ASME Open Journal of Engineering
Online journal at:
https://asmedigitalcollection.asme.org/openengineering

Effects of Incremental Changes to Frame Mass on Manual Wheelchair Propulsion Cost

Jacob Misch[1]
Rehabilitation Engineering and Applied Research (REAR) Laboratory,
Georgia Institute of Technology,
801 Atlantic Drive NW,
Atlanta, GA 30332
e-mail: mischjp@gatech.edu

Stephen Sprigle
Rehabilitation Engineering and Applied Research (REAR) Laboratory,
Georgia Institute of Technology,
801 Atlantic Drive NW,
Atlanta, GA 30332
e-mail: stephen.sprigle@design.gatech.edu

The objective of this study was to assess the effects of small, incremental additions to wheelchair frame mass (0 kg, +2 kg, and +4 kg) on the mechanical propulsion characteristics in both straight and curvilinear maneuvers. A robotic propulsion system was used to propel a manual wheelchair over a smooth tiled surface following rectilinear ("Straight") and curvilinear ("Slalom") trajectories. Three unique loading conditions were tested. Propulsion costs and system rolling resistance estimations were empirically collected using the robotic wheelchair tester. Propulsion cost values were equivalent across all loading conditions over the Slalom trajectory. In the Straight trajectory, adding 2 kg on the axle had equivalent propulsion cost to the unloaded configuration. Adding 4 kg on axle was comparable, but not equivalent, to the unloaded configuration with small (≤4.1%) increases in propulsion cost. This study demonstrates that small (0–4 kg) changes to the frame mass have no meaningful impacts on the propulsion characteristics of the manual wheelchair system. Differences in propulsion cost and rolling resistance were detectable but contextually insignificant. [DOI: 10.1115/1.4062696]

Keywords: manual wheelchair, propulsion cost, rolling resistance, energy loss, frame mass, dynamics, energy efficiency, mechanical engineering, mechatronics and electromechanical systems, rehabilitation devices

Introduction

Long-term manual wheelchair (MWC) users face an elevated risk of injury to the shoulders and joints in their upper extremities due to the constant cyclic high loads applied during propulsion [1–3]. These strenuous conditions can cause discomfort and damage to users, though they can be mitigated through improvements in wheelchair configurations such as the shoulder-to-axle distance [4,5], system mass [6,7], weight distribution (WD) [5–7], wheel selection [8–11], and tire inflation [11–13]. Instead of any of these design choices, the Healthcare Common Procedure Coding System (HCPCS) uses wheelchair frame weight to categorize wheelchairs into the classifications of "standard" chairs (weighing ≥ 36 lbs), "lightweight" chairs (34–36 lbs), high-strength lightweight (<34 lbs), or the highly configurable "ultra-lightweight" (≤30 lbs) chairs. The standard wheelchair is often for short-term use, whereas ultra-lightweight wheelchairs are provided to full-time, active wheelchair users. There is an inherent implication, therefore, that the frame mass has some impact on the performance characteristics of the wheelchair. Activities that involve lifting the frame, such as loading or unloading the chair from a vehicle, could be negatively impacted by heavier frames, depending on the physical capabilities of the user. However, it is unclear if small frame mass differences are related to any over-ground performance differences between MWCs.

From a mechanical standpoint, an increase in system mass increases the load on the wheels, which in turn increases rolling resistance of the drive wheels and casters, as clearly shown in several sources ranging from classical tire and vehicle dynamics [14] to more specific wheelchair-related studies [10,15–17]. However, manual wheelchairs are categorized by frame mass differences of only 1–3 kg (2.2–6.6 lbs) which, when distributed over four wheels, result in minimal change to the mechanical behavior of each wheel [17] and the overall system performance [18,19]. Adding weights far away from the drive wheel axle or center of mass increases the required propulsion forces and is generally more physically demanding [18]. This can be attributed to a combination of reasons related to the mechanics of the wheelchair: shifting mass from a larger-radius wheel to a smaller-radius wheel increases the rolling resistance and scrub torque acting on the wheelchair [15,17,19,20], and the rotational inertia of the system about the center of mass increases [21] which corresponds to increases in the system resistance to turning [6,22,23]. However, the weight distribution is not likely to be impacted by marginal, incremental changes in frame mass, and the performance differences seen with large shifts in weight distribution are near-negligible in comparison with other factors such as tire and caster selection [24]. Additionally, small changes in wheelchair mass are minor compared to the mass of the occupant (around 60–110 kg), which easily dominates the overall mass of the system.

Data from human subject studies with frame mass as an experimental variable suggest that adding mass to the wheelchair frame, up to 5 kg [24] or 10 kg [11,25], has no significant effect on the mechanical performance or biomechanical efficiency of the

[1]Corresponding author.

Manuscript received October 2, 2022; final manuscript received May 30, 2023; published online June 23, 2023. Assoc. Editor: Sesh Commuri.

wheelchair under human operation. de Groot et al. designed an experiment to assess the relative contributions of system mass versus tire type on the power output, heart rate, VO2, mechanical efficiency, and propulsion technique of 11 human subjects on a treadmill at a fixed speed of 1.11 m/s. The extra mass did not increase any of the measured outcome variables, whereas simply changing the tire type from pneumatic to solid had a greater impact on physiological measurements compared with adding 10 kg to the frame [11]. Similarly, Bednarczyk and Sanderson added masses in discrete 0 kg, 5 kg, and 10 kg loading conditions but did not find mass to have an impact on the propulsion efforts exerted by the subjects [25]. Sagawa et al. designed a study focused solely on the effect of small, incremental additions of 0, 1, 2, and 5 kg on the propulsion performance of wheelchair users and able-bodied subjects on the physiological parameters, perceived exertion, and performance during the completion of specific tasks. None of the loading conditions had any significant impact on energy expenditure, heart rate, or task completion rate [24]. These studies [11,24,25] suggest that changes to the frame mass by up to 10 kg had no measurable impact on the performance or efficiency of the wheelchair under human operation.

The utilization of human subjects to study the biomechanics of propulsion is a valid approach. However, studies on this topic are limited by small sample sizes [11,24,25], the use of standardized wheelchair configurations without "fitting" test chairs to individual users [4,11,24], and the use of heavy, instrumented wheels [4,11] which can skew the weight distribution and impact rolling resistance of the wheelchair. When studying a mechanical system, a more controlled, sensitive, and repeatable means for imparting propulsive energy is often necessary. Component-level testing may use or emulate a rudimentary inertial load on a wheel moving along linear paths [12,15,17,20] or curvilinear paths [12,17,19,22,23] with coast-down deceleration or bench-top testing techniques. This style of testing can capture the mechanical energy losses between the tire and floor, yet direct frictional measurements of the components cannot fully capture or estimate the propulsion efficiency of the manual wheelchair as a system [26–28]. To this end, a robotic wheelchair propulsion system was created to provide repeatable loading conditions and propulsive inputs across a wide variety of wheelchairs [29], which expanded upon mechanical coast-down deceleration testing [12,15,17,20] by adding human-like propulsion inputs to generate movement and counteract resistive forces.

Objectivity, repeatability, and reliability of the robotic wheelchair propulsion system have been demonstrated in its initial design article [29] and a preliminary study on occupant mass and weight distribution [6], as well as an extensive investigation on how multiple wheelchair design factors affect the mechanical efficiency of tested wheelchairs [7]. In another study, it was shown that increasing the robotic occupant mass by 20 kg increased propulsion costs by 17–27% during highly controlled straight and curvilinear maneuvers [26]. Propulsion costs measured by the robotic system for folding ultra-lightweight wheelchairs traveling over smooth concrete and carpet (14.6 J/m and 22.6 J/m, respectively) [30] are similar to human propulsion characteristics measured at similar speeds and in similar chairs with instrumented pushrims (14.6 J/m and 18.4 J/m for smooth vinyl and 1 mm thickness textured mat, respectively, estimated from reported weight-normalized work per meter and assuming a combined occupant and chair mass of 93 kg) [31]. Other studies with instrumented pushrims reflect similar magnitudes of work during the propulsion cycle [32] and work per meter [33] to the robotic tester results [7] across a range of surface characteristics. It is possible that using this robotic wheelchair propulsion system and the established metric of propulsion cost [7] (i.e., a metric that reflects the vehicular mechanical efficiency in terms of Joules of energy per meter traveled, analogous to automobile gas mileage), the sensitivity of the wheelchair system to small incremental changes to wheelchair frame mass and weight distribution can be investigated with an objective and highly controllable approach. In this way, the effects of frame mass on propulsion efficiency can be more definitively examined, without the limitations and inherent variability of testing with human subjects.

The objective of this study is to assess the effects of incremental additions to the wheelchair frame mass (+2 kg and +4 kg) on the system-level energy loss and propulsion cost in both straight and curvilinear maneuvers using robotic propulsion. The use of the robotic system is an objective approach that is insensitive to user bias and can precisely detect mechanical differences in propulsion efficiency, if any exist, due to the added loads. This adds to the state of knowledge generated from human subject studies [11,24,25] by investigating similar research questions with more repeatable and precise measurements. Cowan et al. added 9.05 kg to wheelchair frames to mimic the weight differential between standard depot chairs and lightweight chair frames [4]. In a similar vein, our small range of added weights (+2 kg and +4 kg) were chosen to represent realistic but relatively incremental frame mass differences that users may see between two contemporary ultra-lightweight frames. For example, in a recent study on aluminum ultra-lightweight folding and rigid frames, the reported masses differed by 4 kg across the six tested wheelchairs using identical wheels, tires, and casters [30]. Within the context of the current study, added loads are placed on the rear axle to reflect differences near the center of gravity, which could represent changes in frame design, material, or construction. It is hypothesized that changes in mechanical propulsion efficiency will be negligible or minimal, as is the case in many of the referenced studies with human subjects [4,11,24,25].

Methods

The anatomical model propulsion system (AMPS; seen in Fig. 1) was used as a manual wheelchair propulsion device. Construction of the AMPS, as described in Ref. [29], was based on the general anthropometric measurements and body segment parameters of an adult male as well as the generalized wheelchair test dummy defined by ISO 7176-11 [34]. Its weight and weight distribution are configurable by changing positions of the metal bar weights within the AMPS. Motors attached to the push-rim of each drive wheel were commanded with a torque-based feedback system to propel the wheelchair with discrete and highly repeatable pushes. Over-running clutches on the motor shafts imparted torque to the rims during the push-phase, and mechanically disconnected the motors from the wheels between pushes during the coast-phase. The data acquisition subsystem (USB-6341, National Instruments Corp.) collected wheel speeds from axle-mounted encoders on the drive wheels and wheel torques were calculated from measurements of each motor armature current.

Fig. 1 The robotic wheelchair-propelling anatomical model propulsion system seated in the Quickie GT wheelchair

Fig. 2 (Left) Unloaded platform to support up to 4 kg of added mass to the drive wheel axle. (Right) Axle platform loaded with 4 kg of weight plates.

The AMPS was loaded on a Quickie GT (Sunrise Medical, LLC) rigid ultra-lightweight aluminum frame. Components were selected based on examination of the industry standard. Narrow solid pyramidal urethane 5 in. × 1 in. casters (Pr1mo HPR-130, Xiamen Lenco Co.) were used as the front wheels. The drive wheels were configured with standard 24 in. × 1–3/8 in. pneumatic tires (Primo Orion, Xiamen Lenco Co.) and metal spoked rims. The tires were inflated to the suggested pressure of 75 psi. Altogether, the mass of the wheelchair with the front footrest, casters, drive wheels, and anti-tip bars was 13.6 kg.

The "occupant" (the AMPS) was configured to 80 kg with a target weight distribution of 70% over the rear axle, the average weight distribution for ultra-lightweight wheelchair users [35]. Mass and weight distribution were measured by the iMachine inertial measurement device [21]. Three loading configurations were used in this study: a "Default" condition without any additional mass; +2 kg added beneath the seat near the rear axle; another +2 kg (+4 kg total) added at the same axle location. A lightweight metal shelf was fastened underneath the seat of the wheelchair to mount weights at the drive wheel axle, shown in Fig. 2.

The default arrangement, termed the "0A" configuration to represent +0 kg on the axle, was measured at a WD of 71.86% over the drive wheel axle. This weight distribution remained consistent after adding weights to the axle, as did the yaw moment of inertia, reflecting how placing the weight on the center of the rear axle increased the system mass without affecting the other measureable loading conditions. Precise weight distribution values of each configuration, as measured by the iMachine [21], are shown in Table 1 alongside measurements of the yaw moment of inertia of each configuration. To put these values in perspective, shifting the 2 kg and 4 kg masses to the footrest instead of the axle would yield WD values of 68.68% and 65.70%, and the yaw moments of inertia would increase to 6.71 kg m^2 and 7.23 kg m^2, respectively.

Data Collection. Three maneuvers were deployed with the AMPS: (1) a passive coast-down deceleration protocol, (2) a robotically propelled seven-push straight-forward trajectory, and (3) a robotically propelled five-push slalom trajectory.

Table 1 Mass and weight distribution of incremental mass configurations

Config. name	Added mass (kg)	System mass (kg)	Weight distribution over rear axle	Yaw moment of inertia (kg m^2)
"0A"	0	93.58	71.86%	6.15
"2A"	2	95.58	72.02%	6.08
"4A"	4	97.58	72.02%	6.11

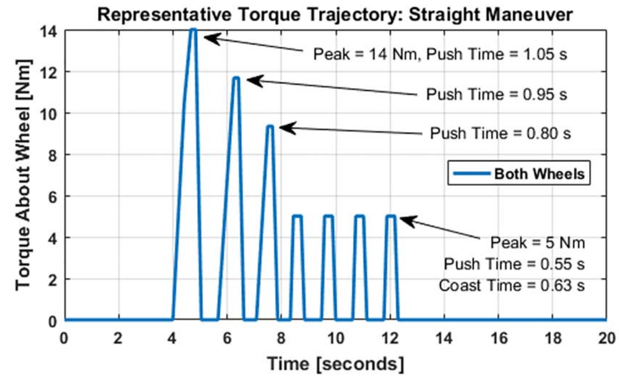

Fig. 3 Torque trajectory for the Straight maneuver

The coast-down deceleration method characterized the rolling resistance of the wheelchair, similar to what has been used in other research on wheelchair energy losses [15,20]. Rolling resistance is an energy loss parameter that helps define the typical coasting behavior of each configuration. Motors were disconnected from the pushrims to mechanically decouple the propulsion system from the wheelchair. The wheelchair was manually pushed via the seat handles to accelerate the chair to approximately 1.0 m/s, and then released to roll down the hallway until it naturally decelerated to a stop. Wheel velocities were recorded by the wheel encoders. Nine trials were taken in each direction (forward and backward, following a "there-and-back" procedure) to account for any slopes or inconsistencies in the floor for each configuration, resulting in 54 total coast-down trials.

AMPS-based maneuvers were deployed over the same surface in a rectilinear ("Straight") trajectory and curvilinear ("Slalom") trajectory. These maneuvers were designed to parametrically mimic the wheel torques used by wheelchair users traveling along common indoor surfaces as measured with instrumented drive wheel systems [4,36–39]. Commanded wheel torques can be seen for the Straight and Slalom maneuvers in Figs. 3 and 4, respectively. The Straight maneuver featured a series of seven pushes: the first three pushes with decreasing torque to accelerate the wheelchair, then four identical pushes to maintain a steady-state travel speed. The Slalom maneuver had five pushes: one small push to propel both wheels forward, align the casters with the chair, and introduce initial velocity, then four stronger pushes on alternating wheels to induce frequent turns. The values for push time and coast time were predetermined for this study and were informed from a variety of sources measuring propulsion torques applied to instrumented pushrims by wheelchair users. A wide range of possible maneuvers can be designed to elicit various movement patterns

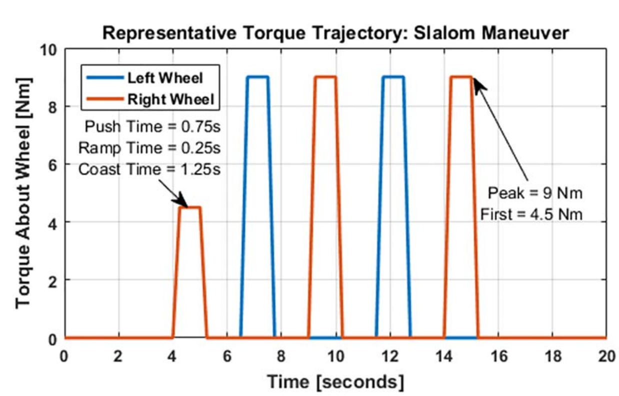

Fig. 4 Torque trajectory for the Slalom maneuver

from the AMPS. Those shown in Figs. 3 and 4 were chosen because they highlight parts of motion (rectilinear and curvilinear) that are influenced by different resistive losses (rolling resistance and scrub torque) and inertial parameters.

Every configuration received the exact same input torques, meaning the AMPS interacted with the wheelchair identically under each loading configuration. Five trials were taken in each direction (forward and backward for Straight; forward-left, forward-right, backward-left, and backward-right for Slalom, with left and right referring to which motor was active during the second push) to normalize over any flooring irregularities. In total, this study required 90 AMPS trials. Data were averaged together by maneuver ($N = 10$ for Straight, $N = 20$ for Slalom) per configuration. Repeatability for each set of trials was assessed. Trial sets with coefficients of variation >10% were retested.

Data Processing. Data from the current-based motor torque sensors, motor encoders, and wheel-mounted encoders were collected at 40 Hz during the over-ground trials. These data were processed in MATLAB. Coast-down data were assessed via linear regression fit as the chair slowed from 0.95 to 0.35 m/s [17]. The velocity data were fit with linear regressions to find the slope of the line representing system deceleration. Coast-down deceleration analysis was expanded to a simple investigation of rolling resistance forces acting on the system, similar to other coast-down test protocols [15,17,20], by multiplying the mean deceleration value by the mass of the system. The product of those terms yields the estimated average force acting against the motion of the chair.

Data Analysis—Propulsion Cost. The main outcome variable of this experiment was propulsion cost, defined as a ratio of the work supplied by the AMPS to the wheelchair over the linear distance traveled by the center of mass. Lower costs are associated with better utilization of the supplied energy and thus better performance. To find the energy supplied to the wheelchair, the wheel torques and wheel speeds are used to calculate rotational power with the equation:

$$\text{Rotational power} = \tau_L \omega_L + \tau_R \omega_R \tag{1}$$

In this equation, τ represents the wheel torque in Newton-meters and ω represents the rotational speed of the wheel in radians per second. Total rotational power, in Watts, is found by summing the individual power inputs of the left and right wheels. Total energy supplied to the wheelchair is found by integrating the time-series power data over the duration of the maneuver:

$$\text{Work in} = \int_{t_i}^{t_f} (\tau_L \omega_L + \tau_R \omega_R) dt \tag{2}$$

The maneuver start time is represented by t_i and the total duration of the maneuver is t_f, in seconds. Similarly, the distance traveled by the center of mass is essentially an integration of the wheel speeds over the duration of the maneuver, derived from a rigid-body analysis of wheelchair motion [40]. Propulsion cost is then found with a simple fraction between the supplied work and the path length or total distance in meters traveled by the center of mass, represented as Δs:

$$\text{Propulsion cost} = \frac{\text{Work in}}{\Delta s} \tag{3}$$

Statistical Analysis. Basic descriptive statistics for each outcome variable were first assessed across the trials for each loading configuration. Cohen's d effect sizes were calculated for each group with the unloaded 0A configuration as the baseline. Effect sizes of $d = 0.2$ indicated that the group means differed by less than 0.2 of the baseline standard deviation and were therefore considered small, $d = 0.5$ were considered medium, and $d = 0.8$ were considered large [41]. Percent change of the means between groups was calculated to tease out the relative contributions of incremental mass and weight placement to the outcome variables. Normality of propulsion cost data were assessed with Kolmogorov–Smirnov goodness-of-fit tests. Both Straight and Slalom propulsion costs were found to have normal distributions ($p > 0.150$).

To assess when the measured wheelchair performances differed by more than a practically important amount, equivalence tests or two-sample two one-sided tests (TOSTs) were conducted within MINITAB (Minitab 18, Minitab Inc.). This style of analysis is technically more appropriate than inferring a lack of a difference when assessed by traditional statistical means [42,43] such as a paired t-test. Operationally, equivalence tests are a two-sided evaluation of differences using confidence intervals (CIs). To inform the practically important bounds of the equivalence analysis on AMPS-based propulsion cost, six relevant studies on biomechanical wheelchair studies were examined [11,13,18,44–46]. Differences in outcome variables across test configurations were used to identify a judicious equivalence interval that reflected differences in propulsion effort obtained across manual wheelchair comparisons. The average differences in outcome variables across these studies were 9.4%. Therefore, the bounds of the equivalence interval used for the AMPS-based propulsion cost were set to $\varepsilon = \pm 5\%$ based on the assumption that mechanical assessment using the AMPS is more precise than biomechanical measurements from human subjects. The TOST analysis compares the 95% CI of the ratio between the test (2A or 4A) and reference (0A) propulsion cost values, relative to the ±5% limits. Configurations are deemed equivalent when the CI falls between the upper and lower bounds, and statistically proven when the p-values for both null hypotheses (ratio $\leq$ lower bound and ratio $\geq$ upper bound) are below the significance level (α), set a priori to 0.05. If the CI extends beyond the equivalence limits, the test configuration cannot be considered equivalent, but are comparable. Only when the CI lies completely outside of the boundaries can test configurations be deemed different from the reference configuration. When the entire CI is greater than the upper equivalence limit, the test configuration can be considered "inferior," as a higher propulsion cost is associated with worse performance.

Results

Coast-Down Decelerations. Table 2 shows that each tested configuration had comparable coast-down deceleration results. Deceleration effect sizes are small to moderate for all configurations. Combining tests from the two directions (forward and backward) yields values most reflective of the chair performance over the particular surface by normalizing for any slopes or inconsistencies in the floor, but also caused high standard deviations of around 18% of the mean, rather than the 1–3% seen when assessing each direction separately. Despite this, the combined datasets were used to prevent any biases due to interactions between masses and surface irregularities.

Propulsion Costs. Basic descriptive statistics for the propulsion cost data are shown in Table 3. Effect sizes mostly reflect the high reliability (i.e., low variances) of the measurements, as seen in prior studies with the AMPS [7]. Cost values were generally tightly grouped with small to negligible percent changes between means.

The equivalence test results of propulsion costs were plotted in Fig. 5 to visualize the propulsion cost distributions. 95% confidence intervals are shown for each propulsion cost ratio. The equivalence threshold (±5% of the mean 0A cost) is shown as a shaded region.

The 0A and 2A propulsion costs were statistically equivalent in both maneuvers. 4A was comparable but not equivalent in Straight motion, and equivalent for Slalom. Most importantly, neither configuration was significantly inferior to 0A.

Table 2 Descriptive statistics of coast-down results

		0A	2A	4A
Mean decel. (m/s^2)	Mean (±std. dev.)	−0.0596 (±0.0103)	−0.0572 (±0.0095)	−0.0589 (±0.0106)
	Effect size	–	0.24	0.07
	% Change	–	−4.0%	−1.2%
Mean force (N)	Mean (±std. dev.)	5.58 (±0.97)	5.47 (±0.91)	5.75 (±1.03)
	Effect size	–	0.12	0.17
	% Change	–	−2.0%	+3.1%

Table 3 Descriptive statistics for propulsion cost in the Straight maneuver

		0A	2A	4A
Straight cost (J/m)	Mean (±std. dev.)	8.89 (±0.53)	8.95 (±0.37)	9.25 (±0.27)
	Effect size	–	0.14	0.87
	% Change	–	+0.7%	+4.1%
	Ratio of group/0A (95% CI)	–	1.01 (0.97, 1.05)	1.04 (1.00, 1.08)
	p-Values			
	Ratio ≤ lower bound	–	0.011	0.000
	Ratio ≥ upper bound		0.045	0.346
Slalom cost (J/m)	Mean (±std. dev.)	9.98 (±0.18)	10.11 (±0.20)	10.10 (±0.17)
	Effect size	–	0.72	0.67
	% Change	–	+1.3%	+1.2%
	Ratio of group/0A (95% CI)	–	1.01 (1.00, 1.02)	1.01 (1.00, 1.02)
	p-Values			
	Ratio ≤ lower bound	–	0.000	0.000
	Ratio ≥ upper bound		0.000	0.000

Note: If the greater of the two *p*-values is ≤0.05, groups have equivalent costs.

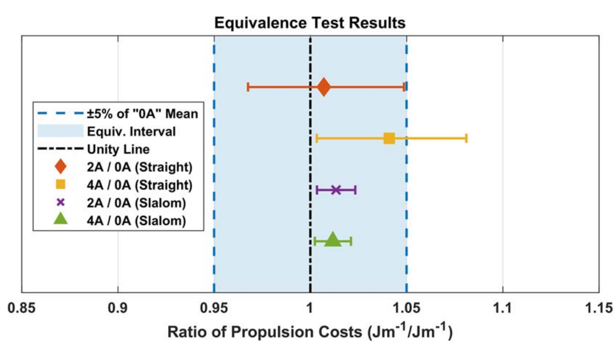

Fig. 5 Overview of equivalence test results for propulsion costs in the Straight maneuver

Discussion

The objective of this study was to assess the influence of incremental changes in frame mass on the system-level performance of the wheelchair. This is the first study to address the research topic using an objective, repeatable robotic propulsion system, ultimately supporting the findings of human subject studies [11,24,25]. During the Straight trajectory, the propulsion cost after adding 2 kg to the frame was equivalent to the default configuration, and adding 4 kg was comparable, though not equivalent. The Slalom propulsion costs of 2A and 4A were both equivalent to the default. Changes in system-level performance only began to appear at +4 kg, and even with the large effect size, the percent change was small. Addition of 1 or 2 kg to a frame should not contribute any meaningful performance differences, with all other factors (tires, weight distribution, and wheelbase) held equal. This result is consistent with human studies which did not find any physiological or biomechanical effects after small mass increases [11,24,25].

Propulsion costs during maneuvers can be used to estimate the added work that might be required throughout a day across configurations. Consider that the average wheelchair user travels between a median of 1.4 km [47] to 1.6 km [48] per day. If 1000 m (i.e., the majority of daily travel) of that motion comprises straight bouts of mobility over smooth surfaces like linoleum tile, the total cost in the 0A configuration would be 8890 J of energy versus 9250 J of energy in the 4A configuration. That 360 J difference is the energetic equivalent of eight full bicep curls with a 5 kg dumbbell, or simply propelling the 4A-configured chair over an extra 40 m per day. This amount of energy is minimal in comparison to other factors, such as the ∼50% propulsion cost increase that was observed when swapping the rear tires from pneumatics to solids [7]. Surprisingly, the Slalom maneuver did not show much difference in cost across configurations despite evidence that wheelchairs with higher system masses require higher torques to propel [6] and incur greater propulsion costs [7] than lighter counterparts. It is believed that the small cornering radii on the Slalom trajectory may have contributed to the similar propulsion cost values, though it is unlikely that the centrally loaded 2 kg or 4 kg would have much impact on propulsion costs in any case. For future studies, it may be more beneficial to test a more rigorous half-turn slalom trajectory rather than just providing alternating wheel torques to the chair.

Energy supplied to the wheelchair is primarily "lost" through the interactions (rolling and scrubbing) between the tires and the ground. The amount of energy loss that occurs per tire is dependent on the tire type, the material properties, and geometric shape of the tire, as well as the load supported by the tire, determined by system mass and weight distribution of the wheelchair. System mass determines the total load distributed over all four wheels, and weight distribution describes how much of the total load is supported at the casters versus at the drive wheels. In general, drive wheels generally exhibit lower energy losses under greater loads than the small front casters due to their substantially larger diameters [7,10,15–17,23]. For example, adding 5 kg to each wheel would increase rolling resistance forces by ≥39% per caster and ≥21% per drive wheel [17]. However, 5 kg per wheel amounts to 20 kg of mass added to the system. In contrast, incremental system mass changes of only 2 or 4 kg would have minimal changes in load per wheel, with much smaller effects on the energy losses.

Energy loss assessment through coast-down testing is a simple, quick, and objectively better way to characterize wheelchairs *by performance* rather than the current approach of categorizing by frame mass. The results presented here illustrate that the very small weight differences which define HCPCS categories do not result in higher propulsion cost. In distinction, the wheelchair coast-down tests used in this study were deployed to measure the energy losses for each configuration. Frame mass was demonstrably insignificant to energy losses at the system level. Rolling resistance forces for 2A and 4A were comparable to 0A, with small effect sizes and percent changes. Additionally, these values were highly correlated with Straight propulsion cost values (Pearson correlation coefficient: $r = 0.912$). As in Ref. [26], energy loss values are powerful predictors of propulsion cost. Use of loaded wheelchair coast-down tests could empower manufacturers and clinicians to assess wheelchair performance in a valid manner.

The HCPCS code uses differences in frame mass as a primary means of categorizing manual wheelchairs and defining eligibility for acquiring one. By focusing on the mass or weight of the wheelchair frame, one may be ignoring more important factors. Specifically, prior work has identified that tires, casters, and weight distribution have the most influence on propulsion cost [7,15,17,19,20,26,49]. A recent study documented a 41% decrease in propulsion cost by upgrading the casters and tires on a K0004 wheelchair [27]. Casters, drive wheels, and weight distribution are more influential because they are directly responsible for the energy losses of the system through rolling resistance and scrub torque. A more valid means to categorized manual wheelchairs would incorporate a simple coast-down test across different configurations (frame adjustments, wheel and tire selections) to determine a measure of system energy losses.

Limitations

Human wheelchair propulsion involves biomechanical factors as well as mechanical factors. Users exhibit movement of the arms and the trunk during propulsion, which may influence the weight distribution and mechanics of the wheelchair system in motion, and expends biomechanical energy; this behavior cannot be replicated by the AMPS. However, human subject testing should generally be attempted after meaningful *mechanical* performance differences are discovered from controlled mechanical tests. The results presented here do not show meaningful propulsion cost or energy loss differences between frame masses.

The rectilinear and curvilinear maneuvers deployed by the AMPS were representative of short bouts of motion that are common trajectories used in everyday mobility, however wheelchair users have infinite options of trajectories during motion. The maneuvers used in this study were meant to demonstrate the most common parts of motion, including inertial parameters such as changes in momentum, and energy losses as the wheels and casters roll and scrub during turns. The components used in this study were selected from common no-cost options offered on ultra-lightweight frames. Performance may vary between wheels and tires based on tire type, material, and size. Components with similar rolling resistance and scrub parameters will exhibit similar performances under the same loading conditions over similar surfaces. Additionally, the results shown in this study exhibited large standard deviations, indicating the possibility that several surface irregularities existed along the test track. The presence of a slope or grade could cause disparate performances of chairs with varying inertial parameters, if only driven in one direction. While this reflects the conditions of many real-world environments, it necessitates the "there and back"-style procedure used during the coast-down deceleration and AMPS propulsion trials.

Conclusion

This study presents an objective comparison of wheelchair performance characteristics between loading configurations with varying incremental frame mass additions and weight distributions. Incremental additions of 2 kg and 4 kg to the wheelchair frame mass had no meaningful impact on propulsion costs in rectilinear and curvilinear maneuvers, based on defined equivalence intervals. An energy loss assessment technique measuring coast-down decelerations of the loaded wheelchair provided simple and efficient measurements that were highly correlated with propulsion cost results. Based on a review of related literature, biomechanical measurements from human subjects are not sensitive enough to detect significant differences between loading configurations with much larger mass changes (≤10 kg). Obsessing over frame mass may cause manufacturers, clinicians, users, and policy-makers to ignore other more important factors. Rather than focus on the frame mass of the wheelchair, it is suggested that future studies investigate the interactions between system mass, weight distribution, and the energy loss parameters of the wheels and tires offered on each wheelchair frame. The described coast-down protocol is recommended for assessing the energy losses of any given wheelchair configuration, as it does not require expensive hardware and can be conducted both in a laboratory setting with dummy-loaded chairs or in a clinic with the individual user.

Funding Data

- National Institute on Disability, Independent Living, and Rehabilitation Research (NIDILRR) (Grant Nos. 90RE5000 and 90IFRE0036; Funder ID: 10.13039/100009157).
- NIDILRR is a Center within the United States Department of Health and Human Services (HHS) Administration for Community Living (ACL). The contents of this article do not necessarily reflect the views of the Department of Health and Human Services.

Conflict of Interest

There are no conflicts of interest. This article does not include research in which human participants were involved. Informed consent not applicable. This article does not include any research in which animal participants were involved.

Data Availability Statement

The data and information that support the findings of this article are freely available.[2]

References

[1] Alm, M., Saraste, H., and Norrbrink, C., 2008, "Shoulder Pain in Persons With Thoracic Spinal Cord Injury: Prevalence and Characteristics," J. Rehabil. Med., **40**(4), pp. 277–283.

[2] Brose, S. W., Boninger, M. L., Fullerton, B., McCann, T., Collinger, J. L., Impink, B. G., and Dyson-Hudson, T. A., 2008, "Shoulder Ultrasound Abnormalities, Physical Examination Findings, and Pain in Manual Wheelchair Users With Spinal Cord Injury," Arch. Phys. Med. Rehabil., **89**(11), pp. 2086–2093.

[3] Mozingo, J. D., Akbari-Shandiz, M., Murthy, N. S., Van Straaten, M. G., Schueler, B. A., Holmes, D. R., McCollough, C. H., and Zhao, K. D., 2020, "Shoulder Mechanical Impingement Risk Associated With Manual Wheelchair Tasks in Individuals With Spinal Cord Injury," Clin. Biomech., **71**(3), pp. 221–229.

[4] Cowan, R. E., Nash, M. S., Collinger, J. L., Koontz, A. M., and Boninger, M. L., 2009, "Impact of Surface Type, Wheelchair Weight, and Axle Position on Wheelchair Propulsion by Novice Older Adults," Arch. Phys. Med. Rehabil., **90**(7), pp. 1076–1083.

[5] Brubaker, C. E., 1986, "Wheelchair Prescription: An Analysis of Factors That Affect Mobility and Performance," J. Rehabil. Res. Dev., **23**(4), pp. 19–26.

[6] Sprigle, S., and Huang, M., 2015, "Impact of Mass and Weight Distribution on Manual Wheelchair Propulsion Torque," Assist. Technol., **27**(4), pp. 226–235; quiz 236–237.

[2]http://hdl.handle.net/1853/64159

[7] Sprigle, S., and Huang, M., 2020, "Manual Wheelchair Propulsion Cost Across Different Components and Configurations During Straight and Turning Maneuvers," J. Rehabil. Assist. Technol. Eng., **7**(1), pp. 1–14.
[8] Bascou, J., Fallot, C., Pillet, H., Hybois, S., Hybois, S., Lavaste, F., and Sauret, C., 2017, "Assessment of Power Losses Due to Ground Contact Forces During Usual Manual Wheelchair Movements," Comput. Meth. Biomech. Biomed. Eng., **20**(Suppl. 1), pp. 7–8.
[9] Chan, F. H. N., Eshraghi, M., Alhazmi, M. A., and Sawatzky, B. J., 2018, "The Effect of Caster Types on Global Rolling Resistance in Manual Wheelchairs on Indoor and Outdoor Surfaces," Assist. Technol., **30**(4), pp. 176–182.
[10] Hoffman, M. D., Millet, G. Y., Hoch, A. Z., and Candau, R. B., 2003, "Assessment of Wheelchair Drag Resistance Using a Coasting Deceleration Technique," Am. J. Phys. Med. Rehabil., **82**(11), pp. 880–889; quiz 890–892.
[11] de Groot, S., Vegter, R. J., and van der Woude, L. H., 2013, "Effect of Wheelchair Mass, Tire Type and Tire Pressure on Physical Strain and Wheelchair Propulsion Technique," Med. Eng. Phys., **35**(10), pp. 1476–1482.
[12] Lin, J. T., Huang, M., and Sprigle, S., 2015, "Evaluation of Wheelchair Resistive Forces During Straight and Turning Trajectories Across Different Wheelchair Configurations Using Free-Wheeling Coast-Down Test," J. Rehabil. Res. Dev., **52**(7), pp. 763–774.
[13] Sawatzky, B. J., Kim, W. O., and Denison, I., 2004, "The Ergonomics of Different Tyres and Tyre Pressure During Wheelchair Propulsion," Ergonomics, **47**(14), pp. 1475–1483.
[14] Pacejka, H., 2005, *Tire and Vehicle Dynamics*, Elsevier, New York.
[15] Bascou, J., Sauret, C., Pillet, H., Vaslin, P., Thoreux, P., and Lavaste, F., 2013, "A Method for the Field Assessment of Rolling Resistance Properties of Manual Wheelchairs," Comput. Methods Biomech. Biomed. Eng., **16**(4), pp. 381–391.
[16] van der Woude, L. H., Geurts, C., Winkelman, H., and Veeger, H. E., 2003, "Measurement of Wheelchair Rolling Resistance With a Handle Bar Push Technique," J. Med. Eng. Technol., **27**(6), pp. 249–258.
[17] Sprigle, S., Huang, M., and Misch, J., 2022, "Measurement of Rolling Resistance and Scrub Torque of Manual Wheelchair Drive Wheels and Casters," Assist. Technol., **34**(1), pp. 91–103.
[18] Alcoléa, V., Medola, F. O., da Silva Bertolaccini, G., and Sandnes, F. E., 2020, "Effect of Added Mass Location on Manual Wheelchair Propulsion Forces," Human Systems Engineering and Design II, **1**(1), pp. 747–753.
[19] Fallot, C., Bascou, J., Pillet, H., and Sauret, C., 2021, "Manual Wheelchair's Turning Resistance: Swivelling Resistance Parameters of Front and Rear Wheels on Different Surfaces," Disabil. Rehabil. Assist. Technol., **16**(3), pp. 324–331.
[20] Sauret, C., Bascou, J., de Saint Remy, N., Pillet, H., Vaslin, P., and Lavaste, F., 2012, "Assessment of Field Rolling Resistance of Manual Wheelchairs," J. Rehabil. Res. Dev., **49**(1), pp. 63–74.
[21] Eicholtz, M. R., Caspall, J. J., Dao, P. V., Sprigle, S., and Ferri, A., 2012, "Test Method for Empirically Determining Inertial Properties of Manual Wheelchairs," J. Rehabil. Res. Dev., **49**(1), pp. 51–62.
[22] Caspall, J. J., Seligsohn, E., Dao, P. V., and Sprigle, S., 2013, "Changes in Inertia and Effect on Turning Effort Across Different Wheelchair Configurations," J. Rehabil. Res. Dev., **50**(10), pp. 1353–1362.
[23] Bascou, J., Sauret, C., Villa, C., Lavaste, F., and Pillet, H., 2015, "Measurement of Wheelchair Adjustment Effects on Turning Deceleration," Comput. Methods Biomech. Biomed. Eng., **18**(Suppl. 1), pp. 1882–1883.
[24] Sagawa, Y., Jr., Watelain, E., Lepoutre, F. X., and Thevenon, A., 2010, "Effects of Wheelchair Mass on the Physiologic Responses, Perception of Exertion, and Performance During Various Simulated Daily Tasks," Arch. Phys. Med. Rehabil., **91**(8), pp. 1248–1254.
[25] Bednarczyk, J. H., and Sanderson, D. J., 1995, "Limitations of Kinematics in the Assessment of Wheelchair Propulsion in Adults and Children With Spinal Cord Injury," Phys. Ther., **75**(4), pp. 281–289.
[26] Misch, J., Huang, M., and Sprigle, S., 2020, "Modeling Manual Wheelchair Propulsion Cost During Straight and Curvilinear Trajectories," PLoS One, **15**(6), p. e0234742.
[27] Misch, J., and Sprigle, S., 2021, "Effects of Wheels and Tires on High-Strength Lightweight Wheelchair Propulsion Cost Using a Robotic Wheelchair Tester," Disabil. Rehabil. Assist. Technol., **1**(1), pp. 1–11.
[28] Bascou, J., Hybois, S., Pillet, H., Lavaste, F., and Sauret, C., 2017, "Mechanical Simulations as a Tool for Assessing the Influence of Wheelchair Settings on the Propulsion Efficiency," Ann. Phys. Rehabil. Med., **60**(Suppl. 1), p. e92.
[29] Liles, H., Huang, M., Caspall, J., and Sprigle, S., 2015, "Design of a Robotic System to Measure Propulsion Work of Over-Ground Wheelchair Maneuvers," IEEE Trans. Neural Syst. Rehabil. Eng., **23**(6), pp. 983–991.
[30] Misch, J., and Sprigle, S., 2022, "Propulsion Cost Changes of Ultra-Lightweight Manual Wheelchairs After One Year of Simulated Use," ASME Open J. Eng., **1**, p. 011047.
[31] Chenier, F., and Aissaoui, R., 2014, "Effect of Wheelchair Frame Material on Users' Mechanical Work and Transmitted Vibration," Biomed. Res. Int., **2014**(1), p. 609369.
[32] Hurd, W. J., Morrow, M. M., Kaufman, K. R., and An, K. N., 2009, "Wheelchair Propulsion Demands During Outdoor Community Ambulation," J. Electromyogr. Kinesiol., **19**(5), pp. 942–947.
[33] Chesney, D. A., and Axelson, P. W., 1996, "Preliminary Test Method for the Determination of Surface Firmness," IEEE Trans. Rehabil. Eng., **4**(3), pp. 182–187.
[34] International Standards Organization, 2012, "ISO 7176 Wheelchair Standards—Section 11: Test Dummies.
[35] Lin, J. T., and Sprigle, S., 2020, "The Influence of Operator and Wheelchair Factors on Wheelchair Propulsion Effort," Disabil. Rehabil. Assist. Technol., **15**(3), pp. 328–335.
[36] Asato, K. T., Cooper, R. A., Robertson, R. N., and Ster, J. F., 1993, "SMARTWheels: Development and Testing of a System for Measuring Manual Wheelchair Propulsion Dynamics," IEEE Trans. Biomed. Eng., **40**(12), pp. 1320–1324.
[37] Koontz, A. M., Cooper, R. A., Boninger, M. L., Yang, Y., Impink, B. G., and van der Woude, L. H. V., 2005, "A Kinetic Analysis of Manual Wheelchair Propulsion During Start-Up on Select Indoor and Outdoor Surfaces," J. Rehabil. Res. Dev., **42**(4), pp. 447–458.
[38] Koontz, A. M., Roche, B. M., Collinger, J. L., Cooper, R. A., and Boninger, M. L., 2009, "Manual Wheelchair Propulsion Patterns on Natural Surfaces During Start-Up Propulsion," Arch. Phys. Med. Rehabil., **90**(11), pp. 1916–1923.
[39] Cowan, R. E., Boninger, M. L., Sawatzky, B. J., Mazoyer, B. D., and Cooper, R. A., 2008, "Preliminary Outcomes of the SmartWheel Users' Group Database: A Proposed Framework for Clinicians to Objectively Evaluate Manual Wheelchair Propulsion," Arch. Phys. Med. Rehabil., **89**(2), pp. 260–268.
[40] Medola, F. O., Dao, P. V., Caspall, J. J., and Sprigle, S., 2014, "Partitioning Kinetic Energy During Freewheeling Wheelchair Maneuvers," IEEE Trans. Neural Syst. Rehabil. Eng., **22**(2), pp. 326–333.
[41] Cohen, J., 1988, *Statistical Power Analysis for the Behavioral Sciences*, 2nd ed., Lawrence Earlbaum Associates, Hillsdale, NJ.
[42] Wellek, S., 2010, *Testing Statistical Hypotheses of Equivalence and Noninferiority*, 2nd ed., CRC Press, Boca Raton, FL, p. 432.
[43] Snow, M. P., Reising, J. M., Barry, T. P., and Hartsock, D. C., 2016, "Comparing New Designs With Baselines," Ergon. Des., **7**(4), pp. 28–33.
[44] Lui, J., MacGillivray, M. K., Sheel, A. W., Jeyasurya, J., Sadeghi, M., and Sawatzky, B. J., 2013, "Mechanical Efficiency of Two Commercial Lever-Propulsion Mechanisms for Manual Wheelchair Locomotion," J. Rehabil. Res. Dev., **50**(10), pp. 1363–1372.
[45] Pavlidou, E., Kloosterman, M. G., Buurke, J. H., Rietman, J. S., and Janssen, T. W., 2015, "Rolling Resistance and Propulsion Efficiency of Manual and Power-Assisted Wheelchairs," Med. Eng. Phys., **37**(11), pp. 1105–1110.
[46] Hilbers, P. A., and White, T. P., 1987, "Effects of Wheelchair Design on Metabolic and Heart Rate Responses During Propulsion by Persons With Paraplegia," Phys. Ther., **67**(9), pp. 1355–1358.
[47] Sonenblum, S. E., and Sprigle, S., 2017, "Wheelchair Use in Ultra-Lightweight Wheelchair Users," Disabil. Rehabil. Assist. Technol., **12**(4), pp. 396–401.
[48] Sonenblum, S. E., Sprigle, S., and Lopez, R. A., 2012, "Manual Wheelchair Use: Bouts of Mobility in Everyday Life," Rehabil. Res. Pract., **2012**(1), pp. 1–7.
[49] Kauzlarich, J. J., and Thacker, J. G., 1985, "Wheelchair Tire Rolling Resistance and Fatigue," J. Rehabil. Res. Dev., **22**(3), pp. 25–41.