Manual Wheelchair Propulsion in Older Adults

By

Rachel Ellen Cowan

BA, University of North Carolina Wilmington, 2000

MS, Wake Forest University, 2003

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This dissertation was presented

by

Rachel Ellen Cowan

It was defended on November 19th, 2007

and approved by

Shirley Fitzgerald, PhD, Rehabilitation Science and Technology, University of Pittsburgh

Alicia Koontz, PhD, Rehabilitation Science and Technology, University of Pittsburgh

Stephanie Studenski, MD, MPH, Department of Medicine, Geriatrics, University of Pittsburgh

Dissertation Director: Michael Boninger, MD, Department of Medicine, Physical Medicine and Rehabilitation, University of Pittsburgh

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Compared to individuals with spinal cord injury (SCI), propulsion by older adults is poorly defined. The goal of this project is to examine the impact of wheelchair, surface, and user characteristics on propulsion mechanics in older adults and individuals with SCI. All participants self-propelled over a series of surfaces at a self-selected velocity and kinetic data collection were provided by the SmartWheel. We described a standard clinical protocol (SCP) for objective assessment of manual wheelchair propulsion and defined reference values for individuals with SCI based this protocol (N=128). The SCP requires self-propulsion over tile, low pile carpet, and up an ADA ramp. In addition we provided a decision framework based on graphical reference data; guiding clinicians through an objective assessment of propulsion, identifying opportunities for intervention and follow-up. We then compared propulsion of individuals with paraplegia (IP, N=54) and older adults (OA, N=53). OA propelled slower than IP; used a greater push frequency and minimum Mz, shorter stroke length, and similar resultant force. When surface difficulty increased, the IP group responded with increased work. This may indicate a lack of capacity in OA to respond to increased resistance. For our cohort of older adults we defined the impact of surface type, wheelchair weight, and rear axle position (N=53). As surface

difficulty or chair weight increased, velocity decreased. Controlling for velocity, push frequency, resultant and tangential force increased as surface difficulty increased; heavier chairs had decreased stroke length and increased resultant and tangential force; and posterior axle positions had increased velocity. Controlling for velocity, posterior axle positions had increased forces. Finally, we examined the impact of strength and gender. Body-weight normalized grip strength was collected. Stronger individuals propel faster than weaker individuals. On low pile carpet, both genders decreased velocity versus tile, but women decreased push frequency while men increased. Surface type has a substantial impact on propulsion velocity and force; magnifying any differences between users and wheelchair configurations. Wheelchair weight and axle position independently affect propulsion mechanics. Gender and strength appear to influence propulsion. Older adults are marginal self-propellers at best; powered mobility may be a more appropriate mobility solution.

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PREFACE

I could spend days composing an all-embracing preface; there is an extensive list of people to thank and experiences to note. For those of you who have found this document, I hope you are enjoying life. Life is an amazing, crazy adventure which we should fully embrace and experience. These years have been characterized to immense growth on my part. I am a far better person now than I was when I arrived. I wouldn't trade or redo a single experience.

Huge thanks to my family for taking great care of me for my entire life. Without your support and understanding, I'd never had made it this far.

For Jen Mercer, now Jenny C., three gold stars for putting up with me during our time at HERL. I'm not going to know what to do without your desk next to mine. Who else will ever answer my weekly questions about simple trig?

Dumb luck landed me an amazing advisor. Dr. B has consistently pushed me when I needed it & kicked me out of the lab when I pushed myself too hard. Words cannot express my gratitude for your guidance, patience, and example.

The Wake Forest Crew, Heather W, Jamie T, Steve G, Aaron S, Tina E…… We've been out of wake for almost 5 years now. I miss every single one of you and look forward to our annual reunions. Weddings, babies, degrees… through it all we'll have a great time!

I wish I could give everyone who has helped me a few lines….. but I'm tired of typing and have some friends to help before I leave. Be assured that every member of HERL; students, staff, and faculty has contributed to this process. You've each had an impact and I'll miss you.

"I just got one last thing, I urge all of you, all of you, to enjoy your life, the precious moments you have. To spend each day with some laughter and some thought, to get you're emotions going. To be enthusiastic every day and as Ralph Waldo Emerson said, "Nothing great could be accomplished without enthusiasm," to keep your dreams alive in spite of problems whatever you have. The ability to be able to work hard for your dreams to come true, to become a reality."

- Jim Valvano-

1. INTRODUCTION

Wheelchair propulsion is an alternative form of mobility with the capacity to facilitate community participation and functional independence. Reliance on wheeled mobility ranges from complete; as often is the case for individuals with paraplegia or tetraplegia, to temporary use; such as individuals with pelvic or femoral fractures, to those who use it as a supplement to ambulation; commonly seen with older adults or individuals with cerebral palsy. Characteristics of the wheelchair, user, and environment can in isolation or in interaction affect the function of an individual. Prescription of a manual wheelchair for a specific individual requires an understanding of the interactions between the capacity of the user, the characteristics of the wheelchair, and the expected environments of use. Only by untangling this paradigm can we begin to objectively determine what characteristics of the user, wheelchair and environment, interact to produce or impede independent mobility.

Of the estimated 1.7 million individuals who use wheeled mobility devices, 87% use manual wheelchairs (1). The largest group of manual wheelchair users (MWU) are older adults (65+) (55.6%), yet relatively little is known about propulsion in this group (2). The research plan of the National Institute on Aging identifies “improvements in the availability and effectiveness of assistive devices” for older adults as a developing initiative, highlighting the need for additional research (3). Mobility limitations results in substantial financial, emotional, and physical burden, subsequent to loss of independence (4-8). Therefore evaluation of manual wheelchair propulsion in the older adult would be relevant to developing initiatives of the National Institute on Aging.

Older adults who find propulsion difficult or impossible in "standard" wheelchairs, especially when confronted with surfaces with increased rolling resistance, such as carpet or ramps, may be able to achieve improved mobility across those same surfaces when fitted with the lightest available wheelchairs in a personally optimized configuration. Standard wheelchairs by definition are >36lbs with a fixed rear axle position; often found in hospitals. Reduced weight, <30lbs, and an adjustable rear axle are two characteristics found only on ultralight wheelchairs. Researchers have conclusively demonstrated that appropriate vertical and horizontal rear axle position reduces the amount of force required for propulsion, decreases the metabolic demand of propulsion, and is related to decreased prevalence of upper extremity injury and pain (9-12). Much of this research has occurred in relatively young populations with spinal cord injuries uncomplicated by systemic reductions in physical strength and conditioning. Current Medicare reimbursement policies, which set the standard, often regulate older adult MWUs to heavier "standard" wheelchairs with fixed rear axle positions. Properly fitted, extremely light wheelchairs have the potential to preserve independence and social participation among the older adult with reduced strength by facilitating independent mobility. The intent of this project is twofold; 1) to evaluate propulsion biomechanics of older adults with various levels of strength in a series of manual wheelchairs with different weight and axle configurations and 2) compare these individuals to a group of community dwelling manual wheelchair users drawn from an international database.

1.1 BACKGROUND

Ultimately, the amount of force required to propel a wheelchair is determined by the rolling resistance (RR). RR is affected by three major factors; the combined weight of the user and wheelchair, wheelchair configuration, and surface RR (13-15). Brubaker and McLaurin discussed the factors affecting wheelchair performance in two early publications, identifying mass distribution as affected by axle position and user shoulder position relative to the axle to be the primary determinants of performance (13-15). Wheelchair configuration, specifically rear axle position, affects RR by altering the distribution of the weight of the system across the front and rear wheels. Moving the rear axle anterior shifts a greater portion of the system weight on the larger rear wheels, decreasing the RR of the system. Rear axle position therefore affects not only the amount of force required to propel a wheelchair across any given surface with any set of wheels, it affects the user's ability to apply propulsive forces. The vertical distance between the user's shoulder and the axle affects the geometry of the push, affecting the ability of the user to apply force to the pushrim(16).

Secondary factors identified by Brubaker and McLaurin include characteristics of the propulsion surface, wheel and caster characteristics and combined weight of the user and wheelchair. As the weight of the system increases, so does the overall RR. Wheel diameter is inversely related to RR, thus when equal amounts of weight are placed on the small front wheels and large rear wheels, there is greater RR acting on the front wheels. Furthermore, tire characteristics affect rolling resistance(17;18). Solid tires result in higher RR than many pneumatic tires(17). Tire pressure of pneumatic tires affects rolling resistance, with lower pressures increasing rolling resistance due to greater deformation of the tire during contact with the ground(18). Surface RR is constant on a given surface, but as with tires, surfaces that deform more on contact with the tire, such as thick carpet, result in higher rolling resistance. Additionally, surfaces that are pliable to the point that wheelchair tires sink into them provide greater RR due to the increased area of contact between the tire and surface, such as occurs when propelling through sand or pea gravel. Surface RR is constant, but when coupled with the effects of rear axle position and wheelchair weight, could impose a demand exceeding the strength of the user.

The majority of biomechanical and functional research addressing these factors has been conducted among MWU with spinal cord injuries. Given older adult individuals represent by far the largest proportion of MWU, and in general receive heavy non-adjustable wheelchairs, it is imperative to begin to understand the impact of wheelchair weight, axle position, and surface rolling resistance on this cohort.

1.1.1 Axle position

Rear axle position affects the magnitude of force, stroke length and push frequency used during propulsion. Both the amount of force and application location has physiological and biomechanical implications. Ideally, the rear axle of a manual wheelchair should be positioned horizontally as anterior as possible without negatively affecting the user's stability(10;11;19). Hughes et al. used a dynamometer to determine the effect of seat position on lever drive and handrim wheelchair propulsion kinematics(12). Lower and rearward seat positions in the handrim propulsion resulted in more joint motion in the saggital plane. Seat position did not have an effect on stroke length or time, although the authors cite the small difference in the seat positions as a possible cause for the lack of changes. In the only study to examine MWU in their own wheelchairs when exploring the impact of axle position on propulsion biomechanics, Boninger et al demonstrated that horizontal axle position was correlated with the frequency of propulsion and the rate of rise of the resultant force(10). Both vertical and horizontal axle position was related to push angle. Kotajarvi et al. examined the effect of seat position on over ground propulsion biomechanics, in contrast to studies examining propulsion on ergometers or dynamometers(11). Generally, lower seat positions (decreased vertical distance between the axle and shoulder) resulted in increased push angel, push time axial and radial forces. To maximize physiological and biomechanical efficiency, seat height (vertical position of the rear axle) should result in 100^o to 120^o of elbow flexion when the hand is placed at top center of the pushrim to maximize physiological and biomechanical efficiency (Figure 1.1) (9-11).

Figure 1.1 Older Adult seated in a test wheelchair with elbow angle at 100 - 120^o

In addition to impacting stroke length and push frequency, multiple studies have demonstrated that rear axle positions located posterior to the user's acromion result in higher peak forces and loading rates(10-12). Additionally such positioning is associated with the prevalence of upper extremity pain and injury, although a causative relationship has not been demonstrated (10). Building on the study described above, Boninger et al. examined the relationship between median function and characteristics of the user and propulsion biomechanics(20). Subject weight was related to pushrim biomechanics and median nerve function. Individuals who weighed more used higher forces to propel at a given velocity. Additionally, weight was associated with the presence of impaired median nerve function. Loading rates and forces required for propulsion at any given velocity can be decreased by shifting the axle anterior and by decreasing the weight of the system through reduction of chair weight, thereby decreasing the demand on the user.

Vertical axle position (seat height) indirectly affects peak forces and loading rates. Physiologically, increasing the vertical distance between a MWU's shoulder and the rear axle increases the metabolic demand (9). Van der Woude et al. examined the relationship between cardiorespiratory response, propulsion kinematics, and seat height in a group of nine non-wheelchair users(9). Seat positioning resulting in 100 - 120 degrees of elbow extension resulted in increased mechanical efficiency and push angle. Physiological response to horizontal axle position has not been documented. Appropriate vertical and horizontal positioning could be the difference between independent mobility and loss of independence in frail, older adult individuals. However, all the evidence to date has been collected on young individuals with and without spinal cord injuries, limiting the accuracy of generalization to other populations, including the older adult.

1.1.2 Weight

Generally, research has demonstrated higher weight, either of that the user or combined user and wheelchair, are associated with larger propulsive forces, regardless of axle position. Larger propulsive forces are associated with the prevalence of median nerve damage and wrist pain among MWU, as described earlier in research by Boninger et al(20). The majority of weight in the user-wheelchair system is provided by the user. However, it is imperative that the wheelchair add as little weight as possible to the entire system, especially with older adults, who generally are weaker than their younger peers. Standard wheelchairs weigh a *minimum* of 36 lbs, and often exceed 40lbs. Research has not established the effect of increased chair weight, independent of axle position, on propulsion biomechanics. The impact of increased chair weight may be minimal in populations with age appropriate function and strength, but may be substantial in populations with compromised strength, such as older adults with mobility limitations.

1.1.3 Rolling Resistance (RR)

Wheelchair configuration and combined weight are generally constant across time, thus the subsequent impact on RR is also constant. However RR is also affected by characteristics of the propulsion surface and tire pressure. Surfaces with higher coefficients of friction, greater deformation, or greater slope increase RR, therefore require more force for propulsion at any given speed. An early study by Wolfe et al examined the effect of carpet on energy expenditure and self-selected velocity during wheelchair propulsion in a group of individuals with a varied history of manual wheelchair use (21). Thirty-five individuals participated, ten without a physical disability (novice users), ten individuals considered to be "deconditioned", and ten individuals with paraplegia. Deconditioned individuals were defined as manual wheelchair users having "disabilities of various degrees and types which had necessitated prolonged hospitalization and bed rest, contributing to general debilitation and deconditioning" [sic]. Subjects completed overground propulsion across concrete and carpeted surfaces in an Everest and Jennings Premier Standard wheelchair. Today, this type of wheelchair is considered a "standard" or "depot" wheelchair. Both the novice and experienced wheelchair users chose a significantly lower velocity for propulsion over carpet versus concrete. Reduction in velocity is an energy conservation strategy. However, even at a reduced velocity, energy consumption remained constant or increased, indicating these surfaces imposed a higher energy demand on the individual at any given velocity(21). This energy demand was as much as 56% greater on carpet in "deconditioned" [sic] manual wheelchair users and 36% greater in individuals with paraplegia(21) A more recent study by Newsam et al. examined differences in over ground propulsion biomechanics between individuals with low paraplegia, high paraplegia, C-7 tetraplegia, and C-6 tetraplegia(22). Seventy men with spinal cord injuries propelled a test wheelchair over tile and carpeted surfaces at a self-selected free and fast pace. Participants also propelled on two simulated inclines, 4% and 8% on a wheelchair ergometer. All groups propelled slower on carpet compared to tile, and on both inclines. As injury level increased, velocities decreased across surfaces, with individuals with C-6 tetraplegia selecting a “fast” velocity slower than what normally is required in a community setting. This highlights the fact that individuals with compromised strength, endurance, or function may benefit the most from reductions in chair weight and anterior axle positioning. Increased slope also results in increased metabolic demand, as demonstrated by Van der Woude et al(23).

Analysis of propulsion biomechanics confirms forces and moments associated with propulsion increase as resistance to propulsion increases, such as occurs when individuals transverse carpets or ramps(24;25). In a series of conference abstracts and subsequent publication drawn from over ground propulsion trials collected during the 2003 and 2004 Veterans Wheelchair Games, greater forces were required as the resistance provided by the surface increased(24-26). This increase in force was coupled with a decreased self-selected velocity. Greater forces require more muscle contraction, translating to increased metabolic demand. It is plausible that propulsion over carpet and ramps could impose a demand on an older adult MWU exceeding their ability, thus preventing independent propulsion. However, small alterations in wheelchair weight and axle position may independently or in combination partially mitigate the increased demand of carpet and ramps, facilitating independent propulsion.

1.1.4 WC Classification & Medicare policies

Wheelchair classifications are mainly defined by two of the previously discussed criteria; degree of axle adjustability and wheelchair weight, both of which can impact the ability of an individual to independently propel a manual wheelchair.

Table 1.1 Manual Wheelchair Classification Codes. HCPCS codes are developed by the Centers for Medicare and Medicaid Services and serve as the primary guidelines for all government and private insurance manual wheelchair reimbursement policies

General Name	**Weight (lbs)**
Standard	>36
Lightweight	34-36
High Strength Lightweight	<34
Ultralight weight	<30

Wheelchair classifications are defined by the Healthcare Common Procedure Coding System (HCPCS), and a general classification is given in Table 1.1. Wheelchair prescription and subsequent reimbursement is based on the expected duration of use and functional ability of the individual in a *specific* wheelchair. Ultralight wheelchairs are traditionally only reimbursed if an individual is unable to complete instrumental activities of daily living in lightweight wheelchairs (IADL), sit greater than three hours daily in the wheelchair, or require non-standard frame dimensions.

Physiologically, propulsion in ultralight wheelchairs imposes a smaller cost on users when compared to standard wheelchairs(27). A group of seventy-four individuals with a spinal cord injury, forty-four with paraplegia, thirty with tetraplegia, propelled an ultralight and a standard wheelchair around an outdoor track at a self-selected velocity for twenty minutes. For all subjects, distance traveled and self-selected speed was greater in the ultralight chair(27). Only individuals with paraplegia demonstrated a lower metabolic cost in the ultralight. However, although the individuals with tetraplegia expended the same amount of energy when propelling both wheelchairs, they traveled farther and faster in the ultralight, an indication of greater efficiency, which has functional implications. Although many older adults generally receive standard wheelchairs, the higher weight of this chair coupled with the fixed axle position may result in physiological and biomechanical demands which exceed their ability, ultimately compromising their mobility.

1.1.5 WC provision and use among the older adult

A portion of older adult Americans use assistive devices for mobility purposes(2). These devices include canes, walkers, and wheelchairs(2). Documentation of the provision and use of such devices is scattered, in part because over half of these individuals acquires the device through self-payment without using Medicare or private insurance(28). Manual wheelchairs are the most common Medicare DME expenditure, representing 39% of all provisions(28). Generally, manual wheelchairs are rented by Medicare for ten months, after which the user can purchase or continue to rent the wheelchair. If the user elects to purchase the chair, Medicare pays for an additional three months, after which the chair belongs to the consumer. The consumer, however, must pay 20% of the purchase price. Medicare pays for a rental an additional five months if the consumer chooses to continue the rental. Rentals are conducted on a monthly basis, with fees for rentals determined state by state. Monthly rental rates are equal to 10% of the total allowable cost of the item. The only manual wheelchair that Medicare will purchase outright is an Ultralight. Rental chairs are unlikely to be fitted to the user, resulting in a scenario where the user might not be able to successful self-propel. However, documentation of such fitting or the lack thereof is not available.

Adding to the difficulty of defining WC use among the older adult is the presence of intermittent disability and the use of multiple mobility strategies, which depend in part on the environment and capacity of the user (29-31). Gill et al. (29) defined mobility disability as self-reported inability to walk one quarter of a mile and climb a flight of stairs without personal assistance. Mobility disability was assessed every month for five years in seven hundred fifty-four community dwelling individuals aged 70 and over. Mobility disability among this cohort was "characterized by frequent transitions between states of independence and disability." Transitions occurred in both directions, from disability to independence and vice versa. However, female gender, older age, and the presence of physical frailty were associated with decreased incidence of transitioning to independence and increased incidence of worsening mobility disability. Use of wheelchairs was not tracked; however the authors noted that programs should, in part, focus on the maintenance of independent mobility. Properly fitting ultralight wheelchairs could serve to preserve independent mobility in older adults, such as the frail, who experience periods of mobility disability. Wheelchairs are used by the older adult to supplement lower extremity disability (32-35). However, among the older adult, only a small percentage relies exclusively on a wheelchair for their mobility needs. For the ambulatory older adult, wheelchair use within the home may not be necessary or possible. In a study of 153 community dwelling individuals who received a new wheelchair, no individuals used their wheelchair in all locations, while only 4% walked in all locations, indicating a mixed use approach to mobility (30). Wheelchair use was the predominant method used in locations far from home, while walking was the predominant method used inside the home. Exploration of wheelchair use within the home led the authors to conclude a mixture of impairments and architectural barriers dictated the choice between ambulation and wheelchair use. The authors concluded selective use of a wheelchair was the normal pattern. Review of Phase 2 data from the 1994-1995 National Health Interview Survey Disability Supplement indicated 13% of individuals over the age of 65 who reported difficulty with one activity of daily living used a wheelchair(36). The majority of these individuals used the wheelchair to go outside, which CMS does not recognize as an acceptable reason for purchase or rental. Furthermore, 97.4% of these individuals relied in part on Medicare as their health insurance. The majority, 65.3% relied on Medicare as their primary insurance with supplemental secondary insurance. Given CMS interpretation of Medicare policy that restricts purchase of DME to what is needed within the home, those who ambulate in the home would not be eligible for MWCs, restricting or preventing their community participation, and isolating them in the home. Due to the selective and intermittent use of manual wheelchairs by the older adult, owing in part to changes in disability, it is simplistic to assume that these users do not need the benefit of a fitted wheelchair. Indeed, this very misunderstanding could be why current policies do not provide ultralights to this population, which may be needlessly impairing their independence.

1.1.5 Manual Wheelchair Propulsion Research in Older Adults

Investigations focused on the biomechanics and physiology of manual wheelchair use among the older adult is sparse at best. However, research by Sawka et al. has indicated manual wheelchair propulsion requires a higher percentage of an older adult individual's physical capacity as compared to middle aged and young individuals (37). All participants were MWU, reporting similar years of wheelchair use. Participants completed a progressive intensity discontinuous exercise stress test on a wheelchair ergometer. Heart rate was monitored continuously and oxygen uptake was sampled every minute. Maximal heart rate, peak VO2, and maximal power output decreased with age, which is not unexpected. However, the authors noted the maximal power output obtained by the older adult group, 7 W, would require older adult MWU to work at their maximal level when crossing a tiled surface and exceed it to transverse a carpeted surface. Among middle aged MWU, the tiled surface would only require 44% of their maximal ability. These results indicate that any increase in the demand of propulsion, as would occur when traveling across carpet, could exceed an older adult individual's capacity, restricting their mobility. Aissaoui et. al. demonstrated improved biomechanical efficiency in a group of older adult wheelchair users by increasing the rearward tilt of the seating system and increasing the recline angle of the backrest (38). Fourteen experienced manual wheelchair users propelled a manual wheelchair fixed to a roller system (rear wheels only). Each user completed a ten meter steady-state propulsion trial in nine different backrest and seat angle combinations. Biomechanical efficiency (tangential force/resultant force) increased with increasing seat and backrest recline angle. Increasing the seat angle and backrest recline angle in their chosen method effectively resulted in an anterior shift in the rear axle position relative to the user's shoulder, which is associated with improved force production.

1.1.6 Strength and Propulsion

Overall strength declines with age and is often further reduced in older adults who are experiencing mobility disability. Although a direct link has not been established between the strength of an older adult and their ability to self-propel, such evidence exists for individuals with SCI. In a longitudinal multi-center Dutch study evaluating changes in fitness and function in newly injured individuals with spinal cord injury, individuals with higher summed manual muscle test scores demonstrated better performance on a wheelchair propulsion test(39). It appears the impact of strength on self-propulsion is greatest for the weakest individuals. Noreau and colleagues demonstrated a strong relationship between strength and functional independence in individuals with tetraplegia, but only a weak relationship in individuals with paraplegia(40). Together, these studies suggest strength is a key component in the ability to self-propel. Strength may be a very important factor determining the self-propulsion success of an older adult. Despite the link between strength and propulsion performance, it has yet to be established if strength affects propulsion mechanics or if strength affects an individual's response to a change in wheelchair configuration or surface type. Assuming all individuals propel in a similar manner despite their strength is a short sighted approach. Identifying strength related differences in propulsion mechanics may allow the refinement of user specific interventions to improve mobility.

1.2 PURPOSE

Older adults (65+) represent the largest group of manual wheelchair users in the United States. However, their propulsion mechanics are among the least well defined. This lack of information may represent a barrier to providing the most optimally configured manual wheelchair. Currently, they often receive heavy, poorly configured manual wheelchairs and report difficulty or inability to self-propel. In addition, this group of users often remains in part ambulatory, identifying them as a unique subset of users, distinct from full-time users. Multiple factors interact to impact propulsion, including the wheelchair configuration, surface of propulsion, and characteristics of the user. Thus, the immediate goal of this project is to document the impact of wheelchair, surface, and user characteristics on propulsion mechanics of older adults. We address this purpose through four manuscripts. Delineating the role of each factor will support the development of manual wheelchair prescriptions specific to the needs and ability of the older adult. We first defined the propulsion mechanics of a group of community dwelling users with a spinal cord injury and outlined a method by which change in propulsion could be objectively assessed in a clinical setting. Defining this group serves to create a profile of a "successful" self-propeller for further comparison purposes. Second, we compared a subset of this group, those with paraplegia, to our cohort of ambulatory older adults, thereby assessing the difference between experienced users and novice. Exclusion of those with tetraplegia or of an undocumented injury level provides a more homogeneous comparison point. The third manuscript addresses the impact of wheelchair weight, axle position, and surface type on the propulsion in the older adult. Evaluation of these factors in combination allows for a more realistic transfer of the result to the clinic. Finally, we explored the role of strength on propulsion mechanics in the last manuscript. Comparison of individuals at the upper and lower ends of the strength continuum within our cohort provides preliminary insight into the role of strength on propulsion mechanics.

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2. PRELIMINARY OUTCOMES OF THE SMARTWHEEL USERS' GROUP DATABASE; A PROPOSED FRAMEWORK FOR CLINICIANS TO OBJECTIVELY EVALUATE MANUAL WHEELCHAIR PROPULSION

Rachel E. Cowan, MS; Michael L. Boninger, MD; Bonita J. Sawatzky, PhD;
Brian D. Mazoyer, PTA; Rory A. Cooper, PhD

From the Human Engineering Research Laboratories (Cowan, Boninger, Cooper); Department of Physical Medicine and Rehabilitation (Boninger, Cooper) and School of Medicine (Boninger) University of Pittsburgh; VA Pittsburgh Health Care System Center of Excellence in Wheelchairs and Related Technology (Cowan, Boninger, Cooper); Department of Rehabilitation Science and Technology (Cowan, Boninger, Cooper), School of Health and Rehabilitation Sciences, University of Pittsburgh; Department of Orthopaedics (Sawatzky), Faculty of Medicine, University of British Columbia, Vancouver, British Columbia, Canada; Banner Good Samaritan Rehabilitation Institute (Mazoyer), Phoenix, AZ

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Address all correspondence to:

Michael L. Boninger, M.D.

Human Engineering Research Laboratories

VA Pittsburgh Health Care System

5180 Highland Drive 151R–1

Pittsburgh, PA 15206

Phone 412-365-4850

Fax 412-365-4858

Email: boninger@pitt.edu

Send reprint requests to:

Michael L. Boninger, M.D.

Human Engineering Research Laboratories

VA Pittsburgh Health Care System

5180 Highland Drive 151R–1

Pittsburgh, PA 15206

Email: boninger@pitt.edu

2.1 ABSTRACT

Objective: To 1) describe a standard clinical protocol for objective assessment of manual wheelchair propulsion; 2) establish preliminary values for temporal and kinetic parameters derived from the protocol; 3) develop graphical references and a proposed application process for use by clinicians

Design: Case series.

Setting: Six research institutions that collect kinetic wheelchair propulsion data and contribute to an international data pool.

Participants: A total of 128 individuals with spinal cord injury.

Intervention: Subjects propelled a wheelchair from a stationary position to a self-selected velocity across a hard tile surface, a low pile carpet, and up an ADA compliant ramp. Unilateral kinetic data were obtained using a force and moment sensing pushrim.

Main Outcome Measures: Differences in Self-Selected Velocity, Peak Resultant Force, and Push Frequency across all surfaces, relationship between 1) weight normalized peak resultant force and self-selected velocity; and 2) push frequency and self-selected velocity

Results: Graphical references were generated for potential clinical use based on the relationship between body-weight normalized peak resultant force, push frequency and velocity. Self-selected velocity decreased (Ramp < Carpet < Tile), peak resultant forces increased (Ramp > Carpet > Tile), and push frequency and stroke length remained unchanged when compared across surfaces. Weight normalized peak resultant force was a significant predictor of velocity on tile and ramp. Push frequency was a significant predictor of velocity on tile, carpet, and ramp.

Conclusion: Preliminary data generated from a clinically practical manual wheelchair propulsion evaluation protocol is presented. A proposed method for clinicians to objectively evaluate manual wheelchair propulsion is described.

Key Words: wheelchair, biomechanics, rehabilitation engineering, rehabilitation, insurance

2.2 INTRODUCTION

Wheelchair propulsion is an alternative form of mobility which can facilitate community participation and functional independence for people with mobility impairments [1]. Reliance on wheeled mobility ranges from complete; as often is the case for individuals with paraplegia or tetraplegia due to spinal cord injury (SCI), to temporary use; such as ambulatory individuals with pelvic or femoral fractures, to people who use it as an ambulation supplement; such as frail elderly or individuals with cerebral palsy. Characteristics of the wheelchair, user, activity, and environment interact to impact successful function. Appropriate wheelchair prescription requires an understanding of the interactions between the capacity of the user, characteristics of the wheelchair, and expected environments of use [2,3]. Objective wheelchair propulsion assessment in commonly encountered environments can supplement clinician opinion.

In the United States, current policies of the Center for Medicare and Medicaid Services (CMS) require clinicians to demonstrate why a wheelchair pre-defined by policy is insufficient to facilitate minimal independent mobility needed to perform mobility related activities of daily living [4-6]. Furthermore, CMS is only concerned with the minimum necessary to facilitate mobility within the home[4 6]. Justifications based on community function, a critical component of independence, can be rejected as not medically necessary by Medicare and third party payers[4,7,8]. Subjective clinical assessments, while valuable and accurate, may be discarded as insufficient evidence for a prescribed wheelchair [8,9]. Increasingly, clinicians are reluctantly tailoring wheelchair prescriptions based on what CMS will approve, rather than to the true rehabilitation needs of each individual [7-10]. The gap between CMS policy and clinical guidelines, which are based on evidence-based practice, needs to be eliminated. Objective assessment of manual wheelchair users propelling across surfaces found in a home environment holds potential to help ameliorate the discrepancy between best practice and third party payer policy.

Historically, a technology gap exists between research and clinical based assessments of manual wheelchair propulsion. Research has advanced our knowledge of manual wheelchair propulsion using tools and techniques either unavailable or not practical for use in the clinic. Such tools include motion capture systems, wheelchair ergometers, dynamometers, treadmills, custom force and moment sensing wheels, and electromyography collection devices [11-19]. Additionally, these tools generate data requiring time intensive processing to produce results. Consequently, clinicians have been unable to use research protocols or tools to evaluate and compare their clients against research findings.

The SmartWheel (Three Rivers Holdings, LLC), a recently commercialized tool, may help close the propulsion assessment technology gap between clinicians and researchers. The SmartWheel Users' Group (SWUG) was formed to guide the clinical development and application of the SmartWheel (SW). The SWUG is an international group of researchers, clinicians, industry, advocacy groups, and end users with the primary goal of ongoing development of evidence driven, clinically meaningful, useful, and practical methods to objectively assess manual wheelchair propulsion (Table 2.1). A secondary goal is facilitation of mutually beneficial communication among the key stakeholders.

Table 2.1 Participants in the SmartWheel Users Group. (Fall 2006). Participants of the SWUG represent 4 countries, 12 states, 3 Veterans Administration Hospitals, 1 VA center of excellence, 5 current or previous Model SCI Centers, 3 members of industry, and 1 advocacy group. All listed facilities have participated in an annual meeting or quarterly conference call within the last two years.

6 Degrees of Freedom, LLC (IL)	Rehabilitation Institute of Chicago (IL)
BES Rehab Ltd (England)	Schwab Rehabilitation Hospital (IL)
Cardinal Hill Rehab Hospital (KY)	Shriners Hospital, Philadelphia (PA)
Denver Veterans Affairs Medical Center (CO)	The Center for Assistive Technology (PA)
Enabling Mobility Center, Paraquad (MO)	The Ohio State University (OH)
Glenrose Rehabilitation Hospital (Canada)	The Ohio State University Medical Center (OH)
Good Samaritan Regional Medical Center (AZ)	Three Rivers Holdings, LLC (AZ)
Human Engineering Research Lab (PA)	TiSport LLC (WA)
Hunter Holmes McGuire VA Medical Center (VA)	University College London (Great Britain)
Jackson Memorial Hospital (FL)	University of British Columbia (Canada)
Kessler Institute of Rehabilitation (NJ)	University of Illinois at Chicago (IL)
Kessler Medical Rehabilitation Research and Education Center (NJ)	University of Pittsburgh (PA)
Mayo Clinic (MN)	University of Washington (WA)
Miami Project to Cure Paralysis (FL)	VA Puget Sound Health Care System (WA)
Minkel Consulting (NY)	Vista Medical, Ltd (Netherlands)
Paralyzed Veterans of America (DC)	Washington University in St. Louis (MO)
Rancho Los Amigos National Rehabilitation Center (CA)	Washington University School of Medicine (MO)

Accomplishment of the primary goal of the SWUG is guided by three ongoing tasks; Development of: 1) clinical manual wheelchair propulsion assessment protocols and applications, 2) clinically relevant manual wheelchair propulsion parameters; and 3) reference values based on the clinical parameters.

Therefore, our specific aims are: 1) Description of a standard clinical protocol for objective assessment of manual wheelchair propulsion 2) Establishment of preliminary values for a subset of parameters produced by the SW clinical software and protocol, and 3) Development of clinical graphical references and a proposed clinical application processes.

2.3 METHODS

2.3.1 Standard Clinical Protocol

The SWUG designed the standard clinical protocol (SCP) to match requirements identified by member clinicians as critical to clinical acceptance and implementation. Four requirements were identified; 1) Use of surfaces common to clinics, 2) Use of multiple surfaces representing varied resistance, 3) Provision of useful information from a single module, and 4) Adaptability to available space and time.

The SCP is a modular assessment which required users to propel a manual wheelchair across 1) level tile, 2) low pile carpet, 3) up an Americans with Disabilities (ADA) compliant ramp (a maximum 1:12 rise to run, 8.3% grade, or 5 degree slope) and 4) in a figure eight on level tile with a SW attached unilaterally to the wheelchair [20]. Use of a SW matching the opposing wheel diameter will maintain the User's wheelchair configuration. A SW weighs 10 lbs, increasing the weight of the system, but providing measures of stroke length and force, which cannot be measured in any other manner in the clinic. In all modules, data collection was initiated before users began to move. For tile and carpet, users began from a stationary position on the selected surface, accelerated to a comfortable self-selected velocity, pushing for a maximum of ten seconds, ten meters, or the end of the surface, whichever occurred first. Data collection was terminated before users left the surface or decelerated. On the ramp, users propelled from level ground directly in front of the ramp, with casters touching the ramp threshold, up the full length until reaching a platform. Data collection was terminated before the user ascended onto the platform. Ramp length and slope varied as allowed under the ADA. The fourth module, the figure 8, assessed the ability of the individual to maneuver and is not included in this analysis.

By design, the SCP does not require clinicians randomize or prioritize the order of the modules. Within a clinical environment randomization may not be possible or reasonable. Additionally, definitions of surfaces were loosely constrained to maintain the practically of implementation. Low pile carpet was defined as closed loop industrial type carpet often found in hospitals, clinics, and some businesses. Tile was any smooth, firm panels lining the floors of hospitals and clinics; often linoleum. Ramps qualified if tiled with a maximum grade of 8.3%, per ADA definition. Clinicians are encouraged to assess clients over any surface they feel would provide relevant information; however submissions to the central data pool (described below) were restricted to collections matching any module of the SCP.

For the purposes of the SW clinical software and SCP, steady-state consists of all strokes occurring after the third stroke, which if target velocity has been achieved, represents a state of propulsion inherently different from the acceleration phase described by “start-up” parameters. Restrictions in space and increasing difficulty of modules (ie, a ramp), may prevent achievement of a “steady-state” condition as it is traditionally defined. A minimum of 5 strokes is required for the SW clinical software steady-state calculations, although all available strokes beginning with stroke 4 are included in stead-state calculations. It is incumbent upon the clinician to compare “start-up” and “steady-state” for each client and module to determine if a “steady-state” condition has been achieved.

2.3.2 Key Parameter Selection

When a module was completed, the SW clinical software automatically generated 21 parameters describing the client’s propulsion [21]. Four parameters of the 21 available were identified by the SWUG as representing the most clinically important and relevant information provided by the SW[a] (velocity, average peak resultant force, push frequency, and stroke length). Clinicians within the SWUG felt all assessments should begin with velocity and all users should be able to achieve a minimum threshold velocity for safe and successful community participation. A velocity of 1.06 m/s, representing the average minimum needed to safely cross an intersection [22], was chosen as the threshold for the purpose of discussion in this manuscript. Force, push frequency, and stroke length were selected by the SWUG based on recommendations from the Clinical Practice Guidelines for the Preservation of Upper Limb Function Following Spinal Cord Injury (CPG)[23]. The CPG recommends the minimization of force and frequency of repetitive upper limb tasks and use of long strokes during propulsion[23].

This analysis is restricted to the four parameters identified by the SWUG plus time and distance for each module. Forces are weight normalized for a subset of statistics. Clinicians can generate weight normalized forces by dividing the output of the SW clinical software by their client's weight. To facilitate clinical application, all parameters presented in this analysis, except for distance covered in the module and time to complete the module, were calculated using MatLab[b] in the same manner as parameters calculated by the clinical software. Distance and time were truncated when necessary to include only 5 strokes, which is the minimum needed by the SW clinical software to generate a full report describing start-up, steady-state, and summary results. We limited our analysis to 5 strokes to mimic what a clinician who could only collect 5 strokes could potentially expect to see as a result.

A series of graphs were planned to assist in clinical understanding and application of this analysis. A generalized representation of these graphs is presented in Figure 2.1 Each graph contains three critical elements, a threshold velocity reference line, a line representing the linear regression between the parameters of interest, and the 75% and 95% covariance ellipses; which define four areas of interest. The intent of the linear regression was to visually represent the significant correlation between velocity and push frequency/force documented during preliminary analysis. Average trial velocity was chosen as the dependent variable for regression based on preliminary analysis identifying it as the strongest correlate of force and push frequency. If a regression was not significant ($p<0.05$), a graph was unnecessary and therefore not constructed. Proposed application and interpretation of these areas is described in the discussion.

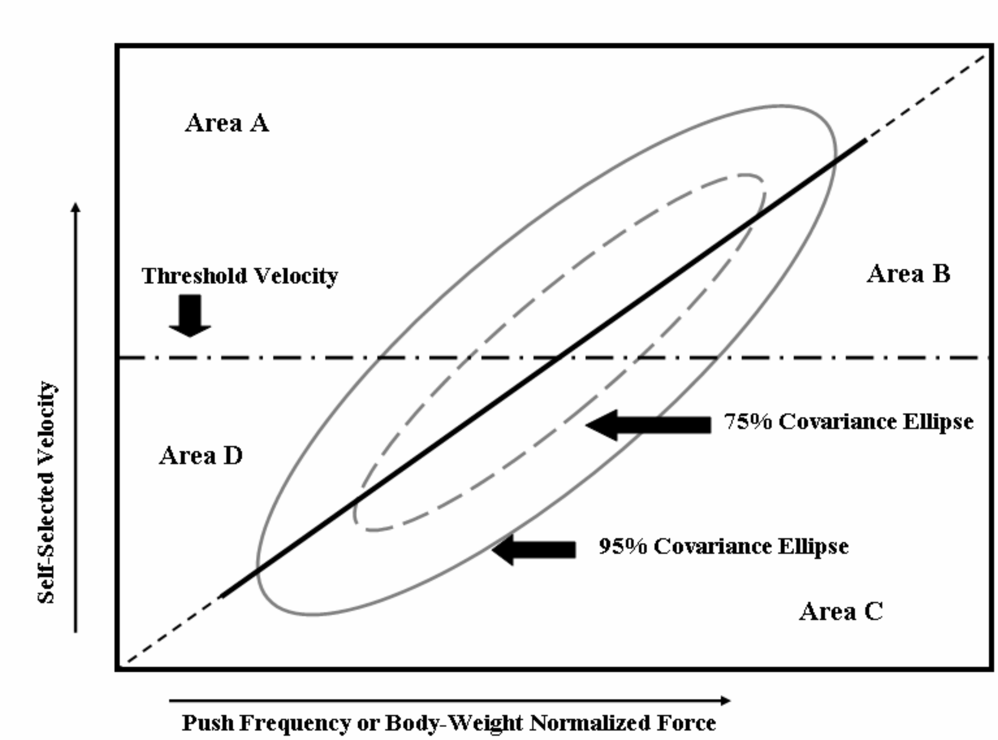

Figure 2.1 Generalized Regression Plot
Solid line is the linear regression line based on existing data. Dashed line is the extrapolated linear regression line. Threshold velocity is represented by a dash-dot line. Solid line ellipse is the 95% covariance ellipse. Dashed line ellipse is the 75% covariance ellipse.
Area A = Above threshold velocity; below average force or push frequency
Area B = Above threshold velocity; above average force or push frequency
Area C = Below threshold velocity; above average force or push frequency
Area D = Below threshold velocity; below average force or push frequency

2.3.3 Central Data Pool

All data used in this analysis was contributed deidentified to a central data pool (CDP) under Institutional Review Board (IRB) approval. The Human Engineering Research Laboratories has an approved IRB to house the CDP. Each individual in the CDP was assigned a unique identifier. All individuals submitted to the CDP met two inclusion criteria: 1) a minimum of one raw SW collection file representing one module and 2) Local IRB approval for contribution to the CDP Exclusion criteria were the failure to meet the inclusion criteria. Restated, an individual was eligible for the CDP if a single SCP module was completed. When available, each submission included user demographics and wheelchair characteristics in addition to the raw SW file generated by the module. User demographics included age (if age < 89), height, weight, gender, primary diagnosis, and years using a wheelchair as the main means of mobility. Wheelchair characteristics included wheelchair manufacturer, wheelchair model, and wheelchair weight. Multiple data collections for each module for an individual could be submitted. For the purpose of this analysis we insured each individual was represented only once by demographics and one raw collection for each module of the standard clinical protocol.

2.3.4 Kinetic Data Reduction and Analysis

This report is focused on the subset of individuals with Spinal Cord Injury or dysfunction (Paraplegia, Tetraplegia, Spina Bifida). For each individual with a SCI, up to the first five strokes of a single data collection session of tile, carpet or ramp submitted to the central pool was selected for analysis. The figure 8 was not included because it is a skill assessment of maneuverability and not accurately described by the parameters. Five strokes represented the minimum needed to complete an assessment and generate a full report by the SW clinical software. Each trial was broken down into start-up and steady state, mimicking the clinical software. The clinical software defines start-up as the first two strokes from a stationary position. Strokes one and two were analyzed and presented separately. The clinical software bases steady-state analyses on the average of all strokes beginning with stroke 4; requiring a minimum of 5 strokes. Statistical examination determined resultant force and velocity were different for start-up and steady-state. Matlab[b] was used to trim data to five strokes, identify the beginning and end of each stroke, define start-up and steady state, and to generate the parameters.

Key parameters were as defined as follows. The resultant force (F), the vector sum of the force applied to the pushrim, was calculated by mathematically combining F_x, F_y and F_z [N] [24,25]. Stroke length was defined as the distance traveled by the hand on the pushrim from the point of contact to the point of release [degrees]. Push Frequency is calculated for the entire trial and is defined as the frequency of pushrim contact[contacts per second]. Steady-State velocity is the average velocity during strokes four and five [m/s]. In the clinical software, all strokes starting with stroke 4 are used in the steady-state average; here this is limited to strokes 4 and 5. Start-up velocity is defined as the peak velocity occurring during the beginning of contact of stroke 2 until the beginning of contact for stroke 3[ms]. Distance covered during the assessment [m] and time to complete the assessment [s] was calculated from the beginning of the first contact to the release of the last contact (up to five strokes).

2.3.5 Statistical Analysis

All analyses were completed using SPSS[c]. The distributions of the data were inspected. Descriptive statistics were calculated for demographics and each parameter for each module. Given the substantial contributions of the Human Engineering Research Laboratories (HERL), we compared data from HERL to the other participants to determine how the overall data set may be affected. It was decided a priori to include all data to capture the population variability, but we wanted to acknowledge potential skew by a heavily represented facility. An ANOVA was used to separately compare subject demographics and steady-state average velocity between HERL and the remainder. A separate ANCOVA for each steady-state peak resultant force, stroke length, and push frequency, was used to compare HERL and the remainder on all three modules, controlling for participant weight and velocity.

To examine the differences in all key variables between surfaces a MANOVA was used. In order to use all available data, we did not use repeated measures, which would have reduced our N to only those subjects who completed all three portions of the protocol. Use of a MANOVA decreases our power to detect differences and is a conservative approach. Linear Regression was used to investigate the relationship between 1) weight normalized steady-state peak resultant force and steady state velocity and 2) push frequency and steady state velocity for each surface for use in the clinically oriented graphs. Covariance ellipses at 75% and 95% were calculated to represent the variability in the parameters. Clinical Reference Graphs based on the linear regression and covariance ellipses were constructed using MatLab[b].

2.4 RESULTS

2.4.1 Demographics

A total of 128 unique individuals were available for inclusion in the analysis. Maximum numbers available for separate surface analyses are as follows: Tile = 123, Carpet = 94 and Ramp = 115. Six facilities contributed individuals: Human Engineering Research Laboratories, Pittsburgh, PA (N=57), University College London, London (N=22), Banner Good Samaritan Rehabilitation Institute, Phoenix, AZ (N=20) University of British Columbia, Vancouver (N=17), Washington University, St. Louis, MO (N=6), and the Mayo Clinic, Rochester, MN (N=6). Demographics are presented for the cohort in Table 2.2.

Table 2.2 Participant demographics

	N	Mean or Count	Standard Deviation
Age (years)	128	40.4	11.2
Height (cm)	122	176.5	10.7
User Weight (kg)	85	80.8	19.9
Wheelchair Weight(kg)	87	13.2	10.2
Duration of Wheelchair Use (years)	128	13.2	10.2
Gender	128		
Male		102	
Female		26	
Diagnosis	128		
Paraplegia		88	
Tetraplegia		22	
SCI – level unknown		11	
Spina Bifida		7	
Wheelchair Make	128		
Quickie		69	
Invacare		26	
TiLite		12	
Colours		6	
Other		15	

2.4.2 HERL versus the remainder

Participants submitted by HERL were significantly older, taller, and heavier than the remainder of the database. When weight was entered into the ANCOVA as a covariate, the differences in kinetic parameters was stroke 1 peak force on carpet, stroke 2 and steady state weight normalized forces on ramp, steady state velocity on tile and carpet, and start-up velocity on ramp. A substantial portion of the participants contributed by HERL were collected at the 2004 and 2005 National Veterans Wheelchair Games. As a group these individuals tended to be older and heavier, but high functioning. All facilities were used in the analysis to allow the reported values to capture the largest amount of variability found within the SCI population.

2.4.3 Description of Key Parameters

Overall means, 95% confidence intervals for the mean, and standard deviations for each parameter for each module are presented in Table 2.3. Generally, stroke 2 peak resultant force was highest and steady-state average peak force lowest. The start-up phase during the ramp condition represents the ascent onto the ramp from the flat area directly in front of the ramp. Steady-State average peak resultant force increased as module difficulty increased. Self-selected steady-state average velocity decreased as module difficulty increased. Start-up peak velocity was significantly different between tile and carpet and ramp, but not between carpet and ramp. Stroke length was similar on all surfaces. Push frequency also was similar across all surfaces, regardless of differences in self-selected velocity or resultant force. Summarized, users selected a lower velocity as surface difficulty increased; achieved through increased forces at the same push frequency and stroke length. The increase in force applied apparently did not offset the increased resistance offered by the surface, resulting in a decrease in velocity.

Table 2.3 Descriptive output for the key parameters. * = significant differences between tile and ramp, † = significant differences between tile and carpet. ‡ = significant differences between carpet and ramp; p ≤ 0.05

	Tile Mean (SD) 95% CI #	**Carpet** Mean (SD) 95% CI #	**Ramp** Mean (SD) 95% CI #
Stroke 1 Peak Resultant Force (N) * †	99.5 (31.1) 93.9-105.2 116	103.9 (33.2) 97.1 – 110.7 92	117.7 (36.8) 110.9-124.5 112
Stroke 2 Peak Resultant Force (N) * † ‡	91.2 (27.8) 86.3 – 96.1 123	106.6 (33.1) 99.8 – 113.3 93	128.3 (35.3) 121.9 – 134.8 115
Steady State Average Peak Resultant Force (N) * † ‡	72.3 (25.3) 67.6 – 77.1 110	87.5 (28.5) 81.2 – 93.8 79	126.2 (34.0) 119.8 – 132.7 106
Start-up Peak Velocity (ms) * ‡	1.2 (0.3) 1.2 – 1.3 123	1.1 (0.3) 1.0 – 1.1 93	1.1 (0.3) 1.0 – 1.1 115
Steady-State Average Velocity (ms) * † ‡	1.2 (0.3) 1.1 – 1.2 110	1.0 (0.3) 0.9 – 1.0 79	0.7 (0.3) 0.7 – 0.8 106
Steady-State Average Stroke Length (degrees)	100.6 (18.0) 97.2 – 104.0 110	97.2 (19.6) 92.9 – 101.5 79	94.1 (20.6) 90.2 – 98.0 106
Entire Trial Average Push Frequency (Contacts per second)	1.0 (0.2) 0.98 – 1.07 123	1.0 (0.2) 0.97 – 1.04 94	1.0 (0.2) 0.98 – 1.06 115
Trial Time for 5 strokes (s)	5.1 (1.1) 4.9 – 5.3 110	5.2 (1.0) 5.0 – 5.4 79	5.2 (1.3) 4.9 – 5.4 106
Trial Distance for 5 strokes (m) * † ‡	5.4 (1.4) 5.2 – 5.7 110	4.5 (1.2) 4.2 – 4.8 79	3.7 (0.9) 3.5 – 3.8 106

2.4.4 Linear Regression and Clinical Graphical Reference

Data points below $Quartile_1$ – 1.5 x the interquartile rage (IQR) or above $Quartile_3$ + 1.5 x IQR were excluded as outliers from the linear regression and covariance ellipse calculations. For each regression and covariance ellipses, this resulted in exclusion of a maximum of 5% of each available dataset. Weight normalized average peak resultant force was a significant predictor of self-selected steady state velocity on tile and ramp conditions. Push Frequency was a significant predictor of average self-selected velocity for the entire trial on all three surfaces. Table 2.4 contains Beta and R squared for each regression model.

Table 2.4 Regression coefficients by model to predict average speed. Only calculated for trials with 5 strokes. For each significant relationship in Table 2.4, a corresponding regression line was plotted (Figures 2.2-2.6). Additionally 75% and 95% covariance ellipses were plotted; allowing clinicians to determine where their client falls in the variability of this population. A reference line was placed at 1.06m/s on the y axis for each regression. This reference line represents the average minimum walking velocity required to safely cross an intersection [22,26].

Predictor	Tile β, (R^2) p=, N	Carpet β, (R^2) p=, N	Ramp β, (R^2) p=, N
Steady-State weight normalized average Peak resultant force	0.043, (0.213) p = 0.000, N = 70	-0.012, (0.008) p = 0.521, N = 54	0.030, (0.151) p = 0.001 N = 70
Entire Trial push frequency	0.384, (0.110) p = 0.000, N = 119	0.506, (0.140) p = 0.000, N=93	0.857, (0.395) p = 0.000, N = 111

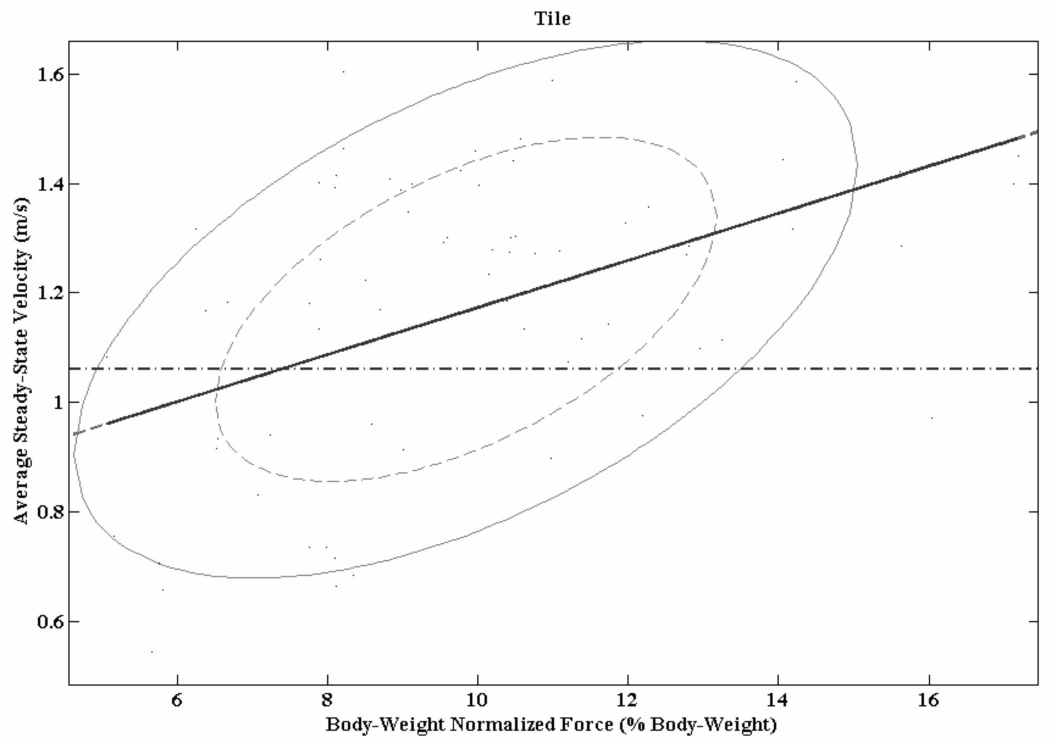

Figure 2.2 Tile Body-Weight Normalized Average Steady-State Peak Resultant Force versus Average Steady-State Velocity

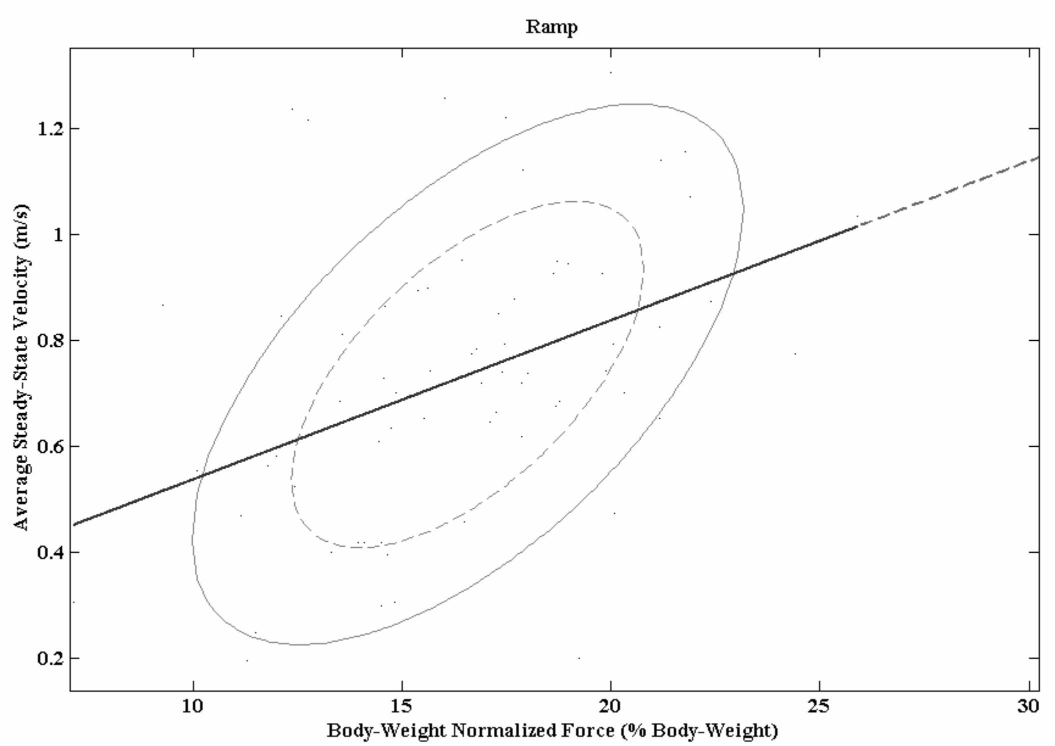

Figure 2.3 Ramp Body-Weight Normalized Average Steady-State Peak Resultant Force versus Average Steady-State Velocity

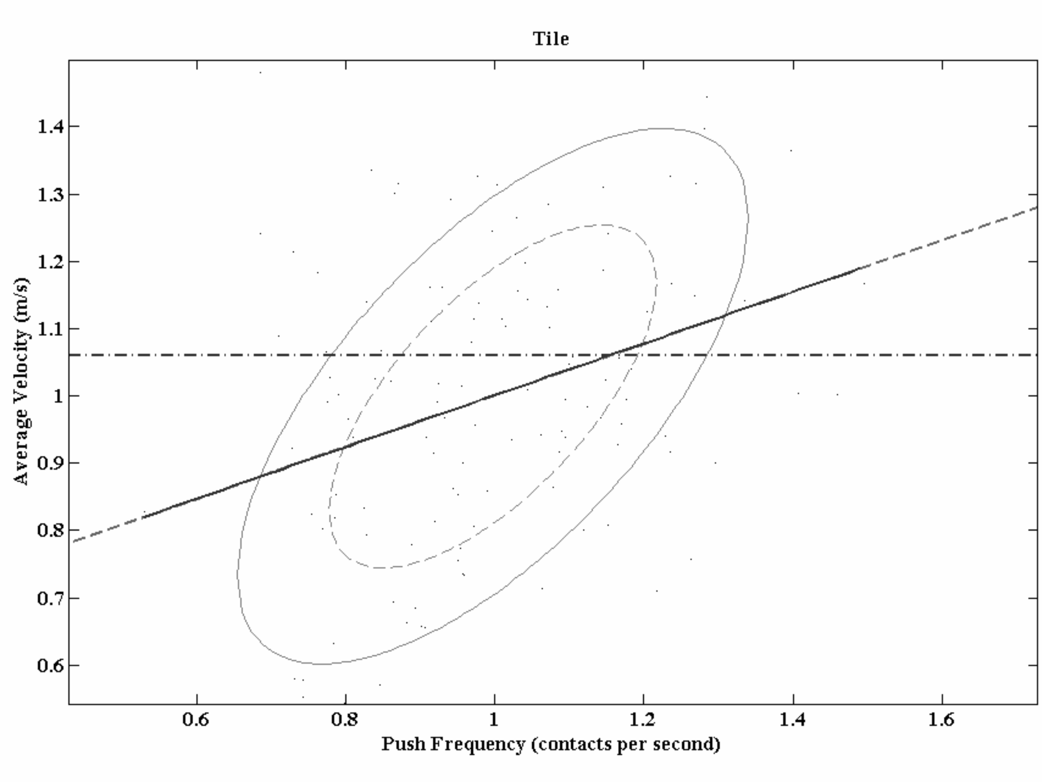

Figure 2.4 Tile Push Frequency versus Average Velocity

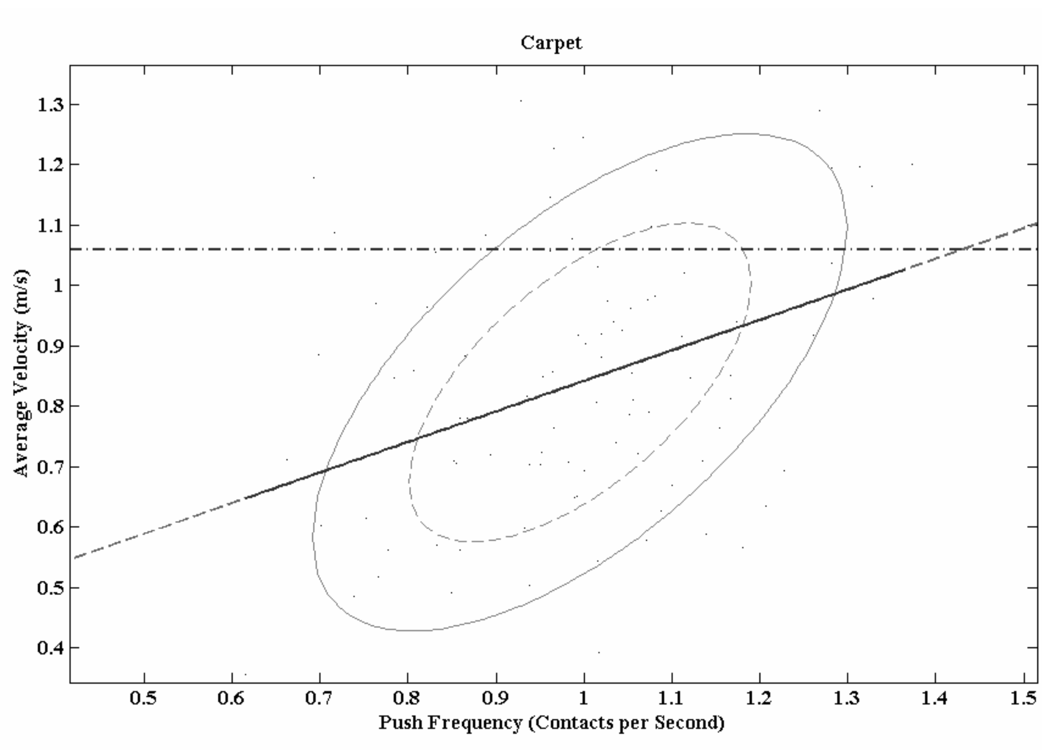

Figure 2.5 Carpet Push Frequency versus Average Velocity

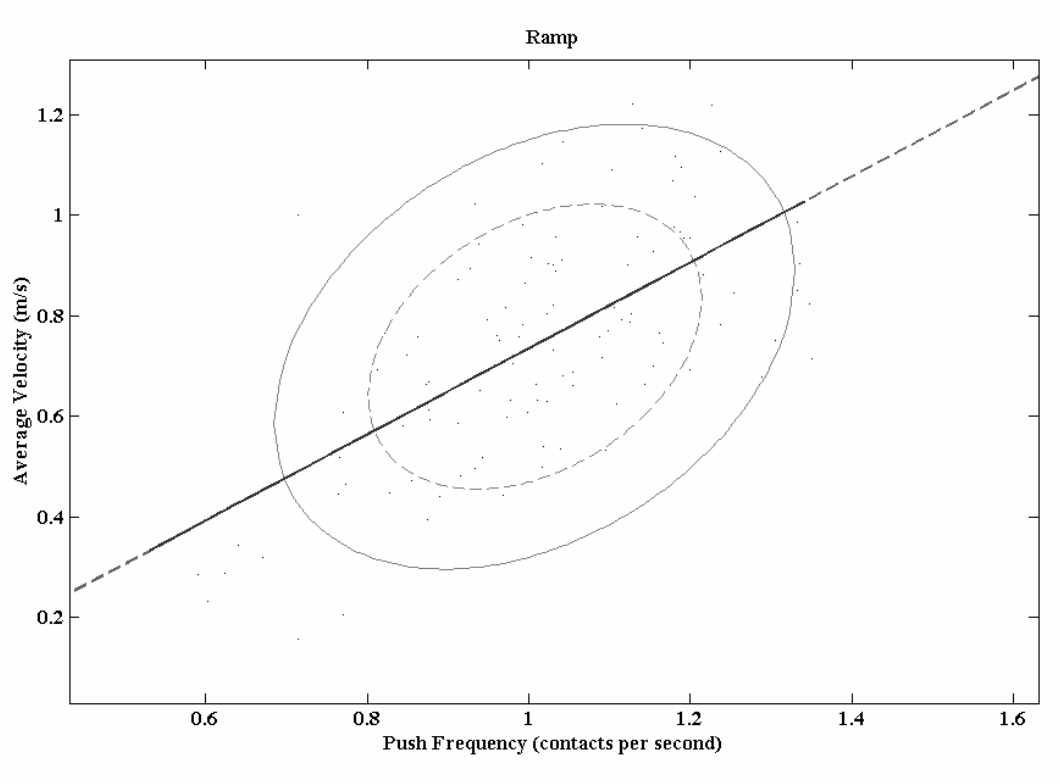

Figure 2.6 Ramp Push Frequency versus Average Velocity

2.5 DISCUSSION

This is the largest multi-site collaboration to evaluate overground propulsion kinetics; represented by over 120 individuals with SCI. It is the first to attempt to use techniques directly transferable to the clinic. Our findings provide preliminary kinetic and temporal values describing overground propulsion for application in clinical evaluations of manual wheelchair propulsion. Guidance to obtain select parameters without the SW is described below (velocity and push frequency). The SW is growing in use and acceptance by clinicians; by the fall of 2007 it is anticipated that 50% of the centers that have them will use it in clinical applications (personal communication).

Primary outcomes of interest were graphical results of the linear regressions for each surface (Figures 2.2-2.6). For each significant relationship, the linear regression line, covariance ellipses, and scatter plot of the predictor parameter (Body-weight normalized average peak resultant force or push frequency) and outcome parameter (velocity) were plotted. Generally, our results fall within ranges reported by previous studies. Body-weight normalized steady state average peak resultant force has been reported as 7.8% to 10.6% [19,27]. Push frequency, often referred to as cadence, has been reported as varying from 0.8 cycles per second to 1.2 cycles per second for a variety of velocities[19,28-32]. Reported self-selected velocity ranges from 0.8 m/s to 1.6 m/s for propulsion on a level surface [19,28-33]. Similarities between our results and those found in the literature are reassuring.

Few studies have examined biomechanics of wheelchair propulsion at a self-selected velocity over surfaces commonly encountered in the community. In general, our results are similar to those of Kotajarvi et al. and Koontz et al[30,34]. Participants in Kotajarvi et al propelled a single wheelchair over a level tile surface in 9 different rear axle positions (N=13) [30]. Average self-selected velocity for all axle positions was 1.48 m/s (+/- 0.16), at a push frequency of 1.23 cycles/s (+/-0.22), using a stroke length of 77.03 degrees (+/-10.21). In comparison, our participants selected a slower velocity, at a lower push frequency with a longer stroke length. A longer stroke length with a constant force will require a lower push frequency to maintain a given velocity. In addition, it is expected that slower velocities are found in conjunction with lower push frequencies. Differences in participants and velocity could account for the differences. Both groups were similar in age, height, weight, and years post injury. However, participants in Kotajarvi et al. were all individuals with low level paraplegia while our participants included individuals with tetraplegia, which could explain the difference in self-selected velocity. Koontz et al examined the propulsion of eleven manual wheelchair users over a series of surfaces at a self-selected velocity[34]. Peak Resultant force during strokes one, two, and steady state over a smooth, level, concrete surface were 103.2 N (+/- 24.4), 101.8 N (+/- 30.7), and 63.6 N (+/-2.9) respectively. In comparison, the results for our tile surface were lower for strokes one and two, but higher for steady-state. Koontz et al steady-state values are the average of stokes 5-7. If strokes 4 and 5 are averaged together, the average steady-state force would be 68.65 N, falling within our 95% confidence interval for tile steady-state average peak resultant force, which is calculated from the average of strokes four and five. Differences in strokes one and two could be a function of self-selected speed and rate of acceleration.

2.5.1 Strengths and Limitations

Clinical application and interpretation of these results requires advance understanding of limitations and strengths of this study. Inherent variability in participants, protocol administration, surface selection, and equipment modification across facilities increases the data variability. Such variability obscures relationships between parameters and differences within a parameter when comparing modules. We believe this limitation has been minimized by the large numbers of participants and future impact will decrease as the database matures. Furthermore we believe this variability is a key strength; allowing this database to encompass the natural variability in users and environments. Approximately half of the current database was contributed by HERL, which may have skewed the results. Any skew is expected to decrease as the database expands through submissions from additional members, limiting the impact of any single facility.

Limiting our steady-state analysis to five strokes increases variability, possibly obscuring relationships/differences. Increasing the number of strokes used in the analysis would decrease the variability. Moreover, individuals may not reach "steady-state" by strokes 4 and 5. In a confined space clinicians should consider comparing multiple collections over a single surface to facilitate determination of what is "typical" for a client. In an attempt to mimic the bare minimum a clinician might have upon which to base their decisions we limited our analysis to 5 strokes; the minimum required by the SmartWheel to generate a clinical report.

Concerning the comparison of start-up and steady state between various modules, clinicians should be aware that "start-up" parameters describing the ramp portion of the protocol represent the transition from level ground to a ramped surface. Individuals interested in propulsion purely on a ramped surface should restrict their inspection to "steady-state" parameters. The transition from level onto a ramp captures by "start-up" parameters may represent a unique challenge for some users; representing a point of evaluation in select instances.

Our analyses were restricted to individuals with SCI, who represent a unique group among the manual wheelchair user (MWU) community. Those who evaluate non-SCI MWU should be aware that their clients may differ. Use of velocity to evaluate the potential of a MWU to achieve successful community function is not diagnosis specific. Any MWU should be able to achieve a minimal velocity for functional purposes, regardless of diagnosis. This is consistent with the CMS National Coverage Determination (NCD) which basis coverage of power and manual mobility on function, independent of diagnosis [35].

2.5.2 Proposed Clinical Application Framework

We present a proposed framework to guide clinicians to intervention opportunities through evaluation of velocity in context with push frequency and force (Figure 2.7). Each clinical reference graph (Figures 2.2-2.6) is divided into four areas by a threshold velocity line and the regression line for the force/push frequency and velocity regression (Figure 2.1). Covariance ellipses allow clinicians to visualize variability in this population and determine how their client compares. Reference values in absence of a velocity context provide a general comparison point (Table 2.3). Clinicians can generate body-weight normalized force used in Figures 2.2-2.6 by dividing output of the SW by their client's body-weight.

Based on clinical guidance the proposed framework prioritizes velocity over force/push frequency. Research indicates forces and push frequency are related to upper extremity injury and minimizing these parameters is recommended to delay upper extremity deterioration [16,23,36]. However, our clinicians report that clinically, low forces were a trademark characteristic of low self-selected speed. In this situation, the clinicians' priority was to increase a user's ability to self-select higher speeds. After a "threshold" speed was achieved, clinicians sought to minimize force and push frequency. The idealized goal for any user is an above threshold velocity coupled with low force and push frequency on multiple surfaces (Diamond box in Figure2.7).

2.5.3 Application Process

Clinical progression through velocity, force, and push frequency evaluations and intervention opportunities is presented in Figure 2.7. Self-selected velocity is a traditional indicator of present and future function in the ambulatory population [37-41]. We selected the average walking velocity required to safely cross an intersection (1.06 m/s) as our threshold. Clinicians may modify the threshold as needed. If a MWU propels below threshold velocity on one or multiple surfaces (Areas C and D on Figure 2.1), the initial goal of the clinician is to design an intervention to achieve threshold velocity (Areas A and B on Figure 2.1). Such interventions could include combinations of strength training, propulsion training, and alterations in their current chair set-up or use of a lighter weight, more adjustable chair. Initial interventions may also include power mobility options if clinical experience indicates the above interventions are inappropriate for their client. Ultimately, the choice between manual or power mobility resides with the user.

Once threshold velocity is reached, clinicians then attempt to preserve velocity while minimizing force and push frequency to help delay the onset of upper extremity pain and dysfunction (Area A on Figure 2.1), following recommendations of the Consortium for Spinal Cord Medicine [23]. Velocity achieved at the expense of high force or push frequency may unnecessarily increase the risk of upper extremity pain and dysfunction (Areas B and C). The ideal goal is a user propelling above threshold velocity at below average force or push frequency across all surfaces (Area A). Users pushing with above average force or push frequency at below threshold velocity may require powered mobility options (Area C).

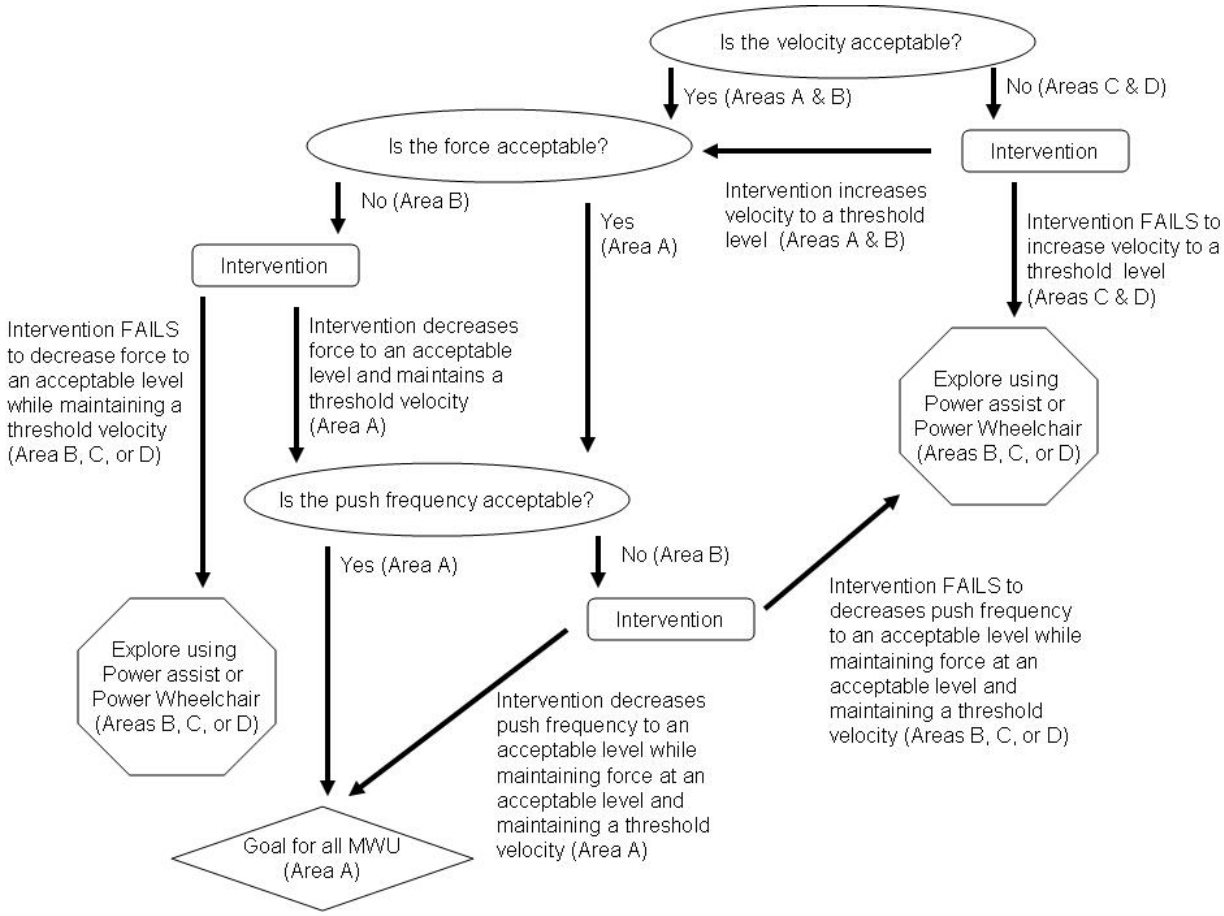

Figure 2.7 Clinician Decision Making Flow Chart. Areas A through D are defined in Figure 2.1

2.5.4 Assessment without a SmartWheel

Clinicians without a SW may still use the push frequency graphical references. Clinicians can mark a ten meter path on a tile, carpet, or ramped surface and record the time to complete. Users start from a stationary position; accelerate to a comfortable velocity, pushing through the finish line. Clinicians count the number of times their client pushes during the distance. Clinicians will need to calculate velocity and push frequency as follows:

Velocity (m/s) = 10m / time to complete 10m (s)

Push Frequency (contacts per second) = # of pushes in 10m / time to complete 10m (s)

As an estimate, these numbers can be used to compare users to the CDP. In addition, clinicians can utilize the proposed framework. Velocity and push frequency assessments and intervention paths are useful without knowledge of force. Clinicians can advise clients to use long, smoothly applied strokes at a low push frequency to minimize force at any velocity [23].

2.5.5 Suggestions for Determining Important Clinical Changes

Clinicians may wonder how much of a change in velocity is important. Unfortunately, concrete numbers do not yet exist. However, examination of the literature indicates small changes in velocity could have a functional impact. Small differences exist in self-selected velocity between SCI levels on a given surface or condition[29,31,42]. Self-selected velocity differed between tetraplegia (0.8 ms) and paraplegia (1.2 ms) in Beekman et al. by 0.4 ms [42]; a difference of 50%.

The absolute and relative difference between a preferred walking velocity (1.22 m/s) and the minimum needed to safely cross an intersection (1.06 m/s) is even smaller; 0.16 m/s [22,26]; a 15% difference. In light of these small absolute but functionally important differences, clinicians may argue that a small, consistent increase in self-selected velocity is important.

Similarly, research has yet to identify absolute or relative force/push frequency thresholds linked to the development or prevention of upper extremity pain. However, the amount of force needed to propel a wheelchair is small, highly repetitive, and related to upper extremity injury [19,36,43,44]. Small reductions in force and/or push frequency would cumulatively decrease exposure; perhaps reducing development of upper extremity pain and injury; providing the basis for recommendations of for the Clinical Practice Guidelines to Preserve Upper Limb Function in SCI [23]. Until thresholds are identified; systematic reductions in force or push frequency are considered beneficial. Therefore it is reasonable for clinicians to argue consistent, but small force or push frequency reductions at a given velocity post intervention represent objective success of an intervention. Interventions reducing both while maintaining velocity possess the strongest evidence.

2.5.6 Future Directions

As the CDP grows, so will development opportunities. Gender, age, and diagnosis specific reference values can be defined. Reference values to evaluate manual wheelchair propulsion without a SW should be developed, thereby assisting all clinicians. A biomechanical focused exploration of the data is warranted in the future. Values for all parameters should be periodically recomputed; ensuring representation of the largest possible population. Eventually these values could become normative, providing a criterion standard for clinicians.

2.6 CONCLUSIONS

A protocol to evaluate manual wheelchair propulsion in the clinic has been described. Preliminary data generated from this evaluation protocol is presented. A proposed framework and application process for clinicians to objectively evaluate manual wheelchair propulsion is described. This method provides a general technique which clinicians may be able to use to compare a client's propulsion to a larger population and/or to compare a client's propulsion before and after an intervention to assess the impact of the intervention.

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Suppliers

[a]SmartWheel: Three Rivers Holdings LLC, 1826 West Broadway Rd. Suite 43
Mesa, AZ 85202
[b]Matlab: The MathWorks, Inc., 24 Prime Park Way, Natick, Massachusetts 01760-1500
[c]SPSS: SPSS Inc., 233 South Wacker Drive, 11th Floor, Chicago, Illinois 60606-6307

3. MANUAL WHEELCHAIR PROPULSION IN NOVICE OLDER ADULTS AND EXPERIENCED INDIVIDUALS WITH PARAPLEGIA

Authors: TBD

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3.1 ABSTRACT

Objective: To identify differences in propulsion biomechanics over two surfaces between experienced individuals with paraplegia and novice ambulatory older adults.

Design: Case Series

Setting: Biomechanics laboratory; international database

Participants: 53 older adults (OA), 54 individuals with paraplegia (IP)

Intervention: Not Applicable

Main Outcome Measures: For all propulsion cycles after the initial three from a stationary position, on tile and carpet: minimum, maximum, and average linear velocity, push frequency, stroke length, distance, total work, peak positive power, peak positive tangential force and Mz, peak resultant force, and peak minimum Mz.

Results: OA and IP slow down on carpet compared to tile, but IP propel faster than OA. On carpet versus tile both groups use similar resultant force; IP increase work, peak power, Mz, and maintain peak minimum Mz; OA increase peak minimum Mz, and maintain work. Comparing the groups, IP use greater work, peak power, tangential force and Mz than OA; OA use greater peak minimum Mz than IP.

Conclusion: Older adults self-selected a lower velocity than individuals with paraplegia; achieved with a higher push frequency, shorter stroke length and similar resultant force. OA demonstrated a greater negative Mz, an indication perhaps their lower velocities are due in part to unintentionally applied braking moments. When surface difficulty increased, only the IP group responded with an increase in work. This may indicate a lack of capacity in OA to respond to increased resistance. Given these findings, optimally configured manual wheelchairs or powered mobility may be appropriate for older adults.

Key Words: Rehabilitation, Wheelchair, Older Adult, Paraplegia

3.2 INTRODUCTION

Older adults who currently, or in the future may utilize a manual wheelchair represent a distinct, yet poorly understood group of manual wheelchair users. Older adults (65+) are the largest group of manual wheelchair users in the United States [1]. This group uses their chair selectively based on their function and environment of use [2]. In addition, many are still ambulatory and experience transitions in mobility disability, implying an intermittent need for a wheelchair [3-5]. Selective use combined with intermittent need for wheeled mobility may prohibit an older adult from developing efficient propulsion mechanics. In addition, this intermittent need and use may contribute to the documented provision of lower quality and less customized manual wheelchairs [6,7]. Combined, these factors may hinder successful independent self-propulsion in older adults. Indeed, inability and difficulty with self-propulsion has been documented [8,9].

Individuals who have a spinal cord injury (SCI) are commonly studied in the field of propulsion biomechanics. Much of this work has the ultimate goal of improving efficiency and decreasing the occurrence of secondary repetitive strain injuries[10]. Research focused on this group has served to provide objective documentation of the impact of optimized axle position [11-13], propulsion training [14,15], and alterations in wheelchair design and configuration[16,17]. Some individuals with SCI could represent the prototype of a "successful" self-propeller, whose wheelchair configuration and personal function combine to facilitate independence in all activities of daily living and successful community participation.

It stands to reason that self-selected propulsion velocity would differ between individuals with SCI and older adults; possibly due to differences in age, function, and experience. If individuals with SCI, specifically those with paraplegia, represent "successful" propellers, then comparison of older adults to individuals with paraplegia may serve to identify avenues for intervention to improve an older adult's ability to self-propel. Comparison across multiple surfaces may highlight additional differences and opportunities, as propulsion velocity and mechanics are influenced by surface type[18,19].

Self-selected walking velocity is commonly used as an indicator of whole body function in older adults and is predictive of future function, disability, and mortality[4,20-22]. Furthermore, walking velocity in older adults is used to differentiate between robust and frail individuals [22]. In the SCI community, self-selected velocity appears to differentiate between quadriplegia and tetraplegia and changes in surface and slope [18,19,23]. Thus, self-selected propulsion velocity may be appropriate measure of the potential for an individual to be "successful" as a manual wheelchair user and to differentiate between groups of users. In addition to velocity, biomechanical differences in walking and propulsion have been noted between age groups, diagnosis, weight ranges, and for propulsion, between experienced and novel users [24-27]. Therefore biomechanical evaluation of propulsion may identify how two groups differ, providing insight which then can be used to design interventions to improve mobility [28].

Therefore, the purpose of this paper is to compare the self-selected propulsion velocity and associated biomechanics of older adults to individuals with paraplegia on tile and low pile carpet, thereby identifying opportunities to improve wheeled mobility for older adults.

3.3 METHODS

3.3.1 Individuals with Paraplegia

Kinetic overground propulsion data for individuals with paraplegia was requested from the SmartWheel User's Group Database (SWUG DB). All data was contributed deidentified under Institutional Review Board (IRB) approval. An individual was eligible for addition to the database if a single portion of a predefined overground propulsion assessment was completed (cite). All individuals with paraplegia who had a body-weight reported were selected for analysis (N=54). The contents and collection methods for this dataset have been described in Chapter 2. Briefly, each individual in the database propelled a manual wheelchair over a series of natural surfaces from a stationary position to a comfortable self-selected velocity with a SmartWheel[a] attached unilaterally. Demographics collected included age, gender, height, weight, duration of injury, and wheelchair manufacturer. For the purpose of this analysis we insured each individual was represented only once by demographics and one trial on each surface.

3.3.2 Ambulatory Older Adults

Fifty-three older community dwelling adults were recruited through flyers at local senior citizen centers, interest groups, bring a friend strategies, and IRB approved research registries (Men = 20, Women = 33). All subjects gave written informed consent prior to participation in the study. The research protocol was approved by the Institutional Review Boards (IRB) of the VA Pittsburgh Health Care System and the University of Pittsburgh. To be eligible for participation, subjects had to self-report 1) the ability to walk without human assistance, 2) the ability to stand-up from a chair, 3) the ability to push themselves in a wheelchair, 4) weight as less than 251 lbs, and 5) score greater than 22 on the mini-mental state exam. Exclusion criteria included self-reported history of stroke or a diagnosis of Parkinson's or Alzheimer's. Participant characteristics can be found in Table 3.1. Participants reported minimal experience with wheelchair propulsion.

3.3.3 Overground Propulsion Data Collection

Subjects drawn from the SWUG DB and older adults propelled a manual wheelchair over hallway tile and low pile carpet. Older adults propelled a manual wheelchair with SmartWheels[a] attached bilaterally. Both groups began in a stationary position on the surface, accelerated to a comfortable self-selected velocity, and pushed through the end of the trial. Each trial distance varied across sites and surfaces. On both surfaces, data collection was initiated before the user began to move and terminated before the user exited the surface.

3.3.4 Older Adults Wheelchair Selection, Fitting, and Practice

A full description of the selection and fitting of the wheelchair is provided in Chapter 4. Three TiLite Model X[b] chairs were secured for the study. Briefly, each older adult was fitted to the smallest possible wheelchair and seat height was adjusted such that each subject demonstrated an elbow flexion angle between 100 and 120 degrees when seated in the chair with their hand placed at top center of the pushrim[29]. Footrests were adjusted to provide support for the feet and thighs, as allowed by the design of the chair. Front and rear seat heights were kept as equal as possible to maintain 0 degrees of seat inclination. Front caster angle was adjusted to minimize caster trail. The horizontal axle position was set to the most rearward available on the manufacturer's bracket system, approximately 1 inch anterior of the backrest. All four configurations were equipped with anti-tippers. Prior to self-propulsion on each surface, each participant completed a 6 minute propulsion task serving as a functional assessment and an opportunity to practice wheelchair propulsion. Non-standardized instruction describing how to propel a chair was given to all participants.

3.3.5 Biomechanical Parameters

Intentional propulsive contact could not be separated from incidental contact occurring during recovery in absence of kinematic data, thus we elected to report our data as a "cycle" average rather than a "stoke" average. A cycle is the period encompassing a propulsive contact and the subsequent recovery. A stroke is a propulsive contact. The initial 3 strokes are thought to compromise the bulk of the initial acceleration from a stationary position and are not included in this analysis[18]. Variables calculated for each cycle from 4 to the end of the data set were averaged together to provide a general representation of propulsion beyond the initial acceleration phase, termed here 'post initial acceleration' (PIA). If less than 4 strokes were recorded, key variables were not computed. For the purposes of this analysis, the following force convention was used: Fx(+) = Forward directed force; Fy(+) = Upward directed force; Fz(+) = Medially directed force; Mz(+) = Moment in the forward direction.

3.3.6 Key Variables

The following variables were calculated for each full cycle available in each trimmed trial: minimum, maximum, and average linear velocity, push frequency, stroke length, total work, peak positive power, peak positive tangential force and Mz, peak resultant force, and peak minimum Mz. Three velocity measures were selected to provide a full representation of the velocity oscillation characteristic of propulsion. Average velocity (m/s) was the defined as the average linear velocity of the wheel during the cycle. Minimum and maximum velocities were the minimum and maximum linear velocity occurring during the cycle. Push frequency and stroke length are measures of the rate and distance of force application, essentially how the person applies force. Stroke length was defined as the angular distance (degrees) traveled by the wheel during the propulsive moment portion of a contact. A propulsive moment was defined as Mz above 0.6 Nm for a minimum of 0.1s. Push Frequency (hz) was calculated as 1 second/cycle time(s). Peak resultant force and Mz were selected as measures of overall force application. A positive Resultant force (N) was defined as the vector sum of Fx, Fy, Fz [18]. Mz contributes to propulsion; conversely a negative Mz contributes to braking[30]. Measures of work, and peak power were selected based on their potential to further explain differences between the groups or surfaces[31]. Work (J) was defined as the integral of Mz over stroke length. Power (w) was defined as Mz * angular velocity. Peak positive power was the maximum positive value occurring during a cycle. Work, power, and all forces and moments were normalized to body-weight prior to analysis. Matlab[c] was used to trim the data, identify cycles, and compute variables.

Variables were calculated for right and left side separately for the older adults, then averaged together to neutralize the net impact of an undulation in the underlying floor resulting in asymmetric force application attributed to a variable cross-slope (maximum = 0.5^{o}). Unilateral data collection was common in the database data, thus the available side was selected for analysis. If both sides were available, the left side was chosen, as it was the most frequently submitted side in the database. A single trial each for tile and carpet, representative of a typical collection was entered into the analysis. For both groups contact was defined as beginning when the resultant force and resultant moment were above threshold (3N and 0.6 Nm). Thresholds were selected to ensure baseline noise was not included in calculations. A contact ended when both dropped below threshold. A search algorithm automated the identification of contact and recovery. Accuracy was verified through visual inspection, with adjustment made on a per cycle basis when necessary.

3.3.7 Statistical Analysis

SPSS[d] was used for all analysis. Data was inspected for normalcy. To determine differences between groups and surfaces, a series of 2 x 2 repeated measures ANOVAs were conducted, one for each of the key variables. Surface was entered as a within subject factor (Tile, Carpet). Group was entered as a between subject factor (Older Adult, Individual with Paraplegia). Significance was set a priori at $p \leq 0.05$. Main effects were calculated for group and surface, with an interaction group by surface. Select pairwise comparisons were used to investigate significant main effects. A Bonferroni correction was applied to control for Type I error.

3.4 RESULTS

A total of 107 individuals were available for analysis (Older Adult (OA) = 54, Individuals with Paraplegia (IP) = 54). Individuals with paraplegia were significantly younger, heavier, and taller than the older adults ($p \leq 0.05$ for all).

Table 3.1 Participant Demographics

	N	Age (yrs)	Height (m)	Propulsion experience (years)	Men (N)	Women (N)
Individuals with Paraplegia	54	42.21 (10.99)	1.79 (0.09)	13.00 (9.86)	48	6
Older Adults	53	73.74 (5.44)	1.69 (0.14)	-----	20	33

3.4.1 Main Effects

Table 3.2 contains all p values for each outcome measure and main effect. Main effects were significant for surface, group, and surface x group for select variables. Post-hoc analysis was performed where significant main effects were noted.

Table 3.2 P values for main effects. Significance set at $p \leq 0.05$. Bold indicates significant main effect

	Surface	Group	Surface x Group
Minimum Velocity (m/s)	**0.000**	**0.049**	0.666
Maximum Velocity (m/s)	**0.000**	**0.000**	0.374
Average Velocity (m/s)	**0.000**	**0.000**	0.769
Push Frequency (Hz)	**0.042**	**0.000**	0.463
Stroke Length (degrees)	0.745	**0.000**	0.190
Total Work (J/Kg)	**0.000**	**0.000**	**0.000**
Peak Pos Power (W/kg)	0.238	0.069	**0.025**
Peak Resultant (% BW)	**0.000**	0.763	**0.049**
Peak Pos Mz (% BW)	**0.000**	0.231	**0.002**
Peak Min Mz (% BW)	0.402	**0.000**	0.332

3.4.2 Surface

All individuals slowed down on carpet compared to tile (Figure 3.1), maintaining stroke length while increasing push frequency, resultant force, and peak positive Mz (Tables 3.3 – 3.5). Total work per cycle was greater on carpet than tile (Table 3.4). All peak and positive peak forces and moments were greater on carpet than tile (Table 3.5). Minimum Mz and peak positive power remained unchanged between surfaces (Table 3.5).

Propulsion across low pile carpet results in decreased velocity, increased push frequency, positive force, positive Mz, and work compared to tile. Stroke length, peak positive power, and peak minimum Mz remained unchanged across surfaces.

3.4.3 Group

Individuals with paraplegia pushed faster than older adults on tile and carpet across all measures of velocity (Figure 3.1). On both surfaces, IP group used a lower push frequency and longer stroke length than OA (Table 3.3). On a per cycle basis, IP demonstrated greater total work (Table 3.4). Peak resultant, positive Mz, and power were not statistically different between IP and OA (Table 3.5). Peak minimum Mz was greater in OA (Table 3.5).

Older adults self-selected a lower velocity than individuals with paraplegia; achieved with a higher push frequency, shorter stroke length and similar resultant force, positive Mz and power. OA demonstrated a greater negative Mz, an indication perhaps their lower velocities are due in part to unintentionally applied braking forces and moments. Work, which was greater in IP, seems attributable to a longer stroke length.

3.4.4 Surface x Group

Surface by group interactions were significant for total work, peak positive power, peak resultant force, and Mz. All measures of velocity; push frequency, stroke length, and peak minimum Mz were not significant for surface by group interactions.

IP demonstrated increased work and power on carpet versus tile while OA remained unchanged (work: $p \leq 0.000$, $p = 0.128$; power $p=0.015$, $p=0.444$). In addition power was greater for IP than OA on carpet, but not tile ($p = 0.012$, $p = 0.408$). IP demonstrated greater increases in resultant force, and positive Mz than OA on carpet versus tile. However, resultant force was statistically non-distinct between the groups on both surfaces (tile, $p = 0.260$; carpet, $p = 0.594$). On carpet, but not tile, IP demonstrated a larger positive Mz (carpet, $p = 0.016$; tile, $p = 0.635$). OA demonstrated increased minimum Mz on carpet, while the IP group remained unchanged ($p=0.016$, $p=0.999$).

To summarize, OA and IP respond in different ways to increased surface difficulty. Both groups decrease their self-selected velocity by a similar magnitude. When force and moments are examined, both groups apply greater peak resultant force and positive Mz on carpet versus tile, but only on carpet do the groups become statistically different, with IP demonstrating greater Mz compared to OA. Resultant force remained similar between the groups on carpet versus tile. Additionally, OA demonstrate a greater peak minimum Mz on carpet versus tile, an indication of increased braking moment, while IP use similar values on both surfaces. Finally, IP demonstrate greater work per cycle and peak positive power on carpet versus tile while OA use similar amounts of total work and peak positive power on both surfaces.

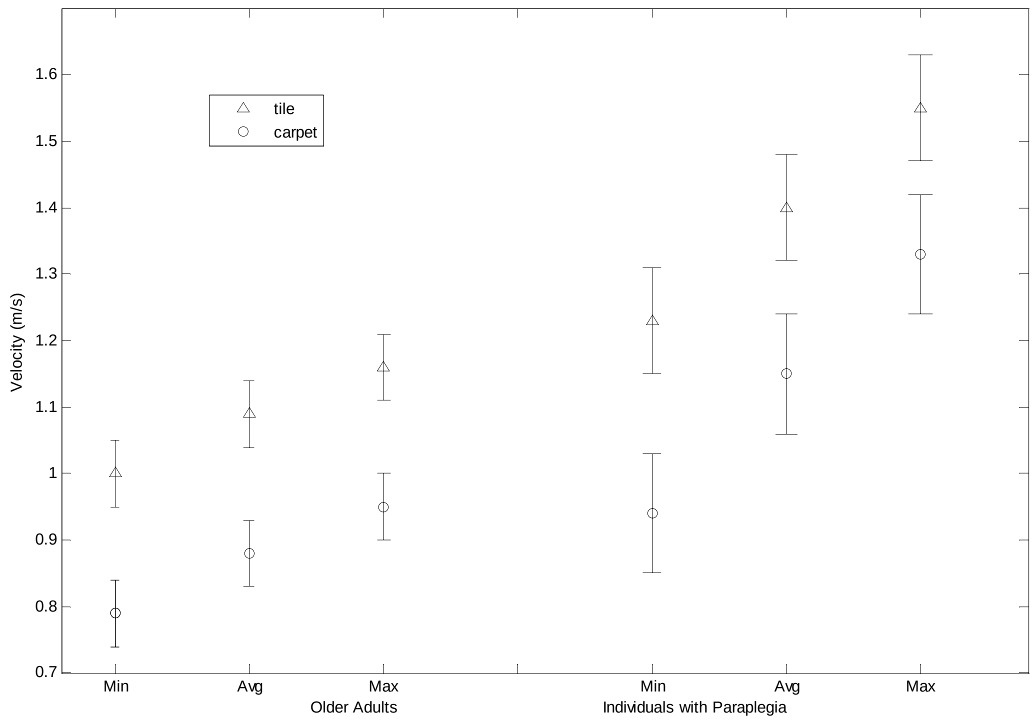

Figure 3.1 Minimum, Average, and Maximum Velocity for older adults and individuals with paraplegia on tile and carpet

Table 3.3 Unadjusted Means and Standard Deviations, Push Frequency, Stroke Length, Distance. * Significant main effect surface. † Significant main effect group. + Significant main effect surface x group.

	Push Frequency (Hz) *,†		Stroke Length (degrees) †		Distance (m) *,†	
	Tile	**Carpet**	**Tile**	**Carpet**	**Tile**	**Carpet**
Individuals with Paraplegia	0.97 (0.24)	0.98 (0.21)	85.25 (14.52)	85.98 (14.79)	1.52 (0.47)	1.21 (0.45)
Ambulatory Older Adults	1.17 (0.30)	1.19 (0.28)	54.58 (13.32)	53.38 (12.36)	1.02 (0.39)	0.78 (0.25)

Table 3.4 Unadjusted Means and Standard Deviations, Total Work, Peak Positive Power. * Significant main effect surface. † Significant main effect group. + Significant main effect surface x group.

	Total Work (J/Kg) *,†,+		Peak Pos Power (W/Kg) +	
	Tile	**Carpet**	**Tile**	**Carpet**
Individuals with Paraplegia	15.98 (7.26)	21.16 (7.44)	8.44 (4.54)	8.64 (3.82)
Ambulatory Older Adults	10.40 (4.87)	11.71 (4.41)	7.20 (3.24)	6.90 (3.02)

Table 3.5 Unadjusted Means and Standard Deviations, Peak Resultant Force, Peak Tangential Force, Peak Positive Mz, Peak Minimum Mz. * Significant main effect surface. † Significant main effect group. + Significant main effect surface x group.

	Peak Resultant (% BW) *, +		Peak Pos Tangential (% BW) *, +		Peak Pos Mz (% BW) *, +		Peak Min Mz (% BW) †	
	Tile	**Carpet**	**Tile**	**Carpet**	**Tile**	**Carpet**	**Tile**	**Carpet**
Individuals with Paraplegia	9.03 (2.83)	10.61 (2.90)	6.92 (2.28)	8.73 (2.34)	1.85 (0.61)	2.34 (0.63)	-0.17 (0.15)	-0.17 (0.08)
Ambulatory Older Adults	9.20 (2.78)	10.30 (2.69)	6.82 (1.93)	7.71 (1.81)	1.82 (0.51)	2.06 (0.48)	-0.26 (0.11)	-0.28 (0.14)

3.5 DISCUSSION

As expected, novice, ambulatory older adults demonstrated a lower self-selected velocity than younger individuals with paraplegia. Unexpectedly, both groups used similar peak forces and moments to achieve different velocities, an indication perhaps of differences in propulsion efficiency. A higher push frequency concomitant with a shorter stroke length was characteristic of older adults compared to individuals with paraplegia. When confronted with a surface that provided greater external resistance, OA and IP responded differently. Both groups slowed down while maintaining similar peak resultant force. However, IP responded to the increased resistance by increasing their peak positive Mz by a larger magnitude than OA. When surface difficulty increased, only the IP group responded with an increase in work and power. OA demonstrated no change in work or power between tile and carpet. This may indicate a lack of capacity in OA to respond to challenging propulsion environments. In contrast, the ability of IP to increase work could be indicative of an ability to respond to increased demand. Given these findings, an increase in external demand which does not result in an increase in work may be an indication that an individual is functioning at the upper end of their physiologic capacity.

3.5.1 Defining Successful Mobility

Independent mobility is a reasonable goal for all users of wheeled mobility. However, quantified numbers representing "successful" independent mobility would differ by the population of interest. As an example, it would not be unreasonable to set as a benchmark the ability to easily self-propel at the comfortable walking velocity of the client's peers. In the case of our individuals with paraplegia, this would be 1.39 - 1.46 m/s[24]; for our older adults it would be 1.27

– 1.33[24]. On both surfaces the average velocity of our older adults fell short of this benchmark, with a greater difference on the carpet. In a clinical setting, this gap would represent an opportunity for intervention. In comparison, the individuals with paraplegia demonstrated a self-selected speed generally on par with their ambulatory peers, an indication that their overall mobility on low resistance level surfaces was comparable. For this group further evaluations would be valuable to identify areas of concern in their propulsion technique or wheelchair configuration which might contribute to upper extremity injury [32].

3.5.2 Selection of Wheeled Mobility

Given the goal of independent mobility, the question arises as to which mobility interventions would be most efficacious. Rehabilitation professionals understand that an individual's physical and social environments are important determinants in the selection of a mobility device. Powered mobility can only facilitate mobility to the extent that it can be used in a given environment. If transportation or structural barriers prevent the use of powered mobility, then a manual wheelchair represents the most appropriate solution. In such a case optimized configuration (such as axle position), custom fit, propulsion training, and strength training could serve to decrease the metabolic demand of propulsion. Lighter weight, adjustable wheelchairs are commonly provided to individuals with SCI [33], with the expressed purpose of facilitating independent mobility by decreasing the physical demand of propulsion. Currently, the typical chair provided to older adults is heavy, is non-adjustable, and is not well fitted [6-8], which undoubtedly contributes to reports of difficulty with or an inability to self-propel [8,9]. Research has demonstrated improved service delivery and customization of wheeled mobility leads to improved function and outcomes in older adults [7,34].

However, not all older adults posses the capacity to become a successful independent propeller, even given the most optimally configured manual wheelchair. In our group 52% obtained an average velocity below that of the typical walking velocity of their peers. Indeed the average maximum for the entire group was lower than the customary walking velocity for their peers. Even in our cohort, who was fully ambulatory, an average increase of 0.17 m/s would be required to facilitate the lower end of the customary walking range on tile. The gap widens as surface difficulty increases and undoubtedly would enlarge as physical frailty increased. The degree to which an optimally configured wheelchair in conjunction with propulsion, skills, and strength training can improve self-propulsion in older adults remains to be determined. Specifically, research exploring the benefits of decreased wheelchair weight, optimized axle position, and propulsion training in older adults is warranted. Such research could provide justification to facilitate the provision of the lightest available wheelchairs through service delivery techniques that ensure proper fit and training has been achieved. Additionally, researchers could identify the subset of older adults for which powered mobility represents the best method to achieve successful independent mobility.

3.5.3 Variables sensitive to differences in Users and Surfaces

Examination of our results provides preliminary indications that select variables describing propulsion are sensitive to differences in surface resistance, while others are sensitive to differences in the user (Table 3.2). The most robust candidates would be those which were significant at $p \leq 0.01$ for both surface and group main effects; per cycle maximum velocity, average velocity, and total work. Of note, total work was the only variable to demonstrate a significant main effect for surface, group, and surface by group interaction. Total work per cycle should be further evaluated to determine its sensitivity to detect changes after clinical interventions, such as alterations in wheelchair configuration and propulsion training.

In summary, we believe these results highlight the need for researchers and clinician to be aware of the impact of the user group of interest and surface type on propulsion parameters. Clinicians may wish to exploit the impact of surface on changes in self-selected velocity and work per cycle to assist in determining the capacity of their client to respond to increasing surface difficulty. A decrease in velocity coupled with little or no change in work when confronted by a surface with increased resistance may be indicative of a client who has little or no capacity to respond to increased environmental demands. Researchers may wish to target different subsets of variables if the question of interest is comparing two groups of users versus comparing two surfaces in one group. Selection of variables which are not sensitive to the research question could lead to type II errors. Further research is warranted to identify variables which are sensitive to differences in user groups, propulsive conditions, propulsion training, and wheelchair configuration. Such variables would be of immense value to the practicing clinician.

3.5.4 Strengths and Limitations

It is unknown if older adults who are part time users of manual wheelchairs demonstrate characteristics associated with experienced wheelchairs users or if due to their intermittent need and use, they fail to develop efficient propulsion techniques. Thus, it cannot be determined how similar or dissimilar our cohort is from an older adult who is an intermittent self-propeller. However, since older adults and individuals with paraplegia exhibited similar peak resultant force coupled with different tangential forces and Mz, resulting in different velocities, with older adults propelling slower with lower peak tangential force and Mz, we are comfortable in inferring that older adults were less efficient propellers. Use of older adults who were experienced in wheelchair propulsion may have provided additional insight. Our subjects were provided with 6 minutes of practice prior to completing propulsion over the tile and carpet, so they were not without any experience. Additionally, if the reason for using a wheelchair is functional declines in strength, then our fairly robust cohort represents an optimistic representation of the capacity of such users. Moreover, our subjects are reasonable representations of individuals who might be forced to use a wheelchair due to a fall related pelvic or femur fracture. Such adults are fully ambulatory up to the point of needing a wheelchair[4]. Lastly, our cohort self-selected a velocity below that of a typical walking speed. We feel it is improbable that a frailer cohort would propel any faster. If anything, we would expect a further reduction in velocity in a frailer group of older adults.

Our study design allowed for the detection of propulsion differences, but did not allow for a determination of the cause of the differences. Propulsion differences between older adults and individuals with paraplegia could be attributed to user characteristics including, but not limited to: dissimilarities in age, propulsion experience, gender distribution and strength/function. For each of these items, our groups were dissimilar. The IP group was composed predominantly of middle aged men with paraplegia who had over a decade of propulsion experience. As a group our older adults were over 30 years older, with minimal propulsion experience and were dominated by women. Gender and age effects exist for many domains of physical function; it is reasonable to assume such effects exist for propulsion. In addition existing research has indicated experienced and novice users select different propulsion strategies. Determination of the cause of propulsion differences could be used to develop target interventions for subsets of users.

For the purpose of this analysis, we standardized, but did not optimize horizontal axle position in our older adults. The position we selected was a clinically relevant position, as it was a position obtainable on a commercially available wheelchair. The selected position, while not optimized to the individual is a more favorable position than that available on the typical chair an older adult propels. Therefore we believe that while an optimized position could have improved the mechanics of our cohort, our selected position is clinically reasonable and applicable, representing a more optimized configuration that what an older adult typically receives. Wheelchair configuration for the international dataset is not available, thus we cannot speculate on the degree to which each individual with paraplegia was seated in an optimized configuration or on the impact attributable to unknown differences in configuration.

Finally, the non-standardized nature of the international data set limits our ability to attribute causality to the specific differences between our cohort and individuals with paraplegia. While we believe our surface selection mirrors that contained in the dataset, we cannot be fully assured of the similarities. However, we believe the inherent variability within the data set due to non-standardization captures clinically relevant variance. Consequently, we believe the differences between our older adults and individuals with paraplegia are genuine and clinically relevant. In addition, data drawn from this dataset represents a newly available criterion for researchers to determine the clinical comparability and generalizability of their research outcomes, especially if their data collection methods mirror that of the data set.

3.6 CONCLUSION

Relatively healthy, ambulatory older adults self-select a propulsion velocity lower than typical walking velocity. Compared to a younger group of individuals with paraplegia, they use similar forces to achieve a lower velocity, indirect evidence of lower propulsion efficiency. When confronted with increased external resistance provided by a carpeted surface, both groups slowed down. Individuals with paraplegia responded with an increase in per cycle total work and peak power primarily by increasing their force application. In contrast older adults maintained their work per cycle. Prescription of the lightest weight chair combined with a user optimized configuration and propulsion training may serve to increase efficiency and minimize the impact of increased surface difficulty. Finally, not all variables are sensitive to differences in surface or users. Further research to define the sensitive and specificity of propulsion parameters is warranted.

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Suppliers

[a]SmartWheel: Three Rivers Holdings LLC, 1826 West Broadway Rd. Suite 43
Mesa, AZ 85202

[b]Model X: TiLite 1426 East Third Avenue, Kennewick, WA 99337

[c]Matlab: The MathWorks, Inc., 24 Prime Park Way, Natick, Massachusetts 01760-1500

[d]SPSS: SPSS Inc., 233 South Wacker Drive, 11th Floor, Chicago, Illinois 60606-6307

4. WHEELCHAIR WEIGHT, AXLE POSITION, SURFACE TYPE AND PROPULSION IN OLDER ADULTS

Authors: TBD

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4.1 ABSTRACT

Objective: To examine the impact of surface type, wheelchair weight, and rear axle position on propulsion biomechanics of older adults during cycles occurring after the initial acceleration period from a stationary position.

Design: Randomized controlled trial

Setting: Biomechanics laboratory

Participants: 53 ambulatory older adults (65+)

Intervention: Participants propelled four wheelchair configurations from a stationary position to a self-selected velocity over four surfaces; hallway tile, low pile carpet, high pile carpet, and up an ADA compliant ramp. Wheelchair weight conditions were un-weighted and weighted (+20lbs). Axle positions were posterior (most rear position on axle plate) and anterior (most forward position +8cm). Configurations included un-weighted anterior (UA), weighted anterior (WA), un-weighted posterior (UA), and weighted posterior (WP). Chair and surface order were randomized.

Main Outcome Measures: Self-selected velocity, push-frequency, stroke length, maximum resultant and tangential force.

Results: Typically, as surface difficulty or chair weight increased, velocity decreased. Controlling for velocity, push frequency, resultant and tangential force increased as surface difficulty increased; heavier chairs resulted in decreased stroke length coupled with increased resultant, and tangential force across all surfaces; and posterior axle positions resulted in increased self-selected velocity. Controlling for velocity, posterior axle positions resulted in increased forces across all surfaces; increased weight muted the benefits of an anterior axle position; and WP demonstrated the highest forces and UA the lowest. Surface difficulty modified the relative impact of weight and axle position.

Conclusion: Wheelchair weight and axle position impose distinct changes on biomechanics. Surface difficulty modifies the impact of weight and axle position. Anterior axle position does not fully mitigate increased weight. Older adults consistently self-selected low velocities, regardless of chair configuration. Heavy chairs with posterior axle positions appear to be a poor mobility choice for older adults.

Key Words: Older Adult, Wheelchair, Biomechanics

4.2 INTRODUCTION

In the United States, older adults (65+) are the largest group of manual wheelchair users, yet are among the least studied (1;2). In regards to wheeled mobility, older adults typically receive a standard or depot wheelchair (1-3), most likely because the ability to ambulate disqualifies them from more advanced wheelchairs (4). The Center for Medicare and Medicaid services (CMS) classifies manual wheelchairs broadly on two basic characteristics; weight and adjustability (5). Weight refers to the heaviness of a chair without riggings, such as footrests, armrests, cushions, or clothing guards. Adjustability generally refers to the ability to select the position of the rear wheels, either through a pre-ordered location or after delivery through a method of the manufactures choice. Additional classification criteria such as seat height, seat dimensions, and frame strength exist, but are beyond the scope of this manuscript and function as sublevels within the two afore mentioned criteria (5;6).

Wheelchairs commonly prescribed for older adults and individuals with SCI typify the extremes of the weight and adjustability criteria. A K0005 is representative of the type of wheelchair an individual with a spinal cord injury (SCI) receives for their mobility needs (7). By definition, a K0005 weighs less than 30 pounds, with multiple levels of customization. Some titanium versions can weigh less than 25lbs, complete with all options. Such chairs are designed to facilitate mobility through decreased weight, optimized rear axle position and custom fit. In comparison, the typical chair an older adult receives is classified as a K0001 (3;6;8). Compared to K0005, K0001s lack rear axle adjustability, custom fit, are heavy (>36lbs) and are designed as temporary mobility solutions. It is not surprising therefore that older adults report inability to self-propel a wheelchair as the most common reason why they choose not to use a wheelchair (9).

Benefits of an optimized axle position include improved metabolic efficiency, increased stroke length, and decreased force production (10-15). The impact of chair weight has yet to be evaluated; however any decrease in weight would be expected to at minimum result in decreased force at a given velocity. Graded surfaces, or simulated grades, have been shown to decrease velocity and stroke length concomitant with increased force (16;17). Additional surfaces have been minimally evaluated, but generally, as surface difficulty increases, velocity decreases while force increases (16). Biomechanics of propulsion in older adults have been investigated in a limited manner with small numbers under conditions which were not intended to mimic propulsion over common surfaces (18;19). All of these studies have advanced our understanding and knowledge base, providing much of the rationale for the current study, but possess inherent limitations in the sample or methodology which have limited their ability to be generalized to older adults.

Therefore, the purpose of the study was to determine the impact of surface type, wheelchair weight and rear axle position on biomechanical parameters describing overground propulsion in older adults. Parameters of interest include self-selected average velocity, push frequency, stroke length, maximum resultant and tangential force. We hypothesize the following: 1) as surface resistance increases, velocity, push frequency, and stroke length will decrease and maximum resultant and tangential force will increase, 2) heavier chairs will result in decreased velocity and push frequency with increased stroke length and maximum resultant and tangential force and 3)

Posterior axle positions, controlling for velocity, resulted in decreased push frequency and increased resultant and tangential force.

4.3 METHODS

4.3.1Participants

Fifty-three older community dwelling adults were recruited through flyers at local senior citizen centers, interest groups, bring a friend strategies, and IRB approved research registries (Men = 20, Women = 33). All subjects gave written informed consent prior to participation in the study. This research protocol was approved by the Institutional Review Boards (IRB) of the VA Pittsburgh Health Care System and the University of Pittsburgh. To be eligible for participation, subjects had to self-report 1) the ability to walk without human assistance, 2) the ability to stand-up from a chair, 3) the ability to push themselves in a wheelchair, 4) weight as less than 251 lbs, and 5) score greater than 22 on the mini-mental state exam. Exclusion criteria included self-reported history of stroke or a diagnosis of Parkinson’s or Alzheimer’s. Participants reported minimal experience with wheelchair propulsion.

4.3.2 Initial Wheelchair Adjustment

Three TiLite[a] titanium folding chairs (Model X) were secured for this study. The seat dimensions of the three chairs were 15 x 16, 17 x 17, and 19 x 18 with 0 degrees of seat inclination and camber. Equipped with fabric seat and back upholstery, a 2 inch foam cushion, plastic removable side guards, anti-tippers, pneumatic rear wheels, solid 4 inch casters, and aluminum hand rims, each chair weighed 25lbs. The three seat dimensions allowed us to accommodate a variety of individuals.

Each individual was seated in the smallest possible width wheelchair. When possible, rear axle position was adjusted vertically so each subject demonstrated an elbow flexion angle between 100 and 120 degrees when seated in the chair with their hand placed at top center of the pushrim. Footrests were adjusted to provide support for the feet and thighs, as allowed by the design of the chair. Front and rear seat heights were kept as equal as possible to maintain 0 degrees of seat inclination. Front caster angle was adjusted to minimize caster trail.

4.3.3 Wheelchair Test Configurations

Two rear axle positions (anterior and posterior, eight cm difference) and two weights (un-weighted and weighted, +20lbs) were tested. The four test configurations were as follows: un-weighted, anterior axle position (UA); weighted, anterior axle position (WA); un-weighted, posterior axle position (UP); and weighted, rear axle position (WP). Chair configuration WP (weighted, posterior axle) is intended to simulate the weight and axle position of a K0001. All four configurations were equipped with anti-tippers for safety purposes.

The forward axle position was the most anterior rear axle position allowed by the manufacturer's bracket system; representing a more optimal configuration (13;14). The rear axle position was eight centimeters posterior; representing the most posterior axle position allowed by the manufacturer's bracket system and; similar to that found on standard wheelchairs. The weight off configuration was the natural weight of the wheelchair (25lbs). The weight on configuration involved the addition of 20lbs of weight to the chair (chair + weight = 45lbs). This addition mimics the weight differential between the lightest weight chairs available and the weight of many standard or "depot" chairs. The weight was added in a manner that maintained the relative weight distribution between the front and rear wheels with the wheels in the most posterior position. Participants propelled all 4 configurations in a randomized order. Prior to propelling the 4 wheelchairs on each surface, each participant completed a 6 minute propulsion task in configuration WP, which served as a novel functional assessment and an opportunity to practice Participants transversed each surface once in each chair configuration for a total of 16 trials per participant.

4.3.4 Propulsion Surfaces

Four surfaces were selected for the testing: hallway tile (T, 12.0 m), low pile carpet on concrete with no pad (LC, 7.3 m), high pile carpet with a typical residential pad on concrete (HC, 7.3 m) and a wooden uncarpeted ADA compliant ramp (R, 2.5m, 8%). These surfaces represent surfaces thought to be commonly encountered in home environments. Surface order was randomized for each chair for each participant.

4.3.5 Propulsion Data Collection

SmartWheels[b] provided kinetic bilateral data collection at 240 Hz. Currently, the SmartWheel provides the only commercially available method by which to measure forces and moments used to self-propel a wheelchair. A SmartWheel weighs ~11 lbs. With bilateral SmartWheels attached, the test weight of the chairs were 40lbs and 60lbs and the weight distribution between front and rear wheels was 28% and 73% with the axle in the posterior position; 23% and 77% with the axle in the anterior position. Without the SW, the weight distribution of the chairs was 36% and 64% respectively with the axle in the posterior position; 20% and 80% respectively with the axle in the anterior position.

Participants were instructed to begin in a stationary position, hands in their lap, accelerate to a comfortable velocity, continuing until they exited the surface or were instructed to stop. Participants began on the surface for T, LC, and HC. For the ramp participants began on the level ground directly in front of the ramp with the front casters within 3 inches of the beginning of the ramp. On all surfaces data collection was initiated before initial contact and was terminated before the chair exited the test surface (LC/HC) or the marked data distance (T/R). Investigators terminated data collection early if it became obvious the participant was experiencing difficulty completing a particular surface in a given chair. Participants completed each surface once in each chair configuration for a total of 16 trials per participant.

4.3.6 Kinetic Data Reduction

Ramp, high carpet, low carpet, and tile trials were trimmed prior to data analysis to remove deceleration that occurred as participants approached the end of the surface or data collection area. The initial 1.5m of each ramp trial was trimmed to remove data which captured the ascent from the level ground onto the ramp. Transitioning from a level surface onto a ramp represents a unique aspect of propulsion beyond the scope of this initial analysis. After all trimming, the following distances were available for analysis: Ramp, 2.0m; Low Pile Carpet, 6.4m; High Pilr Carpet, 6.4m; and Tile; 9.0m.

4.3.7 Biomechanical Parameters

A cycle was defined as the period encompassing a propulsive contact and the subsequent recovery. A stroke was defined a propulsive contact (positive Mz). On HC, LC, and T, variables for the initial three full cycles were calculated separately, but are not included in this analysis. The initial 3 strokes are thought to compromise the bulk of the initial acceleration from a stationary position (16). On the ramp, cycles 1 -3 represent the first three cycles after the chair had ascended onto the ramp. Variables calculated for each cycle from 4 to the end of the data set were averaged together to provide a general representation of propulsion beyond the initial acceleration phase (PIA). If less than 4 strokes were recorded, variables representing PIA were not computed. If a maximum of 4 strokes were recorded, PIA represented the fourth stroke. All analysis was computed using the variables describing PIA.

4.3.8 Key Kinetic Variables

The following variables were calculated for each full cycle available in each trimmed trial: average linear velocity, push frequency, stroke length, maximum resultant force, and maximum tangential force. Maximum values were selected based on existing overground research (14;16).Variables were calculated for right and left side separately, then averaged together to neutralize the net impact of an undulation in the underlying floor resulting in asymmetric force application attributed to a variable cross-slope (average 0.2 degrees). A cycle encompassed the time between the onset of a pushrim contact to the last sample prior to the next contact. A contact was defined as beginning when the resultant force and resultant moment were above threshold (3N and 0.6 Nm). A contact ended when both dropped below threshold. A search algorithm automated the identification of contact and recovery. Accuracy was verified through visual inspection, with adjustment made on a per cycle basis when necessary. Average velocity was the average linear velocity (m) of the wheel during the cycle. Stroke length was defined as the angular distance (degrees) traveled by the wheel during the propulsive moment portion of a contact. A propulsive moment was defined as Mz above 0.6 Nm for a minimum of 0.1 s. Push

Frequency (hz) was calculated as 1 second/cycle time(s). Resultant force (N) was defined as the vector sum of Fx, Fy, Fz (16). Tangential (N) was calculated as Mz/pushrim radius (0.2667 m). Resultant and tangential were normalized to body weight prior to analysis. Matlab[c] was used to trim the data, identify cycles, and compute variables.

4.3.9 Statistical Analysis

SPSS[d] was used for all analysis. Data was inspected for normalcy. To determine differences between surfaces, wheelchair weight, axle position, and cycle, a series of mixed models were conducted, one for each of the outcome measures: velocity, push frequency, stroke length, resultant force and tangential force. First order auto regression was used to model the covariance matrices. Fixed factors and repeated measures included surface (levels = tile, low carpet, high carpet, ramp), wheelchair weight (levels = UA & UP= off, WA & WP= on) and axle position (levels = UA & WA = anterior, UP & WP = posterior). Random factors included subject and number of cycles used to compute the PIA average. Velocity was input as a covariate when appropriate due to its known relationship to the remaining outcome measures. Significance was set *a priori* at $p \leq 0.05$. Main effects were calculated for surface, wheelchair weight, axle position, with interactions entered including wheelchair weight by axle position, surface by wheelchair weight, and surface by axle position. Select pairwise comparisons were used to investigate significant main effects. A Bonferroni correction was applied to control for Type I error.

4.4 RESULTS

A total of 53 older adults participated (Men = 20, Women =33). Average age was 73.6 years (+/-5.4), ranging from 65 to 87. Participants were generally overweight (BMI = 27.6, +/-5.1, height = 1.7m +/-0.1, weight = 76.5kg, +/-16.7). Mean elbow angle after initial fitting was 107.1^{o} +/- 6.3^{o} (full extension = 180^{o}). We were unable to achieve an elbow angle within the specified range of 100^{o} – 120^{o} with one subject due to their anthropometrics (elbow angle = 125^{o}).

4.4.1 Surface

Significant main effects were found for all outcome measures ($p \leq 0.000$ for all) (Table 4.1). All pairwise comparisons for velocity were different ($p \leq 0.000$ for all). As surface difficulty increased, individuals self-selected a slower average velocity (T>LC>HC>R). Controlling for velocity, push frequency was lower on easier surfaces than harder surfaces, while resultant and tangential force generally increased with increasing surface difficulty.

Table 4.1 Unadjusted means (standard deviation) for each surface across all configurations and cycles. Results of pairwise comparisons between Tile and all other surfaces are indicated by p values in the appropriate cell. Velocity entered as a covariate for push frequency, stroke length, maximum resultant and tangential force. Arrows indicate the direction of change compared to Tile when adjusted for velocity. Significant pairwise comparisons for all possible comparisons are indicated in the column headers as follows (p≤0.05): * Tile and Low Carpet different, ** Tile and High Carpet different, * Tile and Ramp different, + Low and High different, ++ Low and Ramp different, # High and Ramp different**

	Average Velocity (m/s) *,**,***,+,++,#	**Push Frequency (Hz)** *,**,***,+,++,#	**Stroke Length (degrees)** **,***,+,++,#	**Maximum Resultant Force (% BW)** **,***,+,++,#	**Maximum Tangential Force (% BW)** *,**,***,+,++,#
Tile	1.05 (0.23) Reference	1.18 (0.30) Reference	55.35 (13.51) Reference	9.33 (2.74) Reference	6.89 (1.95) Reference
Low Carpet	↓ 0.87 (020) p=0.000	↑ 1.19 (0.28) p=0.000	↔ 53.76 (12.50) p=1.000	↔10.38 (2.69) p=0.143	↑ 7.83 (1.97) p=0.003
High Carpet	↓ 0.45 (0.17) p=0.000	↑ 0.94 (0.24) p=0.000	↑ 54.42 (13.47) p=0.000	↑14.72 (2.53) p=0.000	↑ 11.52 (2.08) p=0.000
Ramp	↓ 0.37 (0.15) P=0.000	↑ 0.92 (0.23) p=0.000	↑ 48.31 (12.09) p=0.000	↑16.79 (2.85) p=0.000	↑ 13.31 (2.02) p=0.000

4.4.2 Wheelchair Weight

Significant main effects were found for velocity, resultant force, and tangential force (p=0.000 for all) (Table 4.2). Main effects were not significant for push frequency (p = 0.183) or stroke length (p=0.074). Increased weight resulted in a decreased self-selected velocity. Controlling for velocity magnified the difference in resultant and tangential force between weighted and un-weighted conditions.

Table 4.2 Wheelchair Weight. Unadjusted means (standard deviations) for each axle position across both axle positions and all surfaces. Unadjusted means and standard deviations across all surfaces and cycles. Velocity entered as a covariate for push frequency, stroke length, maximum resultant and tangential force. * Significant difference between wheelchair weights p ≤0.05

	Average Velocity (m/s) *	Push Frequency (Hz)	Stroke Length (degrees)	Maximum Resultant Force (% BW) *	Maximum Tangential Force (% BW) *
Weight On (WA & WP)	0.68 (0.34)	1.05 (0.30)	53.06 (13.34)	12.93 (4.13)	10.04 (3.37)
Weight Off (UA & UP)	0.71 (0.34)	1.08 (0.30)	53.18 (13.04)	12.41 (3.96)	9.51 (3.16)

4.4.3 Axle Position

Significant main effects were found for stroke length, resultant force and tangential force (p = 0.011, p=0.000, p = 0.000) (Table 4.3). Velocity and push frequency were not significant for main effects. (p = 0.271, p = 0.148). The anterior axle position resulted in a longer stroke length and decreased forces when controlling for velocity.

Table 4.3 Axle Position. Unadjusted means (standard deviations) for each weight condition across both axle positions and all surfaces. Unadjusted means and standard deviations across all surfaces and cycles. Velocity entered as a covariate for push frequency, stroke length, maximum resultant and tangential force.*Significant difference between axle positions p ≤0.05

	Average Velocity (m/s)	Push Frequency (Hz)	Stroke Length (degrees) *	Maximum Resultant Force (% BW)*	Maximum Tangential Force (% BW) *
Posterior Axle (UP & WP)	0.70 (0.34)	1.06 (0.29)	52.81 (12.96)	12.98 (4.22)	9.90 (3.36)
Anterior Axle (UA & WA)	0.69 (0.34)	1.07 (0.30)	53.43 (13.41)	12.36 (3.86)	9.56 (3.19)

4.4.4 Weight by Axle Position Interactions

Main effects were significant for stroke length (p = 0.021) when controlling for velocity. There were no significant main effects for velocity, p = 0.426; push frequency, p = 0.271; resultant force, p=0.601; or tangential force, p =0.747, when controlling for velocity as appropriate (Table

4.4). Controlling for velocity, configuration WA had a longer stroke length than all other configurations, which remained statistically non-distinct from each other.

Table 4.4 Weight by Axle Position. Unadjusted means and standard deviations for each chair configuration across all surfaces. Velocity entered as a covariate for push frequency, stroke length, maximum resultant and tangential force. Significant pairwise comparisons for all possible comparisons are indicated in the column headers as follows for push frequency and resultant force (p≤0.05): * UA and WA different, ** UA and UP different, * UA and WP different, + WA and UP different, ++ WA and D different, # UP and WP different. % indicates pairwise comparisons were not performed.**

	Average Velocity (m/s)%	Push Frequency (Hz)%	Stroke Length (degrees) **,+,#	Maximum Resultant Force (% BW) %	Maximum Tangential Force (% BW) %
UA anterior axle weight off	0.70 (0.34)	1.08 (0.30)	53.71 (13.33)	12.02 (3.69)	9.24 (3.02)
WA anterior axle weight on	0.73 (0.34)	1.08 (0.29)	52.65 (12.76)	12.80 (4.18)	9.77 (3.29)
UP posterior axle weight off	0.67 (0.34)	1.06 (0.30)	53.14 (13.52)	12.71 (4.00)	9.87 (3.33)
WP posterior axle weight on	0.68 (0.35)	1.03 (0.30)	52.96 (13.17)	13.17 (4.26)	10.21 (3.42)

4.4.5 Surface by Weight

A main effect existed for resultant ($p = 0.000$) and tangential force ($p=0.000$). Main effects were non significant for velocity, push frequency, and stroke length, and resultant force ($p = 0.874$, $p = 0.915$, $p = 0.215$). As surface difficulty increased, resultant and tangential force increased at different rates for the weight on and off conditions.

4.4.6 Surface by Axle Position

Main effects were significant for velocity, resultant force, and tangential force; controlling for velocity ($p = 0.030$, $p = 0.000$, $p = 0.000$). Main effects were not significant for push frequency or stroke length ($p = 0.780$, $p = 0.321$). Tile was the sole surface where the posterior configuration was propelled faster than the anterior ($p=0.021$). On the remaining surfaces the configurations were statistically non distinct. Resultant and tangential force increased at different rates as surface difficulty increased (Figure 4.1).

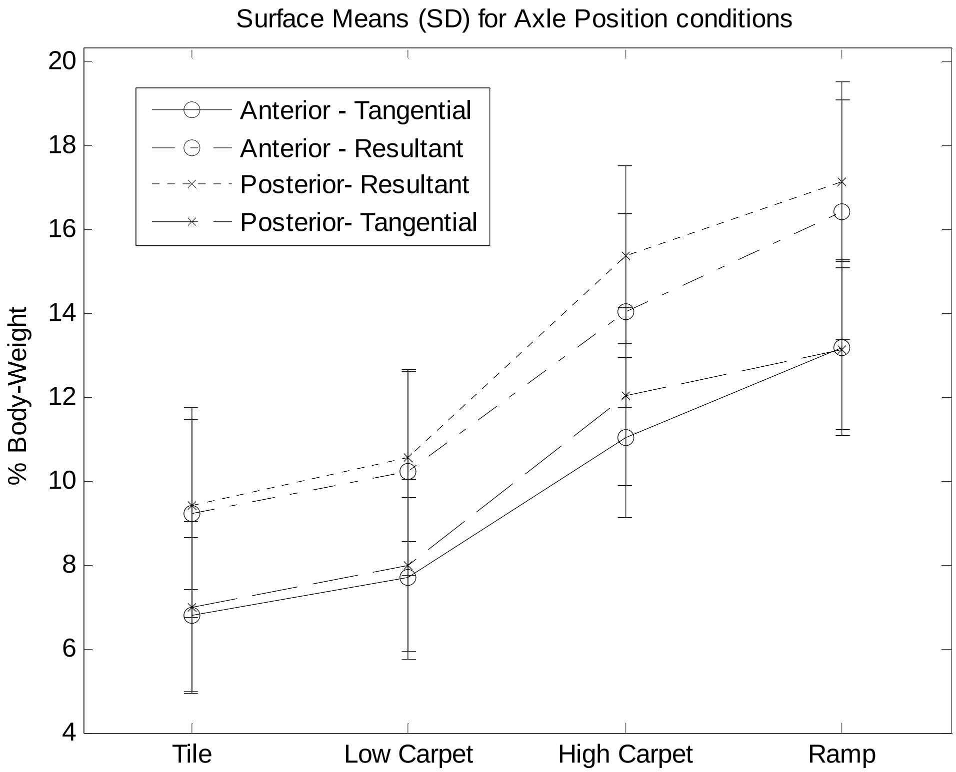

Figure 4.1 Means (SD) for each axle position across both weight conditions for each surface

4.5 DISCUSSION

Our results are the first to examine in a controlled manner the impact and interaction of surface type, wheelchair weight, and axle position on propulsion biomechanics in any population. We have demonstrated that primary variables of interest change across surface types as surface difficulty increases. Wheelchair weight and axle position each have a measurable effect on variables commonly used to describe propulsion. Surface type interacts with weight and axle position differently, highlighting the individual benefits imposed by a lighter weight chair or a more optimized axle position. When unique combinations of weight and axle position are examined, the impact on such variables becomes muted when evaluated without regard to surface type. However, as surface difficulty increases, either via a ramped condition or heavy carpet, the specific impact of a weight and axle position begin to once again manifest.

We confirmed all hypotheses in part. Generally as surface difficulty increased, velocity, and stroke length decreased, while maximum resultant and tangential force increased as hypothesized. Contrary to our hypothesis push frequency increased rather than decreased. As hypothesized, increased weight resulted in decreased velocity and controlling for velocity, stroke length and forces increased. Push frequency decreased and stroke length remained unchanged with increased weight, while an increase was hypothesized for both. Hypothesis three was fully confirmed. Controlling for velocity, forces increased, push frequency decreased. Contrary to hypothesis three, velocity remained unchanged in response to axle position. The effects of wheelchair weight and axle position appear to be subtle, easily obscured or missed if they are not rigorously controlled and examined in context of surface type.

Benefits of an optimized axle position have been documented and confirmed in individuals with spinal cord injury, thus our results associated with axle position were not unexpected, even in a sample of inexperienced older adults (10-15). A lighter chair in theory would result in decreased force use as a minimum result. This study is the first to confirm wheelchair weight does indeed affect propulsion biomechanics and that the impact extends beyond decreased force. Furthermore, the impact of weight appears to become more distinct on surfaces such as high carpet and a ramp. Weight appears to mute the benefits of an anterior axle position and magnify the impact of a posterior position. Surface type seems to modify of the impact of weight and axle position. These results represent an initial attempt to unravel the complex and subtle interaction between the weight of a chair, its axle position, and the surface a user propels across.

4.5.1 Implications for Older Adults

Of greatest concern to older adults are the functional implications of this study. Average self-selected velocity across all surfaces, configurations, and cycles was less than typical walking speed for an older adult (20). Indeed, nearly all average velocities were less than the commonly used frailty criteria of 1.0 m/s (21-23). The most favorable configuration, UA, across the easiest surface, Tile, averaged 1.03 m/s which is still lower than a typical walking speed for older adults (24). Thus, even the best performance by this fairly robust cohort falls short of a full facilitation of a typical walking speed. If our population, who were still fully ambulatory and did not require canes, walkers, or wheelchairs as mobility supplements did not select velocities equal to common walking speed criterion, it is improbable that those older adults who require a wheelchair as a mobility supplement would fair any better. Indeed, reports of inability to self-propel have been documented (9). However, a lighter weight chair coupled with a user optimized axle position could serve to mitigate the difficulty imposed by a carpeted or ramped surface, especially in more frail older adults. It may be the difference between a complete inability to self-propel and an ability to navigate a home environment independently. Future studies are warranted.

4.5.2 Implications for the Larger Wheelchair Community

While our population is distinct from the SCI community from who we have learned much, we believe key components are beneficial to the larger community of users. We have demonstrated that a 20lb increase in wheelchair weight produces a significant increase in resultant and tangential force, regardless of what surface a user propels across. A more favorable axle position produced the expected result of increased velocity with a lower force. However, we believe the interactions with surface difficulty provide the greatest insight. Based on our results, clinicians may want to evaluate their clients over difficult surfaces as a method to magnify the impact of changes to a user's wheelchair, be it weight or axle position. In addition, it appears increased weight can mitigate some benefits of a more forward axle position, especially on surfaces like heavy carpet or a ramp. Therefore, we believe a reasonable argument exists that lighter is indeed better and that axle adjustability does not compensate for a heavier chair. Weight and axle position are separate characteristics and should be evaluated as such during the configuration of a wheelchair.

4.5.3 Best Practice Implications

Provision of standard wheelchairs to older adults may represent the current standard of care, but that standard is far from a best practice designed to benefit the consumer. If anything, this practice may magnify the barriers faced by older adults with mobility difficulties. We have demonstrated that a heavy chair with a posterior axle position requires the greatest force and is propelled at a slower velocity when compared to either lighter chairs or chairs with a more anterior axle position. Research in the SCI population has demonstrated the benefits of an optimized axle position (10-15). We have demonstrated the impact of a heavy chair and rear axle position is magnified on carpeted surfaces representative of potential extremes found in the home. In the most optimal configuration on the easiest surface, our older adults used more force to go slower than many people with SCI (16;25;26). In response to the demands of young and active members of the SCI community, wheelchair design and manufacturing have advanced; producing sub 20lb chairs custom fitted to an individual. Who would argue that an older adult with mobility difficulties would not benefit from such advances? And yet they have not. Even as a temporary mobility solution a heavy chair with a rear axle position is a poor choice for an older adult. Older adults rightly view a wheelchair as a disabler or representative of increasing disability if a standard wheelchair represents the most frequently funded option (6;8). Funding policies which fail to facilitate an older adult's access to the lightest chairs with optimized axle positions should be revaluated(5). In light of these findings, powered mobility equipment may represent a more appropriate temporary and long term mobility solution for an older adult.

4.5.4 Limitations

Our primary limitations reside within our sample. Use of older adults who utilized a wheelchair on a regular basis may have produced different results, however for this initial study we wished to ensure our protocol did not overly burden our subjects. Although our subjects were fairly robust, we considered them to be "potential" future users of wheeled mobility, especially given the physical declines which can occur (27). Therefore we believe they represent a reasonable selection. Our choice of 20lbs may seem extreme, but it represents what we feel is the largest reasonable range of weight of a plausible spectrum. The SmartWheels artificially increase the weight of the system; however the test weight of the un-weighted chairs was 40lbs, a reasonable representation of the weight of standard chairs. Even in that configuration, our cohort failed to achieve a reasonable velocity. Furthermore, our primary intent was to test the effect of a weight change, which was unimpaired by the additional weight of the SmartWheels. Standardized instruction on propulsion technique could have improved their performance. However, it is reasonable to infer that the typical older adult who utilizes a wheelchair as a mobility supplement receives minimal propulsion instruction, thus we feel comfortable that we have not created an artificial situation.

4.6 CONCLUSIONS

Increased weight results in decreased velocity and increased resultant and tangential force while a posterior axle position resulted in increased force and decreased stroke length. An anterior axle position does not appear fully mitigate the impact of increased weight. Wheelchair weight and axle position appear to have distinct impacts which are magnified when surface increased. Older adults achieved low velocities in the most advantageous configuration on the easiest of surfaces, and demonstrated a decline in velocity as surface difficulty increased. Heavy wheelchairs with posterior axle positions appear to be a poor mobility choice for older adults if the goal is successful independent mobility.

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5. WHEELCHAIR PROPULSION IN OLDER ADULTS: IMPACT OF GENDER AND STRENGTH

Authors: TDB

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5.1 ABSTRACT

Objective: To determine the impact gender and strength on an older adult's propulsion mechanics

Design: Randomized controlled trial

Setting: Biomechanics laboratory

Participants: 53 community dwelling ambulatory older adults (65+)

Intervention: Participants propelled two wheelchair configurations from a stationary position to a self-selected velocity over three surfaces; hallway tile, low pile carpet, and high pile carpet. Wheelchair configurations were posterior (most rear position on axle plate) and anterior (most forward position +8cm).

Main Outcome Measures: Body-weight normalized hand grip strength, per cycle averages: self-selected velocity, push-frequency, stroke length, body-weight normalized peak resultant and tangential force per stroke.

Results: Stronger older adults (OA) propel faster with higher forces than weaker OA. Main effects were not significant for a group of strength matched men and women. Stronger OA and

men increase push frequency while weaker OA and women decrease push frequency on low pile carpet versus tile

Generally, increasing surface difficulty results in decreased velocity and push frequency and increased forces.

Conclusion: Older men and women with similar levels of strength propel in a different manner Older women self-select a lower velocity than men of similar strength. Gender appears to influence the selection of the user's push frequency, especially as surface difficulty changes. Surface type and axle position affect propulsion mechanics. Larger sample sizes are required to further clarify the impact of strength on propulsion mechanics.

Key Words: Older Adult, Wheelchair, Gender

5.2 INTRODUCTION

For older adults who are experiencing mobility disability, a manual wheelchair often serves as mobility supplement[1]. As such, it is used selectively, likely based on factors both internal and external to the user[2]. Mobility disability is often corollary to increasing frailty and decreases in whole body strength and endurance[3], although general age related declines in strength do not necessarily result in mobility disability. Poor performance on measures of lower extremity strength, such as walking or rising from a chair, are associated with decreases in upper body strength[4]. Indeed, grip strength has been used alone[5] and in combination with additional measures[6] as an index of frailty.

Research addressing environmental or user factors which may affect an older adult's ability to self-propel is sparse, although reports of inability or difficulty with self-propulsion have been documented[7,8]. It is not unreasonable to assume the ability of an older adult to self-propel is influenced by their strength, characteristics of the wheelchair, and expected environment of use. In individuals with spinal cord injury, changes in strength and endurance have been correlated with propulsion performance[9,10]. In addition, individuals with quadriplegia propel slower than those with paraplegia[11], likely in part to lower levels of strength and conditioning[12]. It remains to be determined how the strength of an older adult affects their propulsion mechanics.

Characteristics external to the user which have a known impact on propulsion mechanics include wheelchair configuration and surface of propulsion[13-15]. In individuals with spinal cord injury, it is well known that a more anterior position of the rear wheels reduces the force needed to propel at a given speed[13,16]. In addition, more anterior positions are correlated with increased contact angles and decreases in push frequency, both of which are considered favorable[13,17]. For the older adult an anterior axle position may facilitate self-propulsion by decreasing the amount of strength needed to achieve independent self-propulsion. In addition, an anterior axle position may facilitate a longer stroke through increased access to the pushrim. Propulsion across surfaces with increased resistance above that of a firm surface, such as tile, result in decreased self-selected velocity and increased force[11,14]. The response of stroke length and push frequency to increased surface resistance has not been fully reported[14,18].

More extensive research is warranted to delineate changes in propulsion mechanics of older adults due to alterations in axle position and increased surface difficulty. Documentation of how older adults with different levels of strength self propel across common surfaces in different wheelchair configurations can provide insight upon which to base interventions improving their ability to self propel.

Therefore the purpose of this study is to determine if older adults with different levels of strength demonstrate different propulsion mechanics across a series of surfaces and wheelchair configurations.

5.3 METHODS

5.3.1 Participants

Fifty-three older community dwelling adults were recruited through flyers at local senior citizen centers, interest groups, bring a friend strategies, and IRB approved research registries (Men = 20, Women = 33). All subjects gave written informed consent prior to participation in the study. This research protocol was approved by the Institutional Review Boards (IRB) of the VA Pittsburgh Health Care System and the University of Pittsburgh. To be eligible for participation, subjects had to self-report 1) the ability to walk without human assistance, 2) the ability to stand-up from a chair, 3) the ability to push themselves in a wheelchair, 4) weight as less than 251 lbs, and 5) score greater than 22 on the mini-mental state exam. Exclusion criteria included self-reported history of stroke or a diagnosis of Parkinson's or Alzheimer's. Participant characteristics can be found in Table 5.1. Participants reported minimal experience with wheelchair propulsion.

5.3.2 Clinical Marker of Strength

Handgrip strength was tested using a Janmar[a] hand grip dynamometer. Participants were instructed to squeeze as hard as possible for five seconds using their dominant hand. A second trial was administered after a 2 minute rest. The average of the 2 trials, normalized to body-weight was used for analysis. The cohort was divided into quartile for analysis purposes.

Preliminary analysis indicated substantial differences in the gender composition between and within most quartiles. Overall, men were statistically stronger than women. To examine potential differences due to gender, a sub-analysis comparing men and women of equal body-weight normalized strength was completed.

5.3.3 Initial Wheelchair Adjustment

Three TiLite[b] titanium folding chairs (Model X) were secured for this study. The seat dimensions of the three chairs were 15 x 16, 17 x 17, and 19 x 18 with 0 degrees of seat inclination and camber. Equipped with fabric seat and back upholstery, a 2 inch foam cushion, plastic removable side guards, anti-tippers, pneumatic rear wheels, solid 4 inch casters, and aluminum hand rims, each chair weighed 25lbs. The three seat dimensions allowed us to accommodate a variety of individuals.

Each individual was seated in the smallest possible width wheelchair. When possible, rear axle position was adjusted vertically so each subject demonstrated an elbow flexion angle between 100 and 120 degrees when seated in the chair with their hand placed at top center of the pushrim (mean elbow angle = 107.1^{o} +/-6.3^{o}, full extension = 180^{o}). Footrests were adjusted to provide support for the feet and thighs, as allowed by the design of the chair. Front and rear seat heights were kept as equal as possible to maintain 0 degrees of seat inclination. Front caster angle was adjusted to minimize caster trail.

5.3.4 Wheelchair Test Configurations

Two rear axle positions (anterior and posterior, 8cm difference) were tested. Both configurations were equipped with anti-tippers. The forward axle position was the most anterior rear axle position allowed by the manufacturer's bracket system. The rear axle position was 8 centimeters posterior of the forward position; representing the most posterior axle position allowed by the manufacturer's bracket system. The anterior axle position represents a more optimal rear wheel configuration[13,17]. Participants propelled both configurations in a randomized order. Prior to propelling the 2 configurations on each surface, each participant completed a 6 minute propulsion task serving as a novel functional assessment and an opportunity to practice wheelchair propulsion. Non-standardized instruction describing how to propel a chair was given to all participants.

5.3.5 Propulsion Surfaces and data collection

Three surfaces were selected for the overground course: hallway tile (T, 12.0 m), low pile carpet on a concrete subfloor with no pad (LC, 7.3 m), and a high pile carpet on a typical residential carpet pad on a concrete subfloor (HC, 7.3 m). These surfaces represent surfaces thought to be commonly encountered in home environments. Surface order was randomized for each chair for each participant. SmartWheels[c] provided kinetic bilateral data collection. Currently, the SmartWheel provides the only commercially available method by which to measure forces and moments used to self-propel a wheelchair.

Participants were instructed to begin in a stationary position, hands in their lap, accelerate to a comfortable velocity, continuing until they exited the surface or were instructed to stop. Participants began on the surface; data collection was initiated before initial contact and was terminated before the chair exited the test surface (LC/HC) or the marked data distance (T). Investigators terminated data collection early if it became obvious the participant was experiencing difficulty completing a particular surface in a given chair. Participants completed each surface once in each chair configuration for a total of 6 trials per participant.

5.3.6 Kinetic Data Reduction

High carpet, low carpet, and tile trials were truncated to 6.4m, 6.4m, and 9.0m respectively to remove deceleration that occurred as participants approached the end of the surface or data collection area. The initial 3 strokes are though to compromise the bulk of the initial acceleration from a stationary position and are not included in this analysis[14]. Variables calculated for each cycle from 4 to the end of the data set were averaged together to provide a general representation of propulsion beyond the initial acceleration phase (PIA).

5.3.7 Key Kinetic Variables

The following variables were calculated for each full cycle available in each trimmed trial: average linear velocity, push frequency, stroke length, maximum resultant force, and maximum tangential force. Variables were calculated for right and left side separately, then averaged together. A cycle encompassed the time between the onset of a pushrim contact to the last sample prior to the next contact. A contact was defined as beginning when the resultant force and resultant moment were above threshold (3N and 0.6 Nm). Thresholds were selected to ensure baseline noise was not included in calculations. A contact ended when both dropped below threshold. A search algorithm automated the identification of contact and recovery. Accuracy was verified through visual inspection, with adjustment made on a per cycle basis when necessary. Average velocity was the average linear velocity (m) of the wheel during the cycle. Stroke length was defined as the angular distance (degrees) traveled by the wheel during the propulsive moment portion of a contact. A propulsive moment was defined as Mz above 0.6 Nm. Push Frequency (hz) was calculated as 1 second/cycle time(s). Resultant force (N) was defined as the vector sum of Fx, Fy, Fz [14]. Tangential force (N) was calculated as Mz/pushrim radius (0.2667 m). Resultant and tangential force were normalized to body weight prior to analysis. Matlab[c] was used to trim the data, identify cycles, and compute variables.

5.3.7 Power Analysis

Data describing overground propulsion for older adults does not yet exist in the literature. Therefore, power calculations were based on data derived from Mulroy et al. (52) for two main outcome variables; velocity and cadence. Mulroy et al. reported the self-selected velocity and cadence of four groups of individuals with different levels of spinal cord injury; C6, C7/8, T2-9, and T10-L3, across two surfaces; tile and low pile carpet (52). Two groups, C6 and T2-9, were selected as a means to mimic distinct levels of strength and function, thereby facilitating a tentative power analysis. Given the conservative estimate of mean and standard deviation of these groups for velocity and cadence respectively, the required sample size of each group is 14 to detect differences between two strength groupings of older adults , for a statistical power of 80% with two sided α=0.05.

5.3.9 Statistical analysis

Data was inspected for normalcy. A series of repeated measures analysis of variance (ANOVA) were conducted, one for each outcome measure. The repeated measures included 2 with subject factors and 1 between subject factor as follows: surface (tile, low carpet, high carpet), chair configuration (UA, UP), and Strength Quartile (Bottom 25%, Second 25%, Third 25%, Top 25%). Main effects included strength quartile, with interactions including strength quartile by surface and strength quartile by axle position. Main effects for suface and axle position are reported in Chapter 4. A second set of repeated measures ANOVAs were conducted to examine the influence of gender, with gender replacing strength as the between subject factor. A sub-group of men and women with statistically similar body-weight normalized strength were selected for the analysis. Gender was the main effect tested with interactions including gender by axle position and gender by surface. Significance was set *a priori* at $p \leq 0.05$. A Bonferonni correction was applied to the multiple main effects comparisons to control for Type I errors.

5.4 RESULTS

5.4.1 Participants

A total of 53 older adults participated (Men = 20, Women = 33). Average age was 73.6 years (+/-5.4), ranging from 65 to 87. Participants were generally overweight (BMI = 27.6, +/-5.1, height = 1.7m +/-0.1, weight = 76.5kg, +/-16.7). Mean elbow angle after initial fitting was 107.1^{o} +/-6.3^{o} (full extension = 180^{o}). We were unable to achieve an elbow angle within the specified range of 100^{o} – 120^{o} with one subject due to their anthropometrics (elbow angle = 125^{o}).

5.4.2 Grip Strength

Main effects were significant for average velocity, peak resultant force, and peak tangential force (Table 5.1, p = 0.006, p=0.002, p=0.001). Push frequency and stroke length were not significant for main effects (Table 5.1, p=0.376, 0.099). The quartiles were not statistically different for age, height, or weight (Table 5.1). However, the bottom, second, and top quartiles had a statistically unequal distribution of men and women (Table 5.1, Binomial test).

Overall, older adults in the top body-weight normalized grip strength quartile pushed faster, using a higher peak resultant and tangential force at a similar push frequency and stroke length than older adults in the bottom quartile (p=0.004, p=0.001, Table 5.1). Differences in force could be due to differences in velocity rather than strength.

Table 5.1 Participant and Propulsion variables, Mean (Standard Deviation), for each body-weight normalized grip strength quartile. * Bottom 25% and Top 25% significantly different, p≤0.05. † Second 25% and Top 25% significantly different, p≤0.05. £ All pairwise comparisons significantly different, p≤0.01. ‡ Statistically unequal gender distribution within the quartile

	Bottom 25%	Second 25%	Third 25%	Top 25%
N	13	14	13	13
Age (yrs)	73.23 (5.37)	73.57 (6.57)	74.62 (5.75)	73.15 (4.26)
Height (m)	1.65 (0.13)	1.64 (0.13)	1.74 (0.16)	1.74 (0.11)
Weight (kg)	82.35 (18.68)	71.16 (15.56)	75.86 (19.77)	77.16 (11.54)
Strength (% body-weight)£	27.00 (2.35)	33.79 (2.00)	43.38 (2.67)	55.72 (8.08)
Gender	‡ men = 0 women =13	‡ men = 2 women = 12	men = 7 women = 6	‡ men = 11 women = 2
Velocity (m/s)*	0.70 (0.37)	0.78 (0.31)	0.82 (0.31)	0.93 (0.30)
Push Frequency (hz)	1.07 (0.29)	1.20 (0.34)	1.13 (0.29)	1.05 (0.23)
Stroke Length (degrees)	52.46 (11.86)	50.85 (14.02)	53.93 (11.00)	61.70 (12.68)
Resultant Force (% body-weight)*	9.66 (2.61)	11.02 (3.48)	11.66 (3.37)	12.80 (3.64)
Tangential Force (% body-weight)*,†	7.47 (2.16)	8.17 (2.72)	8.77 (2.88)	9.75 (2.68)

5.4.3 Gender

Gender had no significant main effects (Table 5.2, velocity, p = 0.110; resultant force, p = 0.43; tangential force, p =0.162; stroke length, p=0.593; push frequency, 0.846). The group of men and women selected for this analysis were statistically non-distinct in age and body-weight normalized strength (Table 5.2, p = 0.581, 0.340). Men were taller and weighed more than women (Table 5.2, p=0.028, p=0.002).

Table 5.2 Participant and Propulsion Variables for each gender, Mean (Standard Deviation). * Men and Women statistically different, p≤0.05

N	Men 12	Women 13
Age (yrs)	74.25 (5.35)	75.69 (7.31)
Height (m)*	1.78 (0.10)	1.65 (0.17)
Weight (kg)*	85.05 (14.31)	65.09 (14.88)
Strength (% Body-weight)	43.25% (4.39)	41.13% (6.25)
Velocity (m/s)	0.88 (0.31)	0.77 (0.31)
Push Frequency (hz)	1.13 (0.21)	1.11 (0.33)
Stroke Length (degrees)	54.73 (10.05)	54.03 (13.72)
Resultant Force (%Body-Weight)	12.23 (3.36)	11.35 (3.61)
Tangential Force (%Body-Weight)	9.27 (2.78)	8.25 (2.75)

5.4.4 Grip Strength Interactions

Body-weight normalized grip strength by surface interactions were significant for push frequency and tangential force (Figure 5.1, $p=0.001$, $p=0.036$). All other variables were not significant for grip strength by surface interactions (velocity, $p= 0.446$; stroke length, $p = 0.370$; resultant force, $p =0.080$). Changes in axle position had no significant interaction with grip strength ($p\geq0.246$ for all). This implies on average individuals responded in a similar manner to changes in axle position.

Push frequency changed in a different manner for the top 2 and bottom 2 quartiles as surface difficulty increased (Figure 5.1). On average, individuals in the bottom 2 quartiles responded to progressive increases in surface difficulty by progressively decreasing their push frequency. In contrast, the response of individuals in the top 2 quartiles varied based on the degree of increased difficulty. A small relative increase, such as low carpet versus tile, resulted in an increased push frequency. A large relative increase, such as the difference between tile or low carpet and high carpet, resulted in a decrease in push frequency. In addition, on lower resistance surfaces, the bottom 2 quartiles generally selected a higher push frequency than the top 2 quartiles (Table 5.2, section B). Differences in tangential force mirrored differences in velocity, reaching significance on the low carpet and high carpet.

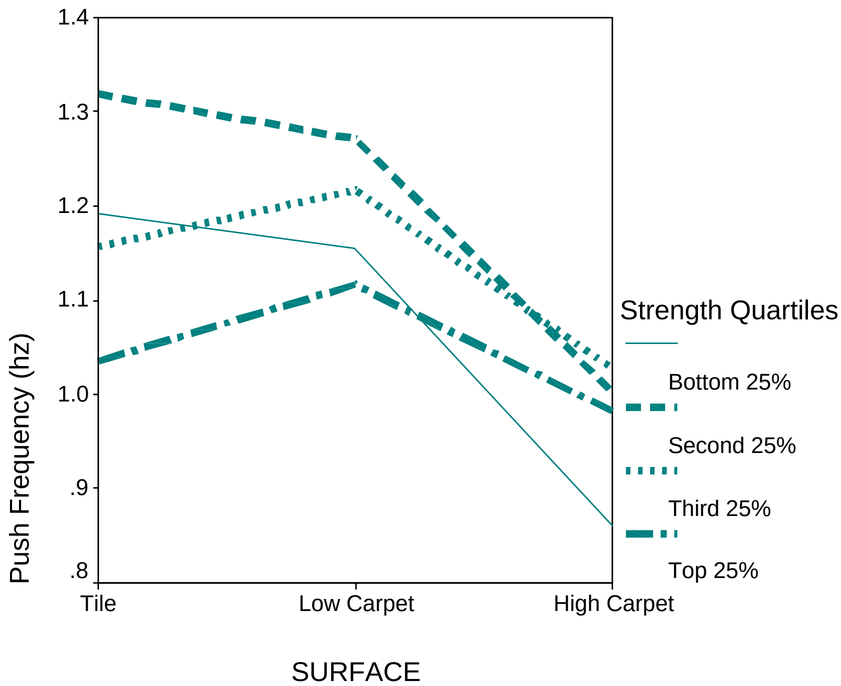

Figure 5.1 Surface by Strength Group Interaction for Push Frequency. Mean push frequency of each strength quartile for self-selected velocity on tile, low pile carpet, and high pile carpet.

5.4.5 Gender Interactions

A surface by gender interaction was significant for push frequency (Figure 5.2, p = 0.004). All other variables were not significant for grip strength by surface interactions (velocity, p= 0.658; stroke length, p = 0.321; resultant force, p =0.672, tangential force, p=0.789). There was no significant gender by axle position interactions (p≥0.169 for all).

Men and women of similar strength respond differently to changes in surface difficulty.

On average, women responded to progressive increases in surface difficulty by progressively decreasing their push frequency. In contrast, the response of men varied based on the degree of increased difficulty. A small relative increase, such as low carpet versus tile, resulted in an increased push frequency. A large relative increase, such as the difference between tile or low carpet and high carpet, resulted in a decrease in push frequency. In addition, on lower resistance surfaces, women generally selected a higher push frequency than men.

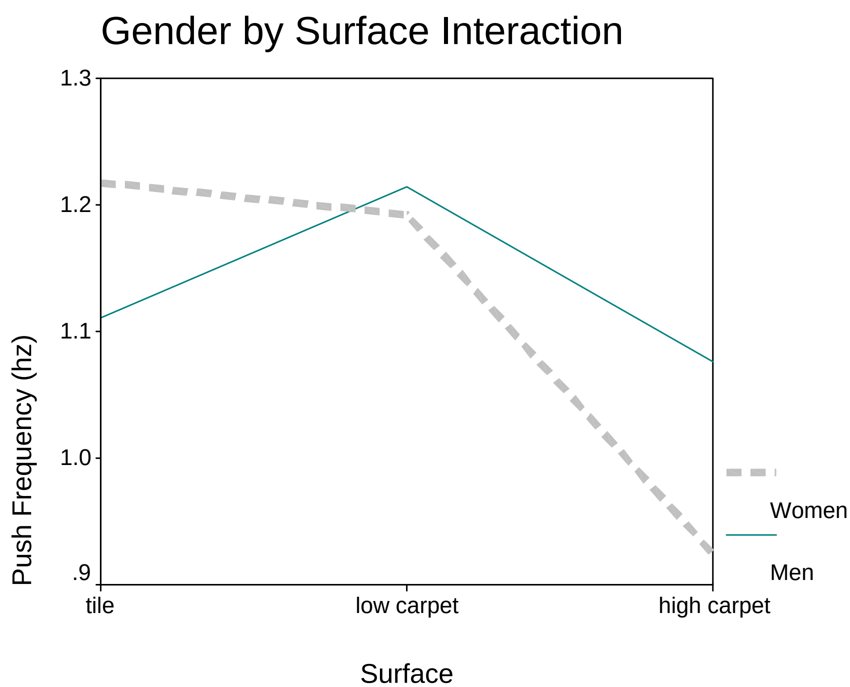

Figure 5.2 Gender by Surface Interaction for Push Frequency. Mean push frequency of men and women for self-selected velocity on tile, low pile carpet, and high pile carpet.

5.5 DISCUSSION

Ambulatory older adults at the lower end of the strength continuum self-select a lower propulsion velocity than their stronger peers. Lower peak forces applied at the same push frequency and stroke length was characteristic of weaker older adults. However, the differences in force may be due to lower velocity rather than lower strength. There were no significant differences in propulsion mechanics between men in women of similar strength. Closer examination of the interactions between the grip strength quartiles, gender, and surface difficulty for the key parameters of velocity, push frequency, and tangential force provides preliminary insight on the potential role of strength and gender in propulsion of novice users.

Comparisons within and across strength quartiles highlights how responses to changes in surface difficulty may be specific to the strength of the user. When the surface of propulsion was considered in conjunction with strength, differences in push frequency manifest. Older adults in the bottom 2 strength quartile responded to increasing surface difficulty with a progressive decrease in push frequency. Given the typical positive relationship between propulsion velocity and push frequency, this change is not surprising. However, these same adults demonstrated a higher push frequency on tile than their stronger peers, who self-selected a higher velocity. In addition, while the top 2 quartiles cohort also slowed down on low pile carpet, they increased their push frequency, in contrast with the expected decrease reflected in their weaker peers. On high pile carpet, the expected relationship between push frequency and velocity held true, with higher velocities coupled with a higher push frequency. Together, these results suggest that the strength of a manual wheelchair user may affect their selection of a push frequency. In addition it appears that decreases in velocity due to increased surface resistance are not necessarily accompanied by a decrease in push frequency.

However, given the differences in the gender composition of the quartiles, the role of gender must be explored. Without regard to surface type or axle position, there were no significant differences between men and women of similar strength in propulsion mechanics. When interactions were examined, this strength matched group of men and women responded differently to changes in surface difficulty. The push frequency response of men mirrored the response of the top 2 strength quartiles, the women's push frequency response mirrored that of the bottom 2 quartiles. Comparison of Figures 5.1 and 5.2 reveal the strikingly similar patterns. Gender based differences beyond strength appear to affect propulsion mechanics.

Normalizing grip strength by body-weight assumes the amount of force required to self-propel at a given velocity is linearly proportional to the user's weight. However, if this is not true, the apparent gender based differences in push frequency my indeed be due to differences in strength. If the relationship between force and weight at a given velocity is non-linear in nature, then absolute strength of the individual would be a more appropriate measure. For example, if the force/weight relationship plateaus at a given weight or velocity, then stronger individuals such as our men will have a distinct advantage over weaker individuals, such as our women. Our men were heavier than our strength matched women and consequently were stronger if absolute strength is considered. Future studies should clarify the relationship between the force required to self-propel, a user's weight, and velocity. Defining the nature of this relationship will facilitate study designs determining if gender based propulsion differences do indeed exist and allow development of targeted interventions to improve propulsion in these groups.

5.5.1 Limitations

We were successfully able to detect a 0.2 m/s difference in self-selected velocity between the top and bottom strength quartiles with 13 individuals in each group. We were unable to detect a difference in stroke frequency between these groups, with a difference of 0.02 Hz between groups. The strength of a user may not affect their selection of a stroke frequency, indicating it may not be an appropriate choice to discriminate between groups. Our findings provide data for power calculations to design studies more fully exploring the impact of gender strength on propulsion mechanics.

Providing standardized training prior to data collection could have both improved the biomechanics of our cohort and reduced the wheelies noted previously. However, we found little evidence that older adults receive propulsion training as a part of their service delivery. Thus we believe our decision to not provide instruction is a realistic reflection of the current standard of practice. Additionally, the anterior configuration represented a configuration that while clinically available, is likely far more unstable than that which an older adult receives or would expect to utilize. Use of older adults who currently used a manual wheelchair may have altered our findings. For the purposes of this initial study we did not wish to overly tax our participants and if frailty is common to current users, then our protocol could have imposed to great a subject burden to be ethical. In addition, our bottom quartile is likely stronger than their peers who use a

MWU due to frailty, thus their mechanics may represent a best case scenario for weak older adults.

5.6 CONCLUSIONS

Older adults with lower body-weight normalized strength select a slower velocity than their stronger peers. Interactions between a user's strength, gender, and surface type were noted. Gender based differences appear to influence the selection of the user's push frequency as surface difficulty changes. Larger sample sizes including men and women of various strengths are required to further clarify the impact of strength and gender on propulsion mechanics.

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6. DISCUSSION

Multiple avenues for intervention to facilitate improved wheeled mobility are available to the clinician. These interventions can target the user, wheelchair, or environment. We have highlighted the differences between two distinct groups, differences due to wheelchair weight and axle position, and explored the role of strength in propulsion mechanics. However, to date a method has not existed to objectively identify both an opportunity for intervention and to measure the change imposed by said intervention. In an attempt to address this gap, we have developed a clinical decision framework. This framework guides clinicians through the identification of opportunities for intervention as well as providing population based reference values. This process represents a preliminary attempt to improve the objective documentation increasingly required by third party payers. Further validation is warranted.

Older adults self-select a low propulsion velocity, even when fitted in a very lightweight manual wheelchair with a more optimal axle position. As anticipated, this chair configuration reduced the amount of force required to propel at a given velocity. Decreasing the amount of force needed to self-propel may facilitate independent mobility in this cohort, albeit at low velocity. Increasing surface difficulty resulted in further reductions in velocity and increased forces. Overall these results bring into question the practice of providing any manual wheelchair, much less anything less than the most optimal configuration, to an older adult if the goal is successful, independent mobility.

When compared to an experienced group of individuals with paraplegia, our novice older adults propelled slower using similar peak resultant force, lower tangential force, a shorter stroke length, higher push frequency and a greater minimum Mz. Individuals with paraplegia represent a group of experienced, “successful” propellers who generally receive more optimally configured wheelchairs(1). When older adults and individuals with paraplegia are confronted with a carpeted surface, both groups decrease velocity and push frequency, and increase peak force. Unexpectedly, older adults maintained the amount of total work per cycle while individuals with paraplegia increased work per cycle. Non-similar responses to changes in surface difficulty could be due intrinsic differences in the cohort, such as strength or experience. Work per cycle may be sensitive to fundamental differences in users or surfaces and should be further explored.

Preliminary results indicate gender may play a role in the propulsion mechanics of novice users. Differences in propulsion mechanics existed between the upper and lower strength quartiles of the older adult cohort. However, these quartiles were dominated by men and women respectively. Examination of strength matched men and women indicated the differences between the quartiles were likely gender based. Men propelled faster, using a longer stroke length, lower push frequency, and higher forces. The higher forces and longer stroke length could be attributed to the higher velocity(2). However, the higher push frequency demonstrated by the women remains to be explained. In addition, men and women responded differently to changes in surface type. Both groups slowed down, but men increased their push frequency for the low pile carpet, while the women decreased push frequency. Together, these results suggest novice older adults may select different propulsion strategies as a function of their gender.

6.1 DEFINING A "SUCCESSFUL" SELF-PROPELLER

Before the success of an intervention can be measured, "success" must first be defined. For our purposes the propulsion mechanics of a group of individuals with paraplegia represented a "successful" self-propeller. As noted above and reported in Chapter 3, this group of individuals was substantially different from our cohort of older adults. Individuals with paraplegia self-selected a higher propulsion velocity using a lower push frequency and longer stroke length than older adults. In addition, they used similar levels of resultant force, but greater peak Mz. Older adults applied larger breaking moments. Differences in propulsion in these groups could be due to differences in experience. Research has documented that discrete differences exist between experienced and novice users(3). The required dose and duration of practice to achieve propulsion mechanics characteristic of experienced users has not been documented. However research does suggest users adopt the most optimal technique quickly, perhaps in the first few minutes of practice(4).

A second, more appropriate criteria defining "successful" mobility is the ability to self-propel at the typical walking pace of your peer group. A wheelchair should *restore* mobility to a *functional* level. Older adults who rely on wheeled mobility should be able to keep pace with their peers in a community setting. If this goal cannot be met then powered mobility options should be explored. By either criterion, many of our older adults failed to achieve a "successful" level of mobility. However, the lower forces imposed by the un-weighted chair and anterior axle position represent more optimal propulsion and should not be discounted.

6.2 POTENTIAL FOR "SUCCESSFUL" PROPULSION IN THE OLDER ADULT

Certainly a portion of older adults who require a manual wheelchair might posses the physical capacity to become successful self-propellers. It would not be unreasonable to expect an older adult who was experiencing a temporary orthopedic impairment, but otherwise was healthy and strong, to be able to self-propel, keeping pace with their ambulatory peers. However, given the age related declines in strength and conditioning, often further reduced in those experiencing mobility disability, is successful self-propulsion a realistic goal? We believe our results indicate full facilitation of a typical walking velocity through manual wheeled mobility is not a realistic goal for many older adults. Our cohort appeared fairly robust, as they were able to travel from their home to our lab for testing, and all walked without canes or walkers. Our inclusion criteria assured us of a fairly healthy group, as the ability to stand up from a chair without human assistance was required. If this group consistently self-selected a low propulsion velocity, it would be unreasonable to expect a frailer cohort to perform better.

Although it is unrealistic to expect a frail older adult to self-propel at speeds comparable to their healthy peers, we believe this group would benefit from better fitting wheelchairs. Previous research indicates older adults who received customized manual wheeled mobility report increased use of and greater satisfaction with their chairs(5;6). Our research provides additional support by demonstrating decreased wheelchair weight and an anterior axle position independently reduce the force needed to self-propel.

6.3 WHEELCHAIR CONFIGURATION AS AN INTERVENTION TO IMPROVE MOBILITY

For many clinicians the wheelchair represents the initial point of intervention to improve propulsion in all users. On chairs with an adjustable axle position, multiple positions can be tried during a single evaluation. Although the clinician and user conceivably perceive the impact of each change, clinically relevant research defining the impact of changes in axle position have been lacking. Clarifying the immediate impact of axle position will allow clinicians to focus on variables known to be responsive to change. In contrast, to axle position wheelchair weight generally can only be adjusted through acquisition of a different chair. However, when a new chair is secured, multiple alterations in addition to weight often exist. These differences can include configuration, design, or manufacturing quality. Therefore, determining the specific impact of weight changes is challenging. Delineation of the effect of weight on propulsion mechanics can provide justification for the provision of very light weight chairs to older adults.

6.3.1 Axle Position

Of the options which can be customized on a manual wheelchair, axle position represents an obvious method by which to alter the propulsion mechanics of an individual. Vertical axle position(7;8) has a known impact on propulsion mechanics and metabolic demand of propulsion, but was not a focus of this study. To minimize the confounding impact of seat height, we adjusted seat height for each individual. Seat height was adjusted such that when seated with their hands placed at to dead center of the push rim elbow flexion was 100^o – 120^o (7). Two clinically available horizontal axle positions spaced 8cm apart were selected for evaluation.

These position were the most anterior and posterior positions available on the manufacturers bracket.

Generally, the anterior position resulted in decreased forces and increased stroke length across all surfaces while velocity remained constant. These findings confirm and contrast results of Kotajarvi et al, who examined 9 different axle positions in users and non-users(9). Consistent with the results of Kotajarvi et al was our documented increase in stroke length in an anterior versus posterior axle position. Kotajarvi et al did not find differences in tangential forces between horizontal axle positions. In contrast, we found an anterior axle position resulted in lower tangential forces across multiple surfaces. Differences between the studies could explain the discrepancies. Kotajarvi et al examined average forces, we selected peak forces for our analysis. Although peak and average forces are correlated, peak forces may be more robust to immediate changes in axle position. Our study samples differed as well. We were fortunate to recruit a fairly large sample (N=53) of inexperienced users. Kotajarvi had a smaller sample (N=13) of experienced users which limited their statistical power. However, experienced users may respond differently to changes in axle position. In addition, potential differences due to changes in horizontal axle position may have been skewed by lack of an optimized seat height in Kotajarvi et al. As noted above, we fitted each participant such that seat height fit as recommended, decreasing external confounders(7). Kotajarvi et al was the first study to evaluate the impact of axle position during over ground propulsion and to date is the only such study to include kinetic and kinematic data. Valuable insight has been gained from their efforts.

Additional researchers have examined the impact of axle position in novice and experienced users(7;8;10). The compiled knowledge gain from these studies provided valuable insight during the development of the current project.

Although considerable effort was expended during the design of this project, we neglected to include standardized propulsion instruction. We believe older adults who receive manual wheelchairs do not receive propulsion training. However documentation supporting or negating this belief does not exist. Some participants popped a series of wheelies when seated in the anterior axle configuration. As this was unanticipated and represented a safety hazard, we provided these participants with limited guidance on how to prevent such wheelies. Such instruction may have affected the results of our study. We expected the un-weighted, anterior axle configuration to result in a higher self-selected velocity due to the lower forces required for propulsion. However, the un-weighted posterior axle position was propelled the fastest on one surface, tile. For all other surfaces, the configurations were statistically similar in velocity. The instability of the anterior axle configuration may have resulted in users selecting a lower velocity as a strategy to prevent wheelies. As we did not document the number of wheelies in each configuration, we are unable to statistically document the impact of such events on propulsion. We recommend future studies incorporate either kinematic or video data collection to document the occurrence of wheelies. Provision of standardized instruction should also serve to reduce chair control errors and the impact of learning effects.

While axle position represents one of the most thoroughly studied aspects of wheelchair propulsion, much work remains. Development of an objective clinical method for determining an "optimal" axle position for each user would be of great benefit, as would an objective definition of "optimal". Until such a method is defined, an iterative approach balancing stability and ease of propulsion is effective in establishing an optimal position. The optimal axle position for each user is unique, but should be as far anterior as possible without compromising the user's rearward stability. The goal is to minimize forces required for propulsion and popping wheelies while maintaining the user's ability to control the wheelchair.

In addition, the impact of axle position has only been studied in straight line propulsion. Future studies should quantify the benefits of horizontal and vertical axle position on the forces, moments, and metabolic demand of maneuvering a manual wheelchair. The benefits of an optimized axle position should be explored in additional populations, such as individuals with spina bifida or those compromised by a stroke. All studies should strive for large sample sizes, as the immediate impact of axle position alterations appears to be small. Long term interventions in multiple populations evaluating the extended benefit of an optimized configuration may provide the most useful information for clinical applications.

6.3.2 Wheelchair Weight

Lighter weight wheelchairs are often provided to individuals with spinal cord injury as a strategy to help prevent the onset of upper extremity pain and dysfunction. However, research addressing the specific impact of wheelchair weight on propulsion mechanics does not exist. Although wheelchairs are classified in part by their weight, as weight decreases, quality of manufacturing, materials, and design tend to improve, introducing a quandary. Is it the weight of the chair the makes the difference or is it the quality of the overall design, manufacturing, and materials?

Comparisons of wheelchairs with substantial differences in design, weight, and materials have suffered from this dilemma(11;12). In an attempt to circumvent these confounding issues, we added 20lbs of weight to a high-end very light weight chair. Our base chair weighed 25lbs, complete with footrests, cushions, wheels, anti-tippers, side guards, and armrests. The weight was added in a manner that reflected the natural weight distribution of the chair to minimize the impact of changes in weight distribution. However, each SmartWheel (SW) weighs ~11lbs, increasing the weight of our test configurations to 40lbs and 60lbs. Additionally, it appears the tire on the SW has a greater influence than the weight of the SW(13). We used commercially available solid treaded tires for our study, similar to those found in the community. The SW undoubtedly alters the propulsion of the user, but remains the only commercially available method to assess forces and moments of propulsion. Our results should be interpreted with this in mind. Nonetheless, the added weight of the SW does not negate our ability to asses the impact of a 20lb change in weight.

We have demonstrated a 20lb change in wheelchair weight immediately affects velocity and force in novice older adults. In a lighter chair, older adults select a higher velocity using lower forces. However, there were no immediate changes in stroke length and push frequency. Changes in weight in absence of changes in axle position would not affect the geometry of the available contact angle, thus lack of change in stroke length and push frequency are not surprising. An argument may be made however that an experienced user may have responded differently. As an example, an experienced user may have responded to a lighter chair by maintaining force and stroke length, but decreasing push frequency, and increasing velocity. Future research should explore how more experienced manual wheelchair users respond to changes in wheelchair weight.

The impact of wheelchair weight has been assumed to be negligible in contrast with the weight of the user (14). Our older adults weighed on average 168.4 lbs. Without the SmartWheels attached, our two configurations were 25lbs and 45lbs, resulting in total system weights of 193.4lbs and 213.4lbs. For each configuration, the chair contributed 12.9% and 21.8% respectively of the total system weight. Such relative increases in weight are not inconsequential. Our results indicate that on well manufactured wheelchairs, reductions in weight can improve propulsion.

In addition, the impact of wheelchair weight was magnified by increasing surface difficulty. As surface difficulty increased, the gap between the light chairs and heavier chairs increased. Heavier chairs were propelled progressively slower with greater forces. Clinicians should take care when evaluating clients within the clinical setting. A chair which performs adequately on tile or low pile carpet may not perform as well at home on more plush carpeting. If a clinician must justify why a poorly configured chair will not suffice for their client, propulsion over surfaces found in the client's home may provide the needed evidence.

Future studies should evaluate the impact of small incremental weight changes and focus on experienced users. Experienced users are likely in tune with a wheelchair and may respond very differently to changes in wheelchair weight. We selected 20lbs in part because it was a clinically relevant difference in weight, but of a magnitude we felt we could detect with a reasonable sample size. Additional insight would be gained by exploring the metabolic demand associated with increased wheelchair weight. In theory, increased forces associated with increased weight should manifest as increased oxygen consumption if propulsion velocity is held constant. Finally, future studies should include several fixed velocities in addition to self-select as a test conditions. The magnitude of difference between a heavy and light chair may vary according to velocity.

6.4 INITIAL INSIGHT INTO THE IMPACT OF SURFACE TYPE

Of the two factors assessed, surface type and wheelchair configuration, surface type had the largest impact on propulsion mechanics. Typically, as surface resistance increased, velocity and push frequency decreased and forces increased. Stroke length was generally unchanged across surface type. Our results confirm the findings of Newsam et al, who reported a decrease in self-selected velocity and push frequency on low pile carpet versus tile in a group of experienced users(15). Our individuals with paraplegia demonstrated similar changes when confronted with low pile carpet. We have expanded in part the findings of Koontz et al who examined start-up propulsion over a variety of surfaces in experienced users(16). As expected, surfaces with increased resistance required greater forces to transverse at a lower velocity (16). To date, Koontz et al represents the most extensive examination of the impact of surface type on propulsion mechanics. Use of novice older adults highlights the universal impact of surface difficulty on propulsion mechanics, namely decreased velocity and push frequency coupled with increased forces. As an environmental condition, surface type is relatively fixed in the daily life of a user. Therefore, fully exploring its role in propulsion mechanics can assist in the selection and configuration of the most optimal wheelchair for an individual.

What remains undefined is the impact of user characteristics on their response to increased surface difficulty. As noted earlier, our cohort of older adults responded to an increase in surface difficulty by maintaining their work per cycle while individuals with paraplegia increased their work per cycle. An obvious difference between the groups is their level of experience. Individuals with more propulsion experience may select a different strategy when confronted with increased surface difficulty. In addition, the gender distribution of the cohorts may have magnified the differences between these groups. Within our older adult cohort, men responded to increased surface difficulty in a manner distinct from women. Men responded to increased surface difficulty with an increase in push frequency contrasted with the decrease exhibited by women. Our cohort of individuals with paraplegia was dominated by men, while our older adult cohort was predominately women. Gender based disparities may explain a portion of the differences between our comparison groups.

To date, propulsion research evaluating the impact of surface type has been limited(15;16). In addition, the majority of research has been focused on steady-state propulsion(8;17) (18). Wheelchair users engage in start up propulsion over 100 times per day(19). Results from chapter 2 indicate forces and moments associated with start-up exceed those found during steady-state propulsion, especially on carpeted surfaces. Future research should focus on the initial few strokes from a stationary position on various surfaces. The daily life of a wheelchair user surely encompasses more than straight-line propulsion on a firm level surface. Maneuvering in the home on carpet, transversing cross-slopes, and up curb cuts define in part the daily experiences of many wheelchair users. Clarifying the biomechanical and metabolic demands of these situations is the next propulsion research challenge.

6.5 PRELIMINARY INDICATIONS FOR THE ROLE OF USER CENTERED INTERVENTIONS

It appears changes in horizontal axle position and wheelchair weight immediately affects self-selected velocity, peak force and stroke length in novice users. Push frequency does not seem to be immediately altered. Lack of immediate change in push frequency has been documented previously in an experienced group of users(9). Kotajarvi et al and our results indicate changes in stroke length due to purely horizontal axle position are small, averaging less than ten degrees. It seems that stroke length and push frequency may be best modified through user centered interventions. Propulsion training could serve to magnify the impact of an optimized axle position. Providing individuals with instruction on more optimal propulsion technique is an obvious starting point which could result in immediate and measurable change. Propulsion training has been shown to increase stroke angle and decrease push frequency in experienced users(20). Preliminary results from our laboratory corroborate these findings and indicate force profiles can be altered by propulsion training(21). However, long term effectiveness of propulsion training has yet to be established.

Strength training may represent another method by which to alter propulsion mechanics(22). Other studies have demonstrated differences between levels of SCI(15) and suggest strength differences may play a role. After a combined intervention including strengthening, individuals propelled using a lower push frequency at a fixed power output (22). In theory, stronger individuals should be able to maintain a given velocity using a lower push frequency because they have the capacity to apply greater forces at each stroke. In addition strength training might allow users to maintain a velocity of their choice on more difficult surfaces at a lower relative physical strain(23;24). Strength training combined with propulsion training may hold potential for the greatest benefit. Strength training would provide the physical ability to execute behavioral changes learned through propulsion training. The benefits of both these interventions would be maximized in an optimal chair configuration and muted in a poorly fitted and configured chair. It must be noted however, that these interventions may not be practical for all older adults. Clinicians should use their professional judgment when tailoring an intervention for their client.

Surprisingly, gender may affect the propulsion mechanics in novice older adults. Our results indicate women propel slower than men using a higher push frequency on hallway tile. When confronted with low pile carpet, both groups slow down, but men increase their push frequency while women decrease. Further exploration of the benefits of these strategies and their cause could assist in the identification of gender specific interventions. However, our decision to weight normalize grip strength assumes the amount of strength required to self-propel is proportional to the user's weight. If this is not true, the apparent gender based differences in push frequency my indeed be due to differences in strength. Men and women of equal weight normalized strength are only equal in absolute strength if their body-weight is also equal. Our men were heavier than our strength matched women and consequently were stronger if absolute strength is considered. Future studies should compare the propulsion of men and women of similar height, weight, and strength, across a spectrum of strength and body-dimensions to determine if gender differences truly do exist.

6.6 IDENTIFYING A NEED FOR AN INTERVENTION

Chapter 2 defines a method by which clinicians can identify a need for an intervention. This process was developed after the 2006 SmartWheel User's Group Meeting where it became apparent that achieving a functional velocity was the primary goal of clinicians involved in seating and mobility interventions. The clinical decision framework is based on 3 ranked questions (Chapter 2 - Figure 2.7). "Is the velocity acceptable?" "Is the force acceptable" "Is the push frequency acceptable?" Clinicians need not progress to the next question until a yes answer has been achieved for the current question. This evaluation highlights the importance of first evaluating velocity as an index of function. Establishing an "acceptable" velocity is incumbent upon the user and clinician. However, we suggest 1.06 m/s as a minimum level. This velocity represents the average minimum needed to safely cross an intersection (25).

As noted above, more appropriate velocity criteria may be the typical walking velocity of the client's peer group. For our older adults this is 1.27 – 1.33 m/s; for our individuals with paraplegia it would be 1.39 - 1.46 m/s(26). Many of our older adults failed to reach this threshold, even when fitted in a more optimal configuration. Our SW increased the weight of each chair by 15lbs, which likely slowed our group, especially on the heavy carpet. We believe many would have still selected a velocity below this threshold if the regular rear wheels were used in place of the SW. However, if we follow our clinical decision framework, once the most optimal configuration failed to elicit an acceptable velocity, we would explore powered mobility options. Justification of a powered mobility when an optimal manual configuration does not suffice is consistent with the criteria for powered mobility adaptive equipment set forth by the Centers for Medicare and Medicaid Services(27). It must be noted that powered mobility can only facilitate mobility if the environment of use is accessible. Powered mobility solutions are generally not as portable as manual wheelchairs, although select models can be broken down for transport. Lack of accessible transportation and/or an accessible home environment may be barriers to successful use of powered mobility. In such situations, a very lightweight manual wheelchair with an optimized configuration may represent the best possible solution.

Conversely, not all older adults self-propel or desire to self-propel. A common sight in the community is an older adult in a wheelchair being pushed by a companion. If an older adult truly has no desire to self-propel, a very light weight chair may still facilitate mobility. As stated before, these chairs are characterized by a higher quality of design, materials, and manufacturing, which would make being pushed easier. The lower weight of these chairs would decrease the strain experienced by their companions when the chair is loaded in and out of a car, or transported up and down steps. A successful intervention accommodates the needs, goals, and desires of the client. Allowing the user to define their goals will allow for the development of a successful solution. Compared to individuals with SCI independent self-propulsion may not be a priority for older adults. Researchers and clinicians should remain mindful of the goals of their subjects and clients when designing and evaluating the success of an intervention. An idealized wheelchair configuration, be it manual or power, can only facilitate independent mobility if the user desires to be independent.

For individuals with SCI, the importance of weight loss and maintenance interventions should not be overlooked. Although our individuals with paraplegia were younger than our older adults, they were significantly heavier. Generally, weight increases with age. Obesity is more common in individuals with paraplegia due to a greater sedentary lifestyle and decreased lean tissue mass. The greater weight of our individuals with paraplegia is indicative of poor weight management and increases the load on the upper extremity during propulsion and transfers. This cumulative loading likely increases their risk for upper extremity dysfunction, ultimately compromising their independence. Weight management and reduction strategies should be considered by clinicians as a compliment to proper chair prescription to reduce the risk of developing upper extremity pain and injury in individuals with SCI.

Examination of Figures 2.2, 2.4, and 2.5 in Chapter 2 reveals a substantial portion of our cohort with Spinal Cord Injury or Dysfunction self-select a propulsion velocity below our self-imposed threshold of 1.06 m/s. This may be due in part to the impact of the SW and the nature of our data analysis. The additional resistance provided by the SW above and beyond a user's natural wheel and tire configuration could have resulted in a self-selected velocity lower than the user's typical pace. In addition, we limited our analysis to the fourth and fifth stroke collected, at which point the user may not have fully accelerated to their typical speed, especially given the increased resistance of the SW. Reference values for typical propulsion velocities should be established with the SW attached, allowing for the impact of the SW on self-selected velocity to be defined. However, if the low velocities exhibited by a portion of our group are indeed representative of their typical velocity, they may not be as functional as they could be and powered mobility options may be warranted.

6.7 SUGGESTIONS FOR A CLINICALLY MEANINGFUL DIFFERENCE

Regardless of the absolute amount of change resulting from an intervention, clinical relevance is of primary importance. Statically significant changes may not have clinical relevance and vice versa. Unfortunately, clinically meaningful changes have yet to be established for all propulsion parameters. However, research from other fields indicates a change in a self-selected walking velocity of 0.1 m/s represents a measurable improvement in function(28;29). In addition, small differences between certain critical reference values further indicates small absolute changes in velocity are clinically meaningful. As mentioned in chapter 2, small differences exist in self-selected velocity between SCI levels on a given surface or condition(11;30;31). Self-selected velocity differed between tetraplegia (0.8 ms) and paraplegia (1.2 ms) in Beekman et al. by 0.4 ms (11); a difference of 50%. The absolute and relative difference between a preferred walking velocity (1.22 m/s) and the minimum needed to safely cross an intersection (1.06 m/s) is even smaller; 0.16 m/s (25;32); a 15% difference.

Although changes in wheelchair configuration did not elicit an appreciable increase in self-selected propulsion velocity, the reductions in velocity which occurred as a result of increasing surface difficulty could be considered functionally important. Removal of environmental barriers to independent propulsion may be as critical as proper wheelchair configuration to independent mobility. Provision of firm, level surfaces in public facilities would benefit all users of manual wheeled mobility.

6.8 FEDERAL POLICY IMPLICATIONS

Our results indicate alterations in two areas of federal policy could help facilitate improved wheeled mobility for all manual wheelchair users. Current Centers for Medicare and Medicaid Services (CMS) policy facilitates the provision of heavy, non-adjustable wheelchairs to older adults. Our results highlight the impact of such characteristics on older adults who may one day rely on manual wheelchairs. In addition, the American with Disabilities Act Accessibility guidelines loosely defines what constitutes an “accessible” carpeted surface. It appears the current definition of “accessible” carpet should be revisited and further clarified. Based on our findings, we believe changes in both policies could facilitate improved mobility for all manual wheelchairs users by reducing barriers to service delivery and within the environment.

6.8.1 Centers for Medicare and Medicaid Services

The Centers for Medicare and Medicare services define manual wheelchairs primarily according to their weight and adjustability. Typically, the lighter and more adjustable the chair, the greater the price of the chair. As a cost savings approach, Medicare often rents manual wheelchairs for users. However, Medicare will only purchase the lightest and most adjustable wheelchairs, an implication temporary users would not benefit from such chairs. If Medicare purchases a chair for a user, Medicare pays 80% of the maximum allowable cost. This “maximum allowable” cost is often less than the actual price of the chair. The user must pay for the remaining cost, which at minimum is 20% of the cost of the chair. For older adults on a restricted income, the remaining cost of the lightest and most adjustable chairs could exceed their ability to pay. All this assumes that justification of such chairs has occurred. Currently, justification must be provided as to why the “least costly alternative” does not suffice for the user(27). Generally, the least costly alternative is a heavier and less adjustable chair. It does not appear that most older adults receive their wheelchair through a specialized seating and mobility clinic, which would be needed to provide such justification(5;33). Together, these policies and practices seem to drive older adults toward the heaviest, poorly fitted and configured wheelchairs. If this is the “standard of care” an older adult can expect to receive, then we are doing our older adults a disservice.

We believe our results support the provision of very light weight, adjustable chairs to older adults. Such configurations reduce the force needed to propel at a given velocity on any given surface. As noted above, these chairs are only purchased by Medicare, and Medicare often provides rentals. A solution therefore, would be to develop a rental fleet of such chairs and require such rentals be fitted by a qualified specialist. Although the initial investment cost for such chairs would be high, theses chair posses a higher durability, increasing their lifespan of use(34). Providers would conceivably profit over the long term through fewer repairs and a lower replacement rate. In addition, recent research shows developing a rental fleet of high end wheeled mobility for use in progressive diseases is a cost saving approach which fully supports the needs of the user(35). Requiring rentals to be fitting by certified specialists is simply an extension of the new Medicare policy for powered mobility. This would ensure and "optimal" configuration truly has been achieved. Together, these policy changes could alter service delivery for older adults in a manner which would facilitate improved mobility when a manual chair is required.

6.8.2 Americans with Disabilities Act Accessibility Guidelines (ADDAG)

Current ADAAG guidelines for carpet are as follows:

> **Carpet**. If carpet or carpet tile is used on a ground or floor surface, then it shall be securely attached; have a firm cushion, pad, or backing, or no cushion or pad; and have a level loop, textured loop, level cut pile, or level cut/uncut pile texture. The maximum pile thickness shall be 1/2 in (13 mm). Exposed edges of carpet shall be fastened to floor surfaces and have trim along the entire length of the exposed edge. Carpet edge trim shall comply with 4.5.2.(36)

Both carpets used in this project had less than a ½ pile. The low pile carpet was an uncut loop without a pad, while the high pile carpet was a cut loop with typical plush residential padding. Even the low pile carpet required more force to transverse at a lower speed compared to the tiled surface. The high pile carpet as configured with a pad did not meet ADA guidelines. However, it does represent a potential surface found within the home. Additional commentary with the ADA guidelines indicates a serious need for "quantitative and qualitative criteria for carpeting(36)." We support this call for such criteria. Our results confirms previous findings indicating carpet imposes increased demand on the user (11;16;37). A quantitative system rating surface difficulty would fill a much needed information gap. Such information could then be used to develop more precise guidelines to ensure consistency across facilities.

In facilities where mobility is important, i.e. airports, shopping malls, conference centers, and hotels, efforts should be made to ensure a low resistance surface is used. Where possible, all surfaces should be surfaced in firm tile, thereby eliminating surface type as a potential barrier to patrons who rely on manual wheeled mobility. If carpeting must be used, then firm low pile industrial carpet without a pad should be used. If the facility prefers a plush carpeted surface, efforts should be made to include a tiled pathway to all key areas to ease access by members of the community who use manual wheelchairs.

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