Published in final edited form as:
Arch Phys Med Rehabil. 2009 July ; 90(7): 1076–1083. doi:10.1016/j.apmr.2008.10.034.

Impact of Surface Type, Wheelchair Weight, and Axle Position on Wheelchair Propulsion by Novice Older Adults

Rachel E. Cowan, PhD, **Mark S. Nash, PhD**, **Jennifer L. Collinger, BS**, **Alicia M. Koontz, PhD**, and **Michael L. Boninger, MD**
Human Engineering Research Laboratories (Cowan, Collinger, Koontz, Boninger), Department of Physical Medicine and Rehabilitation (Collinger, Boninger), and School of Medicine (Boninger), University of Pittsburgh, Pittsburgh, PA; VA Pittsburgh Health Care System Center of Excellence in Wheelchairs and Related Technology, Pittsburgh, PA (Cowan, Collinger, Koontz, Boninger); Department of Rehabilitation Science and Technology (Cowan, Koontz, Boninger), School of Health and Rehabilitation Sciences, and Department of Bioengineering (Collinger, Koontz), University of Pittsburgh, Pittsburgh, PA; and Department of Neurological Surgery and the Miami Project to Cure Paralysis, Miller School of Medicine, University of Miami, Miami, FL (Nash).

Abstract

Objective—To examine the impact of surface type, wheelchair weight, and rear axle position on older adult propulsion biomechanics.

Design—Crossover trial.

Setting—Biomechanics laboratory.

Participants—Convenience sample of 53 ambulatory older adults with minimal wheelchair experience (65−87y); men = 20, women = 33.

Intervention—Participants propelled 4 different wheelchair configurations over 4 surfaces; tile, low carpet, high carpet, and an 8% grade ramp (surface, chair order randomized). Chair configurations included: (1) unweighted chair with an anterior axle position, (2) 9.05kg weighted chair with an anterior axle position, (3) unweighted chair with a posterior axle position (Δ0.08m), and (4) 9.05kg weighted chair with a posterior axle position (Δ0.08m). Weight was added to a titanium folding chair, simulating the weight difference between very light and depot wheelchairs. Instrumented wheels measured propulsion kinetics.

Main Outcome Measures—Average self-selected velocity, push-frequency, stroke length, peak resultant and tangential force.

Results—Velocity decreased as surface rolling resistance or chair weight increased. Peak resultant and tangential forces increased as chair weight increased, surface resistance increased, and with a posterior axle position. The effect of a posterior axle position was greater on high carpet and the ramp. The effect of weight was constant, but more easily observed on high carpet and ramp. The effects of axle position and weight were independent of one another.

Reprint requests to Michael L. Boninger, MD, Human Engineering Research Laboratories, VA Pittsburgh Health Care System, 5180 Highland Dr. 151R–1, Pittsburgh, PA 15206, e-mail: boninger@pitt.edu..

Conclusion—Increased surface resistance decreases self-selected velocity and increases peak forces. Increased weight decreases self-selected velocity and increases forces. Anterior axle positions decrease forces, more so on high carpet. Effects of weight and axle position are independent. Greatest reductions in peak forces occur in lighter chairs with anterior axle positions.

Keywords

Aged; Biomechanics; Rehabilitation; Wheelchairs

Adults aged 65 and older are the largest group of manual wheelchair users in the United States.[1] In 2006, the Centers for Medicare and Medicaid Services (CMS) reported spending more money on standard manual wheelchairs (K0001) for older adults than any other manual wheelchair classification (>$84 million).[2] Hoenig et al[3] and Hubbard et al[4] report older veterans are most commonly provided with manual wheelchairs matching the CMS K0001 category. Together, these data suggest older adults typically use standard wheelchairs classified by CMS as a K0001.

Wheelchairs in the K0001 classification weigh greater than 36lbs, are not adjustable, and are commonly referred to as standard or depot wheelchairs.[4] These chairs are the heaviest offered, weighing 20lbs (9kg) more than the lightest available wheelchairs. Older adults who have a manual wheelchair, but do not use it, report chair weight as a primary reason for non-use.[5] Studies examining the impact of wheelchair weight are limited. Bednarczyk and Sanderson[6] reported 5kg and 10kg weight additions did not affect propulsion kinetics of adults and children with spinal cord injury across a tiled floor. The authors suggested propulsion over more fatiguing surfaces might enhance weight-imposed differences and that kinetic or energy expenditure measurements might be more sensitive to weight additions than kinematic measures. To date, no studies have evaluated the impact of wheelchair weight on propulsion kinetics. The numerous differences between standard and ultralight/lightweight wheelchairs, plus the impact of the user-wheelchair interface, renders isolating wheelchair weight impact a challenge. However, all else equal, physics dictates that a lighter chair requires less force to propel. Energy expenditure studies have determined ultralight/light weight wheelchairs improve performance compared to standard wheelchairs[7-9], but have not isolated which feature (i.e. weight, axle position, manufacturing, etc..) resulted in the observed differences.

As noted, wheelchairs in the K0001 class are not adjustable. By contrast, non-K0001 wheelchairs allow adjustments to rear wheel axle vertical and horizontal position. All research evaluating the physiological and biomechanical impact of wheel axle position has studied either persons with spinal cord injury or non-disabled young adults.[10-14] Increasing vertical distance between the rear wheel axle and the position of the users shoulder, i.e. increasing seat height, increases oxygen cost[12], heart rate[12], push frequency[14], and peak force[10], and decreases stroke angle[10-12]. Boninger et al[10] reported increased stroke angle, decreased push frequency, and decreased peak forces are correlated with increasing anterior axle positions. In contrast, Kotajarvi et al[11] reported self-selected speed, push frequency, stroke time, stroke distance, tangential force, peak propulsion moments and mean fraction effective force were not affected by clinically reproducible horizontal axle position changes. However, higher axle placements (lower seat height) increased stroke length and axial forces. As evidenced by the results of Kotajarvi et al[11], clinically relevant changes in axle position may impart minimal changes in propulsion biomechanics.

Few studies have investigated propulsion biomechanics of older adult manual wheelchair users. Aissaoui et al[15] evaluated the effect of nine combinations of seat tilt (3 seat tilt angles) and backrest angle (3 backrest tilt angles) on the stationary propulsion mechanics of 14 manual wheelchair users over age 62. A seat tilt or backrest recline of 10° improved fraction effective force by 10%. In a follow-up study[16], shoulder loads remained constant as seat tilt and backrest angle change, as long as the position of the shoulder relative to the axle remained constant. Studies based on this cohort represent the sum knowledge to date of propulsion mechanics in older adults.

Daily wheelchair propulsion occurs on many surfaces, yet the impact of surface type on propulsion mechanics remains poorly understood. Most propulsion biomechanical assessments are conducted on stationary systems, such as ergometers, dynamometers, and treadmills. These systems confer methodological benefits, and can simulate graded propulsion, but cannot fully replicate overground propulsion. Graded surfaces, or simulated grades increase forces[17,18], decrease self-selected velocities[17,18], and decrease stroke length.[18] Koontz et al[17] reported that ramped, grass, and interlocking paver surfaces required greater wheel torques than indoor tile, low-pile, high-pile, and wood surfaces. No differences were observed among the indoor surfaces of tile, low pile carpet, and high pile carpet. In contrast to Koontz et al[11], Wolfe et al[19] observed a decrease in self-selected velocity of 0.29 m/s between level concrete and carpeting of the same type used in hospitals and nursing homes, in a mixed group of deconditioned, normal, and paraplegic persons. This decrease in velocity served to minimize differences in energy cost between the surfaces.

The wheelchair propulsion characteristics of older adults have been understudied. Given the largest group of manual wheelchair users in the United States is older adults, it is important to define how wheelchair characteristics and surface conditions affect their propulsion biomechanics. Understanding their propulsion mechanics may serve to highlight new approaches to wheelchair selection and configuration for this population. Therefore, the purpose of this project is to assess the impact of surface type, wheelchair weight, and horizontal axle position on the propulsion biomechanics of older adults.

METHODS

Participants

Fifty-three older community dwelling adults were recruited through local senior center flyers, interest groups, bring-a-friend strategies, and Institutional Review Board- (IRB) approved research registries. This research protocol was approved by the IRB of the VA Pittsburgh Health Care System and the University of Pittsburgh. All subjects gave written informed consent prior to participation. To be eligible for participation, subjects had to self-report: (1) age ≥ 65 years, (2) ability to walk without human assistance, (3) ability to independently stand from a chair, (4) ability to independently self-propel a wheelchair, (5) weight ≤ 113.12 kg (250 lbs), and (6) have a score ≥ 21 on the mini-mental state exam.[20] Exclusion criteria included self-reported history of stroke or a diagnosis of Parkinson or Alzheimer disease. Inclusion and exclusion criteria were selected to maximize participant safety and protocol completion.

Propulsion Surfaces

Four common surfaces were selected for testing: hallway tile (12.0m), low pile carpet on concrete without pad (7.3m), high pile carpet with a 1.11 cm pad on concrete (7.3m) and a wooden uncarpeted Americans with Disabilities Act-compliant ramp (2.5m, 8% grade). Rolling resistance was not quantified, but each surface represented a distinct level of difficulty. Surface order was randomized for each participant and for each chair configuration.

Initial Wheelchair Adjustment

Three TiLite[a] titanium folding chairs (model X) were secured for this study. Seat dimensions were 38.1 × 40.6, 43.2 × 43.2, 48.26 × 45.72cm (15 × 16, 17 × 17, and 19 × 18 in). Each weighed 11.31kgs (25lbs); with fabric seat/back upholstery, a 5.08cm foam cushion, removable plastic side guards, anti-tip devices, pneumatic rear wheels, solid 10.16 cm diameter casters, aluminum hand rims, and had 0° seat inclination and camber.

Individuals were seated in the narrowest possible wheelchair. Vertical rear axle position was adjusted so each subject achieved an elbow flexion angle of 100° to 120° when seated with hands at pushrim top center.[12] Footrests were adjusted to provide support for the feet and thighs as chair design allowed. Front and rear seat heights remained as equal as possible to maintain seat inclination at 0°. Front caster angle was adjusted to minimize caster trail.

Wheelchair Test Configurations

Two weight configurations (unweighted and weighted, Δ9.05kg (20lbs)) and 2 rear axle positions (anterior and posterior, Δ0.08m (3.1in)) were tested. The 4 test configurations were: (1) unweighted with an anterior axle position; (2) weighted with an anterior axle position; (3) unweighted with a posterior axle position; and (4) weighted with a posterior axle position. The unweighted configuration was the delivery weight (11.31kg/25lbs). The weighted configuration added 9.05kg (20lbs) of weight to the chair (chair + weight = 20.36kg/45lbs). The 9.05kg mimicked the weight differential between the lightest chairs available and standard/depot chair weight. The weight consisted of custom cut lengths of 2.54cm square brass bars attached to the cross bars and seat rails via industrial velcro and clamps. Weight was added in a manner that maintained the relative front/rear wheel weight distribution of the posterior configuration. The anterior axle position was the most forward rear axle position allowed by the manufacturer's bracket system, and had less rolling resistance and stability than the posterior axle position.[11] All configurations included anti-tip devices for safety. The posterior axle position was 0.08m rearward of the anterior position and was the most posterior axle position allowed by the manufacturer's bracket system. The weighted, posterior axle configuration simulated the weight and axle position of a K0001 wheelchair. Participants propelled all 4 configurations in a randomized order. Prior to data collection, each participant completed a six- minute propulsion task in the weighted, posterior axle configuration. The task consisted of propulsion at a self-selected speed, while turning in a self-selected direction on a 60 meter loop on a linoleum tile hallway.

Propulsion Data Collection

SmartWheels[b] provided bilateral 6 degrees of freedom kinetic data collection (240 Hz). Each SmartWheels had a solid treaded tire and weighed ~4.98 kg (11lbs). With bilateral SmartWheels attached, the test weight of the chairs was 18.10kg (40lbs) unweighted and 27.15kg (60lbs) weighted. With the SmartWheels attached and the axle in the posterior position, the weight distribution between front and rear wheels was 28% and 72%; with the axle in the anterior position, 23% and 77%. Without the SmartWheels, the weight distributions of the chairs were 36% and 64% with the posterior position; and 20% and 80% in the anterior.

Participants were instructed to begin in a stationary position with hands in their lap, accelerate to a comfortable velocity, and continue until they exited the surface or were instructed to stop. Participants began on the surface for tile, low pile carpet, and high pile carpet. For ramp testing, participants began on concrete level ground directly in front of the ramp with the front casters within 7.6 cm of the ramp. Data collection for all testing was initiated before initial hand-to-

---

[a] TiLite Model X; TiSport, LLC, 1426 East Third Avenue, Kennewick, WA 99337.

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

rim contact, and terminated before the chair exited the test surface (low and high pile carpet) or the marked distance (tile and ramp). Data collection was terminated if the participant experienced difficulty completing a surface. After completing all surfaces in a configuration, participants rested for 2 minutes while the chair was changed to the next configuration. In 1 day of testing, participants completed testing on 4 surfaces for each of 4 chair configurations, making a total of 16 trials per participant.

Kinetic Data Reduction

The SmartWheels sign convention was defined as Fx + = anterior, Fy+ = superior, Fz+ = medial (left wheel), lateral (right wheel). Positive moments were defined as counterclockwise about the respective force vector. A stroke was defined as a propulsive contact (negative Mz). A cycle was defined as the period encompassing a propulsive contact and the subsequent recovery.

The initial 3 cycles from a stationary position and the deceleration that occurred as participants approached the end of the level surfaces were trimmed prior to key variable computation. The initial 3 cycles are thought to comprise the bulk of the initial acceleration from a stationary position.[17] All trimmed data thus began with cycle 4 and extended to however many cycles occurred prior to deceleration. Variables were calculated for each of these cycles and then averaged to provide a general representation of propulsion beyond the initial acceleration phase of the first 3 cycles. If fewer than 4 cycles were recorded, which occurred intermittently on the ramp, variables were not computed. If a maximum of 4 cycles were recorded, the variables were represented by the fourth cycle.

Key Kinetic Variables

The following variables were calculated bilaterally and averaged for each complete cycle in each trimmed trial: average linear velocity, push frequency, stroke length, peak resultant force[11,17], and peak tangential force.[10,11] These variables provide a comprehensive description of propulsion. A search algorithm automated the identification of contact and recovery. Accuracy was verified through visual inspection, with necessary adjustments made on a per cycle basis. Average velocity was the average linear velocity (m/s) of the wheel during the cycle derived from the onboard encoder. 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 below −0.6 Nm for a minimum of 0.1s. Push frequency (Hz) was calculated as 1/cycle time(s). Resultant force (N) was defined as the vector sum of Fx, Fy, and Fz.[10] Tangential force (N) was calculated as the absolute value of Mz/pushrim radius (0.2667m). Resultant and tangential forces were normalized to body weight prior to analysis. Matlab[c] was used to trim the data, identify cycles, and compute variables. All data were filtered by the SmartWheels software prior to export to Matlab.

Statistical Analysis

SPSS[d] was used for all analyses. Data were inspected for normality. To assess the effect of surface, data were collapsed across the four wheelchairs and entered into a one-factor repeated measures analysis of variance (ANOVA). A separate ANOVA was calculated for each of the 5 outcome measures. Significant main effects for surface were tested with pairwise comparisons. To determine differences due to wheelchair weight, axle position, and their interaction, 2 (weight) × 2 (axle position) repeated measures ANOVAs were calculated for each of the 5 outcome measures for each surface. While the α level for significance was ≤ 0.05,

---

[c] Matlab; The MathWorks, Inc, 24 Prime Park Way, Natick, MA 01760−1500.

[d] SPSS; SPSS Inc, 233 South Wacker Dr, 11th Fl, Chicago, IL 60606−6307.

a Holm-modified Bonferonni correction was applied to control for type I error due to multiple comparisons.

RESULTS

Fifty-three older adults who reported minimal previous experience propelling a manual wheelchair participated (men = 20, women =33). Average age was 73.6 years (+/−5.4y), ranging from 65 to 87. Participants were on average overweight (body mass index = 27.6+/−5.1 kg/m^2, height = 1.7+/−0.1 m, weight = 76.5+/−16.7 kg). Mean elbow angle after initial fitting was 107.1+/−6.3° (full extension = 180°). One participant was unable to achieve an elbow angle within the specified range of 100° − 120° (angle = 125°).

Surface

Surface main effects were significant for all outcome measures ($p < 0.000$ all) (tables 1, 2). All pairwise comparisons of average velocity were significant ($p < 0.000$) (see table 1). As rolling resistance increased, average velocity decreased (ramp < high pile carpet < low pile carpet < tile). Push frequency was lowest on the high carpet and ramp, and highest on tile and low carpet. Stroke length was shorter on the ramp than all other surfaces. All pairwise comparisons of peak resultant and tangential force were different: As rolling resistance increased, peak force increased (see table 2). To summarize, as rolling resistance increased, participants self-selected a lower velocity using greater forces. Decreased velocity on high carpet and ramp was accompanied by a lower push frequency. The ramped condition was traversed at the slowest velocity using the highest forces, lowest push frequency and shortest stroke length.

Wheelchair Weight

On all surfaces the un-weighted condition was pushed faster, with lower resultant and tangential forces than the weighted condition (see tables 1, 2). The un-weighted condition resulted in a higher push frequency on high carpet and ramp than the weighted condition ($p<0.000$) (see table 1). There was no difference in push frequency between the weight conditions on tile and low carpet (see table 1). Stroke length was unchanged by weight on all surfaces (see table 1).

Axle Position

The posterior condition was propelled faster than the anterior condition on the tile and ramp (see tables 1). Velocity was unaffected by axle position on low carpet and high carpet. On all surfaces, push frequency and stroke length were unaltered by axle position (see table 1). On all surfaces, the anterior condition required lower resultant and tangential forces than the posterior condition (see table 2).

Weight by Axle Position Interactions

There was no significant weight by axle position interaction. However, on all surfaces, the unweighted anterior axle configuration required lower forces to travel at similar or higher velocities than the weighted posterior axle configuration.

DISCUSSION

The key findings of this study are that (1) surface type substantially affects self-selected velocity, peak resultant force, and peak tangential force; (2) a 9kg weight addition reduces self-selected velocity while increasing peak forces on each surface; and (3) an anterior axle position decreased peak forces on every surface, with the largest decrease occurring on high carpet. In contrast to velocity and force, push frequency and stroke length are less affected by surface type, wheelchair weight, and axle position. These variables may be more responsive to changes in vertical axle position or propulsion training interventions than alterations in horizontal axle position, wheelchair weight, and surface type.

Surface

Surface type has a substantial impact on self-selected velocity and peak forces in older adults. Absolute differences in self-selected average velocity on the surfaces ranged from 0.06 m/s (high carpet vs ramp) to 0.64 m/s (tile vs ramp) (table 3). These absolute differences are equivalent to relative changes of 13.95% and 63.37%, and Cohen's d effect sizes of 0.96 and 6.48 respectively (see table 3). Perera et al[21] suggests 0.05 m/s and 0.10 m/s, respectively represent clinically meaningful small and substantial changes in 10-foot gait speed in older adults. We thus propose self-selected propulsion velocity changes ≥0.05 m/s is clinically meaningful. These decreases in velocity may be a strategy to minimize the oxygen cost of propulsion.[19] Clinicians cannot change a user's environment, but they can alter the user's wheelchair. Defining how clinically reproducible changes in wheelchair configuration affect propulsion on common surfaces may provide clinicians and consumers with objective evidence to guide chair selection and configuration.

Decreased velocity was accompanied by substantially increased peak resultant force; ranging from 12.37% (tile vs low carpet) to 88.86% (tile vs ramp), with effect sizes of 1.01 and 6.47 respectively (see table 3). This contrasts the results of Koontz et al[17] who reported no difference in peak velocity or peak force between tile, low pile carpet, and high pile carpet. We attribute our different result to study population differences. Our larger, homogenous sample (N=53 vs N = 13[17]) provided greater power to detect differences. In addition our novice wheelchair users may have responded differently to increased surface difficulty than the experienced users of Koontz et al.[17] However, our results of decreased self-selected velocity coupled with increased peak force, decreased push frequency, and decreased stroke length in the ramp vs. tile comparison corroborate findings based on a different group of experienced users.[18]

Decreasing velocity was generally accompanied by reduced push frequency and an unchanged stroke length. However, the velocity decrease from tile to low carpet was not accompanied by a decreased push frequency (see table 1). This may indicate the correlation between push frequency and velocity is specific to propulsion surface[10]. Stroke length essentially remained unchanged on level surfaces despite changes in velocity and peak force. During level propulsion, stroke length may primarily be a function of user-chair interface and user experience. As noted above, the smaller stroke length on the ramp is in agreement with a previous study[18]. Given that our sample and methodology differed substantially from this study[18], it appears ramps induce a predictable change in propulsion biomechanics.

Weight

On each surface, older adults self-selected a lower speed using greater peak resultant and tangential force in the weighted condition. The lower velocity and higher force of the weighted condition was coupled with a lower push frequency on high carpet and ramp. The lower push frequency could be attributed to the lower velocity. Absolute, relative, and Cohen's d effect sizes of weight for resultant force appear constant across surfaces (table 4). In addition the absolute effect of weight on velocity appears constant (see table 4). However, weight has a greater relative effect on velocity for the high carpet and ramp compared to the tile and low carpet (see table 4). Together, these results suggest increased wheelchair weight has the same impact on every surface, but is most readily observed on a high carpet or ramped condition.

Axle Position

To isolate the effects of horizontal axle position changes, each user was individually fitted to achieve the recommended 100° to 120° elbow angle.[12] In addition, insofar as possible, a level seat inclination was maintained for both axle positions. Seat inclination affects anterior/ posterior weight distribution, which in turn affects rolling resistance; force required for propulsion and may moderate the effect of axle position.

The anterior configuration required less peak resultant and tangential force than the posterior configuration on all surfaces. However, these differences were significant for only 3 of the 4 surfaces (low carpet, high carpet, ramp). The force difference between axle positions changed as surface difficulty increased (see table 4). For peak resultant force, a medium effect size was observed on high carpet and ramp, and a small effect observed on tile and low pile carpet (see table 4). These data suggest an interaction between axle position and surface type. The greatest difference in peak force between the axle positions occurred on the most difficult level surface. We feel these results support the suggestion of Bednarczyk and Sanderson[6] to evaluate propulsion on more fatiguing surfaces as a technique to magnify the effect of chair configuration changes.

Push frequency and stroke length were unaffected by horizontal axle position. On more difficult surfaces, lighter chairs resulted in lower push frequencies than heavier chairs. Stroke length appears unaffected by changes in horizontal axle position and wheelchair weight and level surface type. As previously noted stroke length may be primarily affected by user experience and vertical axle position. We suggest that clinicians who wish to increase client stroke length and/or decrease push frequency explore vertical axle position adjustments and propulsion training.

It is important to note the axle positions evaluated represent opposing extremes commonly available to clinicians. Realistically, most users will have an axle position in between these positions. In addition, the adjustments made by clinicians will be much smaller than the 8cm we evaluated. While effect size of axle position on peak force appears modest, energy expenditure measurements may be more sensitive than kinetic or kinematic measures to clinically relevant changes. Use of self-selected velocity allowed us to determine the natural response of our subjects to changes in axle position, wheelchair weight, and surface type. However, fixing velocity may enhance the effect of surface type, wheelchair weight, and axle position. We suggest future research employ both self-selected and fixed velocity conditions.

Weight and Axle Position Interactions

A lack of interaction between wheelchair weight and axle position suggests a decrease in wheelchair weight will reduce peak resultant force regardless of axle position. Clinicians can maximize reductions in peak resultant force by securing the lightest possible wheelchair and shifting axle position as far forward as tolerated by their client. If wheelchair stability is a concern, securing an ultralight wheelchair and selecting a posterior axle position will maintain stability while decreasing required propulsion forces. Of note, the most posterior axle position available on untralight wheelchairs is anterior of the axle position on standard wheelchairs. Therefore, the weighted posterior condition we used to simulate a K0001 may perform better than most standard/depot wheelchairs; especially considering we provided fitting not available on standard/depot wheelchairs.

Study Limitations

The study findings are delimited by testing of older persons who were naïve to wheelchair propulsion, limited time for training, additional weight imposed by the SmartWheels, evaluation over short propulsion distances, and lack of quantification of surface rolling resistance. Use of older adults who were inexperienced in wheelchair propulsion may limit generalization to younger persons or experienced users. However, Koontz et al[17] found increased force and decreased velocity in a group of younger, experienced wheelchair users, suggesting the effect of surface is universal to all manual wheelchair users. Formalized propulsion training could have benefited our participants, but conversations with clinicians suggested older adults receive little formal instruction. Thus while we believe formal instruction and guided practice should be the clinical norm and would have improved the propulsion of our cohort, we attempted to create an opportunity for learning prior to data collection without introducing an artificial level of instruction. While the SmartWheels add weight, change the weight distribution, and undoubtedly affect propulsion, they remain the only commercially available method to assess the kinetics of propulsion, which were primary outcome variables of interest. Alternative methods to quantify the impact of surface, axle position, and wheelchair weight include electromyographic, oxygen consumption, and motion capture evaluations. Use of these systems would facilitate measurement without fundamentally altering the user-wheelchair system. Short propulsion distances may invoke responses different from longer distances, but simulates evaluation conducted in limited clinic space and mimics propulsion in confined spaces found in many homes. Finally, quantification of the rolling resistance of each surface would have improved the precision of our study.

CONCLUSIONS

Surface type has a substantial impact on self-selected velocity and peak resultant and tangential forces in older adults. The effect of a heavier wheelchair on self-selected velocity and peak forces is most pronounced on high carpet and ramp. Anterior axle positions decrease peak forces. The magnitude of this effect increases as surface difficulty increases. The effects of weight and axle position appear to be independent. The greatest reductions in peak resultant force will be obtained by securing the lightest possible wheelchair and then shifting the axle as far forward as tolerated by the client.

Acknowledgments

Supported by the National Institutes of Health (grant nos. 1 F31 HD053986-01, P30 AG024827, AG023641), National Science Foundation (DGE0333420, Graduate Research Fellowship), and the Department of Veterans Affairs Rehabilitation Research and Development (grant no. B3142C).

We certify that we have affiliations with or financial involvement (eg, employment, consultancies, honoraria, stock ownership or options, expert testimony, grants and patents received or pending, royalties) with an organization or entity with a financial interest in or financial conflict with, the subject matter or materials discussed in the manuscript. Rachel Cowan has a non-financial affiliation with Three Rivers Holdings, Inc. in the form of sub-contracted grants.

List of Abbreviations

ANOVA, analysis of variance; CMS, Centers for Medicare and Medicaid Services; IRB, Institutional Review Board.

References

1. Kaye, HS.; Laplante, MP. Disability Statistics Report. 14. US Department of Education, National Institute on Disablity and Rehabilitation Research; Washington, DC: 2000. Mobility Device Use in the United States..
2. Centers for Medicare and Medicaid Services. Part B Physician/Supplier National Data – CY 2006, Top 200 Level 11 Healthcare Procedure Coding System (HCPCS/Alpha-Numeric) Codes. 2006 [September 3, 2008]. from http://www.cms.hhs.gov/MedicareFeeforSvcPartsAB/Downloads/LEVEL2CHARG06.pdf

3. Hoenig H, Landerman LR, Shipp KM, Pieper C, Pieper C, Richardson M, et al. A clinical trial of a rehabilitation expert clinician versus usual care for providing manual wheelchairs. J Am Geriatr Soc 2005;53:1712–20. [PubMed: 16181170]
4. Hubbard SL, Fitzgerald SG, Reker DM, Boninger ML, Cooper RA, Kazis LE. Demographic characteristics of veterans who received wheelchairs and scooters from Veterans Health Administration. J Rehabil Res Dev 2006;43:831–44. [PubMed: 17436170]
5. Mann WC, Goodall S, Justiss MD, Tomita M. Dissatisfaction and nonuse of assistive devices among frail elders. Assist Technol 2002;14:130–9. [PubMed: 14651251]
6. Bednarczyk JH, Sanderson DJ. Limitations of kinematics in the assessment of wheelchair propulsion in adults and children with spinal cord injury. Phys Ther 1995;75:281–9. [PubMed: 7899486]
7. Hilbers PA, White TP. Effects of wheelchair design on metabolic and heart rate responses during propulsion by persons with paraplegia. Phys Ther 1987;67:1355–8. [PubMed: 3628489]
8. Beekman CE, Miller-Porter L, Schoneberger M. Energy cost of propulsion in standard and ultralight wheelchairs in people with spinal cord injuries. Phys Ther 1999;79:146–58. [PubMed: 10029055]
9. Parziale JR. Standard v lightweight wheelchair propulsion in spinal cord injured patients. Am J Phys Med Rehabil 1991;70:76–80. [PubMed: 2009163]
10. Boninger ML, Baldwin M, Cooper RA, Koontz A, Chan L. Manual wheelchair pushrim biomechanics and axle position. Arch Phys Med Rehabil 2000;81:608–13. [PubMed: 10807100]
11. Kotajarvi BR, Sabick MB, An KN, Zhao KD, Kaufman KR, Basford JR. The effect of seat position on wheelchair propulsion biomechanics. J Rehabil Res Dev 2004;41:403–14. [PubMed: 15543458]
12. van der Woude LH, Veeger DJ, Rozendal RH, Sargeant TJ. Seat height in handrim wheelchair propulsion. J Rehabil Res Dev 1989;26:31–50. [PubMed: 2600867]
13. Mulroy SJ, Newsam CJ, Gutierrez DD, Requejo P, Gronley JK, Haubert LL, et al. Effect of fore-aft seat position on shoulder demands during wheelchair propulsion: part 1. A kinetic analysis. J Spinal Cord Med 2005;28:214–21. [PubMed: 16048139]
14. Masse LC, Lamontagne M, O'Riain MD. Biomechanical analysis of wheelchair propulsion for various seating positions. J Rehabil Res Dev 1992;29:12–28. [PubMed: 1640378]
15. Aissaoui R, Arabi H, Lacoste M, Zalzal V, Dansereau J. Biomechanics of manual wheelchair propulsion in elderly: system tilt and back recline angles. Am J Phys Med Rehabil 2002;81:94–100. [PubMed: 11807343]
16. Desroches G, Aissaoui R, Bourbonnais D. Effect of system tilt and seat-to-backrest angles on load sustained by shoulder during wheelchair propulsion. J Rehabil Res Dev 2006;43:871–82. [PubMed: 17436173]
17. Koontz AM, Cooper RA, Boninger ML, Yang Y, Impink BG, van der Woude LH. A kinetic analysis of manual wheelchair propulsion during start-up on select indoor and outdoor surfaces. J Rehabil Res Dev 2005;42:447–58. [PubMed: 16320141]
18. Richter WM, Rodriguez R, Woods KR, Axelson PW. Stroke pattern and handrim biomechanics for level and uphill wheelchair propulsion at self-selected speeds. Arch Phys Med Rehabil 2007;88:81–7. [PubMed: 17207680]
19. Wolfe GA, Waters R, Hislop HJ. Influence of floor surface on the energy cost of wheelchair propulsion. Phys Ther 1977;57:1022–7. [PubMed: 142993]
20. Mitrushina M, Satz P. Reliability and validity of the Mini-Mental State Exam in neurologically intact elderly. J Clin Psychol 1991;47:537–43. [PubMed: 1939698]
21. Perera S, Mody SH, Woodman RC, Studenski SA. Meaningful change and responsiveness in common physical performance measures in older adults. J Am Geriatr Soc 2006;54:743–9. [PubMed: 16696738]

Table 1

Means and Standard Errors of Average Velocity, Push Frequency, and Stroke Length for All Levels of Each Main Effect and Interaction, by Surface.

		Average Velocity (m/s)				**Push Frequency (Hz)**				**Stroke Length (Deg)**			
Effect	**Mean (Standard Error)**	**Tile N = 53**	**Low Carpet N =53**	**High Carpet N =53**	**Ramp[*] N = 39**	**Tile N = 53**	**Low Carpet N = 53**	**High Carpet N = 53**	**Ramp[*] N = 39**	**Tile N = 53**	**Low Carpet N = 53**	**High Carpet N = 53**	**Ramp[*] N = 39**
Weight × Axle	Unweighted Anterior	1.05 (0.03) P= 0.874	0.89 (0.03) P=0.977	0.47 (0.02) P=0.963	0.35 (0.02) P=0.375	1.19 (0.04) P=0.963	1.20 (0.04) P=0.127	0.98 (0.04) P=0.724	0.94 (0.04) P=0.069	55.91 (1.84) P=0.891	54.20 (1.60) P=0.291	55.00 (1.99) P=0.460	46.01 (1.74) p=0.306
	Weighted Anterior	1.02 (0.03)	0.86 (0.03)	0.44 (0.02)	0.32 (0.02)	1.18 (0.04)	1.23 (0.04)	0.93 (0.03)	0.89 (0.04)	55.84 (1.96)	53.25 (1.85)	54.04 (1.77)	44.39 (1.54)
	Unweighted Posterior	1.09 (0.03)	0.88 (0.03)	0.47 (0.03)	0.38 (0.02)	1.17 (0.04)	1.20 (0.04)	0.95 (0.03)	0.99 (0.04)	54.78 (1.84)	53.32 (1.73)	54.39 (1.89)	45.45 (1.41)
	Weighted Posterior	1.05 (0.03)	0.85 (0.03)	0.44 (0.03)	0.33 (0.03)	1.16 (0.04)	1.17 (0.04)	0.91 (0.03)	0.88 (0.04)	54.88 (1.83)	53.73 (1.83)	54.24 (1.80)	45.76 (1.77)
Weight	Unweighted	1.07 (0.03) **P=0.012**[†]	0.89 (0.03) **P=0.026**[†]	0.47 (0.02) **P<0.000**[†]	0.37 (0.02) **P<0.000**[†]	1.18 (0.04) P=0.769	1.20 (0.04) P=0.959	0.97 (0.03) **P<0.000**[†]	0.97 (0.04) **P<0.000**[†]	55.34 (1.75) P=0.983	53.76 (1.60) P=0.704	54.70 (1.90) P=0.452	45.73 (1.46) P=0.460
	Weighted	1.03 (0.03)	0.86 (0.03)	0.44 (0.02)	0.33 (0.02)	1.17 (0.04)	1.20 (0.04)	0.92 (0.03)	0.89 (0.04)	55.36 (1.83)	53.49 (1.76)	54.14 (1.73)	45.08 (1.54)
Axle	Anterior	1.03 (0.03) **P = 0.029**[†]	0.88 (0.03) P=0.606	0.45 (0.02) P=0.967	0.33 (0.02) **P=0.002**[†]	1.19 (0.04) P=0.312	1.21 (0.04) P=0.128	0.96 (0.03) P=0.078	0.92 (0.04) P=0.386	55.87 (1.84) P=0.221	53.73 (1.64) P=0.804	54.52 (1.82) P=0.746	45.20 (1.52) P=0.616
	Posterior	1.07 (0.03)	0.87 (0.03)	0.45 (0.02)	0.36 (0.02)	1.17 (0.03)	1.19 (0.04)	0.93 (0.03)	0.93 (0.04)	54.83 (1.78)	53.53 (1.73)	54.31 (1.79)	45.60 (1.46)
Surface	Surface Mean[*] N = 39	1.01 (0.02) **p<0.000**[†,‡]	0.84 (0.01)	0.43 (0.01)	0.37 (0.01)	1.21 (0.02) **p<0.000**[†,‖]	1.22 (0.02)	0.96 (0.02)	0.92 (0.02)	52.54 (0.86) **p<0.000**[†,§]	51.61 (0.85)	51.60 (0.88)	48.29 (0.90)

Note. P values are listed in the first row of each effect or interaction. Significant P-values are in bold. Unweighted and weighted means are from data collapsed across axle position. Anterior and posterior means are from data collapsed across weight condition. Surface means are from data collapsed across all four chair configurations. For surface pairwise comparisons, the α level was adjusted using a Holm modified Bonferonni correction to evaluate the significance level of each pairwise comparison.

[*] 14 individuals were dropped from the analysis because they completed less than 4 cycles on the ramp in at least 1 chair configuration, resulting in missing data. This limited all analysis which included the ramp to the 39 individuals who had at least 4 cycles in each chair configuration on the ramp.

[†] Significant main effect.

[‡] All pairwise comparisons p < 0.000.

[§] Ramp significant from all other surfaces, p < 0.000 for all. All other comparisons non-significant (tile vs. low carpet., p = 0.047; tile vs. high carpet., p = 0.112, low carpet vs. high carpet, p = 0.969).

[‖] All pair wise comparisons p < 0.000, except for tile versus low carpet, p = 0.309.

Table 2

Means and Standard Errors of Peak Resultant and Tangential Force for All Levels of Each Main Effect and Interaction by Surface.

Effect	Mean (Standard Error)	Peak Resultant Force (%BW)				Peak Tangential Force (%BW)			
		Tile N = 53	**Low Carpet N = 53**	**High Carpet N = 53**	**Ramp[*] N = 39**	**Tile N = 53**	**Low Carpet N = 53**	**High Carpet N = 53**	**Ramp[*] N = 39**
Weight × Axle	Unweighted Anterior	9.00 (0.34) P=0.700	10.04 (0.36) P=0.628	13.83 (0.32) P=0.727	15.78 (0.42) P=0.213	6.60 (0.26) P=0.844	7.50 (0.27) P=0.580	10.78 0.26) P=0.8.26	12.52 (0.26) P=0.123
	Weighted Anterior	9.47 (0.35)	10.37 (0.33)	14.28 (0.31)	16.55 (0.46)	6.96 (0.25)	7.89 (0.26)	11.29 (0.27)	13.25 (0.31)
	Unweighted Posterior	9.25 (0.38)	10.31 (0.37)	15.22 (0.34)	16.99 (0.44)	6.85 (0.27)	7.70 (0.25)	11.77 (0.27)	13.14 (0.31)
	Weighted Posterior	9.61 (0.43)	10.79 (0.43)	15.56 (0.37)	17.20 (0.40)	7.16 (0.30)	8.23 (0.31)	12.26 (0.31)	13.49 (0.31)
Weight	Unweighted	9.12 (0.33) **P=0.024**[†]	10.18 (0.35) **P=0.011**[†]	14.53 (0.31) **P=0.011**[†]	16.38 (0.41) **P=0.009**[†]	6.72 (0.24) **P=0.017**[†]	7.60 (0.25) **P=0.001**[†]	11.27 (0.25) **P<0.000**[†]	12.83 (0.28) **P<0.000**[†]
	Weighted	9.54 (0.37)	10.58 (0.36)	14.92 (0.32)	16.88 (0.40)	7.06 (0.26)	8.06 (0.27)	11.77 (0.28)	13.37 (0.30)
Axle	Anterior	9.23 (0.23) P=0.749	10.21 (0.33) **P=0.032**[†]	14.05 (0.30) **P<0.00**[†]	16.17 (0.42) **P<0.000**[†]	6.78 (0.24) P=0.168	7.69 0.26) **P=0.015**[†]	11.02 (0.25) **P<0.000**[†]	12.89 (0.30) **P=0.001**[†]
	Posterior	9.43 (0.39)	10.55 (0.38)	15.39 (0.33)	17.09 (0.39)	7.00 (0.26)	7.96 (0.26)	12.01 (0.28)	13.31 (0.30)
Surface	Surface Mean[*] N = 39	8.89 (0.18) **p<0.000**[†,‡]	9.99 (0.17)	14.48 (0.17)	16.79 (0.21)	6.53 (0.13) **p<0.000**[†,‡]	7.49 (0.12)	11.26 (0.14)	13.31 (0.15)

Note. P values for each effect or interaction are listed in the first row of each effect or interaction. Significant P-values are in bold. Un-weighted and weighted means are from data collapsed across axle position. Anterior and posterior means are from data collapsed across weight condition. Surface means are from data collapsed across all four chair configurations. For surface pairwise comparisons, the α level was adjusted using a Holm modified Bonferonni correction to evaluate the significance level of each pairwise comparison.

Abbreviation: BW, body weight.

[*] 14 individuals were dropped from the analysis because they completed less than 4 cycles on the ramp in at least 1 chair configuration, resulting in missing data. This limited all analysis which included the ramp to the 39 individuals who had at least 4 cycles in each chair configuration on the ramp.

[†] Significant main effect.

[‡] All pairwise comparisons significant, p < 0.000 for all.

Table 3

Average Velocity and Peak Resultant Force Effects Sizes for Surface Type

	Average Velocity		
	Low Carpet	**High Carpet**	**Ramp**
Tile	0.17m/s	0.58m/s	0.64m/s
	16.83%	57.43%	63.37%
	1.72	5.87	6.48
Low Carpet	-	0.41m/s	0.47m/s
		48.81%	55.95%
		6.57	7.53
High Carpet	-	-	0.06m/s
			13.95%
			0.96
	Resultant Force		
Tile	1.1 %BW	5.59 %BW	7.90 %BW
	12.37%	62.88%	88.86%
	1.01	5.11	6.47
Low Carpet	-	4.49 %BW	6.80 %BW
		44.94%	68.07%
		4.23	5.70
High Carpet	-	-	2.31 %BW
			15.95%
			1.94

Note. Data presented in descending order are absolute difference (m/s or %BW), relative difference (% change), and Cohen's d. Absolute difference = Absolute value (row mean – column mean). Relative difference = absolute difference/(row mean) * 100. Cohen d = absolute difference/(pooled standard deviation of row and column means).

Abbreviation: BW, body weight.

Table 4

Average Velocity and Peak Resultant Force Effects Sizes for Weight and Axle Position on Each Surface

	Average Velocity			
	Tile	**Low Capet**	**High Carpet**	**Ramp**
Weight	0.04m/s 3.74% 0.18	0.03m/s 3.37% 0.05	0.03m/s 6.38% 0.21	0.04m/s 10.81% 0.26
Axle	0.04 m/s 3.88% 0.18	0.01ms 1.14% 0.05	0.00m/s - -	0.03m/s 9.09% 0.24
	Resultant Force			
Weight	0.42%BW 4.61% 0.18	0.40%BW 3.93% 0.15	0.39%BW 2.68% 0.17	0.50%BW 3.05% 0.16
Axle	0.20 %BW 2.17% 0.08	0.34%BW 3.33% 0.13	1.34%BW 9.54% 0.58	0.92%BW 5.69% 0.36

Data presented in descending order are absolute difference (m/s or %BW), relative difference (% change), and Cohen's d for each effect. Absolute difference = Absolute value(un-weighted mean – weighted mean)or Absolute value(anterior mean – posterior mean). Relative difference = absolute difference/(un-weighted mean or anterior) * 100. Cohen d = absolute difference/(pooled standard deviation of both weights or axle position).

Abbreviation: BW, body weight.