Rot. Moment	6.6 SE 2.6	5.3SE 2.5	4.7 SE 1.3	6.1SE 3.1	13.0SE 2.9	5.3 SE 1.6	8.5 SE 2.9	8.3 SE 2.7	5.1 SE 1.6
Int. Rot. Moment	16.7 SE 2.3	19.8 SE 1.8	15.7 SE 2.5	12.4 SE 2.6	18.1 SE 2.1	19.2 SE 3.0	9.0SE 2.5	9.2 SE 2.1	8.7 SE 3.0
Flexion Moment	9.6 SE 2.7	7.3 SE 2.2	5.9 SE 1.7	4.5 SE 3.2	9.2 SE 2.5	5.5 SE 1.8	7.6 SE 3.1	7.0SE 2.4	5.8 SE 1.8
Extension Moment	20.0 SE 3.2	21.3SE 2.8	24.2 SE 2.4	13.5SE 3.7	18.1SE 3.2	15.6SE 2.8	10.1 SE 3.6	12.0 SE 3.1	13.4 SE 2.9

Table 5. Fr Handrim

FB compared to CG	
Time Point	**Score/Significance** F(2,166.3)=7.0, p=.001
Short Term*	t(183)=-3.1, p=.002
Long Term	N.S. p=.16
IO compared to CG	
Time Point	**Score/Significance** F(2,163.5)=5.3, p=.006
Short Term*	t(183)=-1.9, p=.05
Long Term*	t(145.2)=2.2, p=.02
FB compared to IO	
Time Point	**Score/Significance** F(2,164.9)=0.9, p=.39
Short Term	**N.S.**
Long Term	**N.S.**

Table 6. Shoulder Maximum Medial Force (Max Fz)

FB compared to CG	
Time Point	**Score/Significance** F(2,18.5)=6.7, p=.006
Short Term*	t(44.0)=3.3, p=.002
Long Term	N.S., t(12.3)=1.5, p=.1
IO compared to CG	
Time Point	**Score/Significance** F(2,17.2)=1.9, p=.17
Short Term*	**N.S.**
Long Term*	**N.S.**
FB compared to IO	
Time Point	**Score/Significance** F(2,17.8)=5.3, p=016
Short Term	**N.S.,** t(44.0)=1.8,p=.07
Long Term*	t(11.7)=2.7, p=.01

Table 8. Shoulder Posterior Force (Min Fx)

FB compared to CG	
Time Point	**Score/Significance** F(2,34.7)=6.7, p=.003
Short Term*	t(44.0)= -3.4, p= .001
Long Term	N.S., t(31.4)=-1.5, p=.12
IO compared to CG	
Time Point	**Score/Significance** F(2,33.4)=2.1, p=.13
Short Term	N.S.
Long Term	N.S.
FB compared to IO	
Time Point	**Score/Significance** F(2,34.5)=1.3, p=.26
Short Term	**N.S.**

Table 9. Shoulder Inferior Force (Min Fy)

FB compared to CG	
Time Point	**Score/Significance** F(2,34.1)=2.2, p=.1
Short Term	N.S.
Long Term	N.S.
IO compared to CG	
Time Point	**Score/Significance** F(2,33.5)=1.9, p=.15
Short Term	N.S.
Long Term	N.S.
FB compared to IO	
Time Point	**Score/Significance** F(2,33.8)=5.5, p=.008
Short Term*	t(44.0)=-3.2, p= .002
Long Term	**N.S.**

Table 10. Contact angle group by time

FB compared to CG	
Time Point	**Score/Significance** F(2,169.5)=51.1, p=.0001
Short Term*	t(183)=-7.2, p=.0001
Long Term*	t(149.3)=-8.3, p.0001
IO compared to CG	
Time Point	**Score/Significance** F(2,164.6)=30.6, p=.0001
Short Term*	t(183)=-5.7, p=.0001
Long Term*	t(147)=-6.3, p=.0001
FB compared to IO	
Time Point	**Score/Significance** F(2,167.0)=2.9, p=.05
Short Term	t(183.0)=-1.3,p=.1
Long Term*	t(148.1)=-2.2, p= .02

Table 11. Stroke frequency group x time

FB compared to CG	
Time Point	**Score/Significance F(2, 177.8)=28.3,p=.0001**
Short Term*	t(183.0)=7.1, p=.0001
Long Term*	t(150.7)4.2, p=.0001
IO compared to CG	
Time Point	**Score/Significance F(2,170.8)=42.0, p=.0001**
Short Term*	t(183.0)=7.8, p= .0001
Long Term*	t(148.2)=6.6, p=.0001
FB compared to IO	
Time Point	**Score/Significance** F(2,174)=3.0, p=.05
Short Term	t(183.0)=-0.9,p=.32
Long Term*	t(149.4)=-2,4,p=.01

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