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Comparison Between Overground and Dynamometer Manual Wheelchair Propulsion

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Abstract

Laboratory-based simulators afford many advantages for studying physiology and biomechanics; however, they may not perfectly mimic wheelchair propulsion over natural surfaces. The goal of this study was to compare kinetic and temporal parameters between propulsion overground on a tile surface and on a dynamometer. Twenty-four experienced manual wheelchair users propelled at a self-selected speed on smooth, level tile and a dynamometer while kinetic data were collected using an instrumented wheel. A Pearson correlation test was used to examine the relationship between propulsion variables obtained on the dynamometer and the overground condition. Ensemble resultant force and moment curves were compared using cross-correlation and qualitative analysis of curve shape. User biomechanics were correlated (R ranging from 0.41 to 0.83) between surfaces. Overall, findings suggest that although the dynamometer does not perfectly emulate overground propulsion, wheelchair users were consistent with the direction and amount of force applied, the time peak force was reached, push angle, and their stroke frequency between conditions.

Keywords

biomechanics; kinetics; spinal cord injury

Stationary propulsion is conveniently used to study manual wheelchair propulsion physiology and biomechanics. Simulators are ideal for controlling experimental variables in propulsion studies such as speed, power output, and resistance and for the collection of

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kinetic, kinematic, electromyography, and metabolic data. Researchers have gone to great lengths to design systems that realistically mimic the inertial effects of propelling overground and a variety of terrains (Aissaoui et al., 2002; Mulroy et al., 1996; van der Woude et al., 2001; Veeger et al., 1992); however, because these systems are stationary, they do not provide the same visual cues that would be present when pushing in a free-living environment and they minimize the need for postural adjustments to maintain stability of the wheelchair since the chair is physically tethered (Vanlandewijck et al., 2001; Veeger et al., 1992). Recent studies on overground propulsion indicate that propulsion kinetics and techniques vary widely depending on the types and nature of real-world surfaces commonly traversed in daily living (Hurd et al., 2008b, 2008c; Koontz et al., 2005). However, no studies to date have directly compared wheelchair propulsion on a simulator to propulsion overground. This type of study is important to better understand the generalizability of data collected exclusively with simulators in laboratory settings.

Simulators commonly used to study propulsion include ergometers, dynamometers, and wheelchair treadmills. Ergometers and dynamometers are systems that measure work and power, are capable of applying a load, and may or may not have the ability to add power to the system to simulate friction losses (DiGiovine et al., 2001; Mulroy et al., 1996; Niesing et al., 1990; Veeger et al., 1992). If the system is not computer controlled (e.g., equipped with an electronic control system), simulation of inertia to match the user/wheelchair system is accomplished with weights and flywheels (Newsam et al., 1999; Mulroy et al., 1996; Rodgers et al., 2000) and simple friction or electric brakes for adjusting propulsion difficulty (Kotajarvi et al., 2006). Motor driven treadmills change workload by changing the angle of inclination or adding resistance to the chair by use of a pulley system (van der Woude et al., 2001). With treadmills the wheelchair is not held stationary, and thus the inertial properties of the individual during movement are preserved (Richter et al., 2007); however, they are not truly representative of real-world propulsion since the user is forced to keep pace with the constant velocity setting of a motor-driven treadmill. With most ergometers, to obtain pushrim kinetics requires users to transfer into a test chair that is equipped with or attached to the instrumentation (Mulroy et al., 1996; Newsam et al., 1999; Rodgers et al., 2000; Veeger et al., 1992). Dynamometers and treadmills allow for testing the individual in their own chair and the pushrim kinetics are collected from instrumented wheels (DiGiovine et al., 2001; Koontz et al., 2007; Kotajarvi et al., 2006; van der Woude et al., 2001). Despite the differences among simulators, stationary propulsion seeks to mimic overground propulsion as realistically as possible.

Gait has been studied extensively on treadmills and overground. Research comparing ambulation on a simulator to overground has shown differences in muscle activation patterns, kinematics, and joint moments (Lee & Hidler, 2007). These differences were attributed in part to the absence of such visual cues as terrain, walls, and doorways during treadmill walking (Warren et al., 2001; Van Ingen Schenau, 1980). Because surroundings do not move with respect to the subject, there may be changes in kinematics and energy consumption, as visual feedback is important to maintaining balance and stability (Van Ingen Schenau, 1980). It is possible that like gait, propulsion may also differ between simulators and overground conditions.

Few studies have compared propulsion on simulators to surfaces in the natural environment. Hurd et al. (2008a) found no significant differences in the bilateral asymmetry of kinetic and temporal propulsion data between a dynamometer that simulated level ground and any of the indoor community and laboratory terrain tested including flat, smooth tile, low-pile carpet, and a ramp. In a preliminary study, we found stroke-to-stroke variability was more pronounced on indoor surfaces (i.e., flat, smooth tile, ramp, carpet) compared with a dynamometer (Timcho et al., 2009). These studies used discrete point statistics or averages to compare biomechanical differences across surfaces and as a result changes in the underlying stroke pattern may have been overlooked (Stergiou, 2004). The goals of this study were to (1) investigate the relationship between key kinetic and temporal discrete point variables and (2) compare qualitative and quantitative characteristics of the force and moment curves between a dynamometer and a level, smooth surface. We hypothesized that key variables would be highly correlated and stroke patterns would be similar between the two conditions.

Methods

Subjects

Subjects were recruited from several IRB-approved research registries as well as through flyers and word of mouth. Subjects signed informed consent and inclusion criteria that required participants to be over the age of 18 and use a manual wheelchair as their primary means of mobility (>80% of locomotion).

Experimental Setup

The dynamometer used in this study was designed and fabricated in-house and was modeled after the one described in DiGiovine et al. (2001) without the computer-controlled motors and torque sensors. The dynamometer consists of two independent, steel tubular rollers, one for each wheel, supported by pillow-block bearings mounted to steel channel. While it is equipped with tachometers to provide visual speed and direction feedback, participants received no feedback when propelling on the dynamometer in this study.

Two instrumented wheels (SmartWheel, Three Rivers Holdings, Inc., Mesa, AZ) were attached to the individual's own wheelchair. The SmartWheel records three-dimensional (3D) pushrim kinetics and temporal data and does not alter wheelchair setup (Cooper et al., 1997). The SmartWheel is 60.7 cm (24 inches) in diameter, has a urethane solid-foam tire, a 1.27 cm (1/2 inch) round anodized pushrim, and weighs 48.5 N (10.9 lbs). Two wheels were used to maintain weight distribution. Adding two SmartWheels in place of the regular wheels adds approximately 50 N to the total weight of the wheelchair.

Propulsion Testing

Trials on the dynamometer and overground were performed in random order. Each subject's wheelchair was attached to the dynamometer with a four-point securement system. Subjects were allowed to acclimate to the dynamometer for at least five minutes and afterward started from a rest position and propelled up to a self-selected speed for one minute. The overground testing consisted of propelling down a long, level, straight hallway on tile

flooring (e.g., industrial type found in hospitals). The subject started from rest and propelled up to a self-selected speed for a distance of 40 m. For both conditions, data were collected from the SmartWheel corresponding to the subject's nondominant arm as previous studies found biomechanical variables to be similar between sides during straightaway propulsion (Hurd et al., 2008a, Boninger et al., 2004). SmartWheel data were collected at 240 Hz for the entire duration of each trial. Subject weight was self-reported.

Data Analysis

Kinetic data were filtered using an 8th-order Butterworth low-pass filter, zero-lag, and a 20 Hz cutoff frequency (Cooper et al., 2009). The propulsion cycle was divided into two phases: nonpropulsive and propulsive according to the methods described in Kwarciak et al. (2009). The nonpropulsive phase includes initial contact and release and was excluded from further analysis. Consistent with Cowan et al. (2008) the first three strokes of each trial were considered start-up strokes and were excluded from further analysis. The peak resultant force was calculated for each propulsive phase and for all remaining strokes. An average and standard deviation for all strokes was calculated and individual strokes which were not within two standard deviations of the mean were characterized as extraneous corrective forces (e.g., steering forces), miss-hits, or braking forces rather than propulsive forces and were excluded from further analysis. Output variables were calculated for each "acceptable" stroke and then an average over five strokes for each trial was computed. The 3D forces recorded by the SmartWheel were transformed into a pushrim coordinate system (Cooper et al., 1997). Output variables included the maximum resultant force (F_R), radial force (F_r), tangential force (F_t), medial-lateral force (F_z), moment about the hub (M_z), push angle, stroke frequency, average wheel velocity, and average mechanical effective force (mef). The term mef is the contribution of tangential force to overall resultant force (F_t^2/F_R^2) (Cooper et al., 1997). Normality was assessed by the skewness of the distribution. Variables that fell within a skewness range of -2 to $+2$ were considered normally distributed. Depending on the nature of the distribution either a Pearson correlation test (normally distributed variables) or a Spearman rho test (nonnormally distributed variables) was performed to examine the relationship between propulsion variables obtained on the dynamometer and the overground condition. Paired comparison tests were not applied due to inertial differences that may have existed between the two conditions as the dynamometer was not adjusted to match the inertial characteristics of each individual who participated in the study. Linear regression was used to determine if dynamometer variables and subject weight were significant predictors of propulsion kinetics overground. The M_z and F_R curves for each stroke were body weight and time normalized to 100% and ensemble curves were created. The ensemble curve was determined by computing a point-by-point average of the five strokes which were previously identified as representative of the user's inherent stroke technique. From the ensemble curve, the peak rate of rise and descent for the first and last third of the push phase was identified as well as the percent of push phase when the peak value was reached. These data were also examined for normality and compared between conditions using either Pearson's or Spearman's rho correlation test. A cross correlation was performed to determine the degree of similarity between normalized M_z and F_R during the push phase between the two conditions for each subject (Stergiou, 2004). The shapes of the normalized

Mz and F_R were also evaluated qualitatively. Three distinct curve patterns were identified and the distribution of each across the two conditions was evaluated using a chi square test.

Results

Twenty-four subjects participated in the study. The majority of the subjects had a spinal cord injury (7 cervical, 13 thoracic, 2 lumbar) while one participant had spina bifida and another had a double below-the-knee amputation. Subject demographics were 21 males and 3 females with an average age of 40 (*SD* 13) years, weight of 792 (*SD* 189) newtons, and average duration of wheelchair use of 17 (*SD* 11) years.

When we applied our criteria for identifying extraneous strokes, we found a total of four strokes to be unacceptable: one stroke each from two subjects on the dynamometer and one stroke each from two subjects on tile. All variables met the skewness criteria for normality except for the medial-lateral force component overground. Individuals produced larger peak force and moment on the dynamometer compared with tile (Table 1) and all of these variables were positively correlated for the two surfaces except for peak medial-lateral force. A secondary analysis of the average medial-lateral forces, however, indicated significant correlation between the two surfaces. The average F_z force produced on the dynamometer was 7.4 (*SD* 10.4) N compared with 6.2 (9.9 *SD*) N overground ($r = .534$; $p = .007$). Self-selected velocity for tile was higher than for the dynamometer and was not correlated. Mechanical efficiency, push angle, and frequency were positively correlated between conditions. Subject body weight was significantly correlated with all maximum forces and maximum moment about the hub overground with the exception of medial-lateral force (R ranging from 0.427 to 0.782, $p < .01$). The same was true for dynamometer with the additional exception that maximum radial force was not correlated with body weight (R ranging from 0.467 to 0.623, $p < .01$).

Peak resultant force on tile was selected as a key dependent measure for our linear regression model as this kinetic parameter has been previously correlated to wrist injuries in wheelchair users (Boninger et al., 2005). Two predictor variables were considered based on the sample size and the correlation findings in Table 1. The predictor variables tested in separate models using the enter approach included peak F_R on the dynamometer and body weight, peak F_R and push angle on the dynamometer, and peak F_R and stroke frequency on the dynamometer. Entering two force or torque predictor variables together into the model was not considered an option due to their high level of interdependency. We found that dynamometer maximum resultant force and body weight best predicted maximum resultant force on tile. The regression model equation for tile resulted in an R value of 0.826, p -value of <0.001 , as follows.

$$\text{Tile Resultant Force} = -14.671 + (0.487) (\text{Dynamometer Resultant Force}) + (0.231) (\text{Subject Body Weight})$$

The normalized Mz curves were positively correlated (R ranging from 0.74 to 0.99, $p < .001$) between conditions. The cross correlation coefficient was greater than 0.90 for 21 of the 24 subjects (88%), demonstrating that the curve shapes were nearly identical for a vast

majority of subjects between surfaces. F_R was also positively correlated between conditions (R ranging from 0.5 to 0.99, $p < .001$). For F_R , 16 out of 24 subjects (67%) had a cross correlation coefficient greater than 0.9. Below are normalized Mz curves representative of the range of correlation coefficients found between the two conditions (Figure 1a–c). Figure 1a shows a subject who had nearly identical curves and Figure 1b is a subject with the same curve shape but different magnitudes. Figure 1c is characteristic of curves with the lowest correlation.

Rate of rise and time to reach peak normalized Mz and F_R were significantly correlated (R ranging from 0.402 to 0.66, $p < .05$) (Table 2).

Upon qualitative examination, we found three unique distributions of Mz curves: unimodal, bimodal, and flat (Figure 2). Unimodal describes a curve with one peak while bimodal is a curve with two peaks. An example of a curve with a flat distribution can be seen in Figure 2 as well.

Bimodality has been described in the propulsion literature with the first peak, termed the *impact spike*, caused by the user forcefully contacting the pushrim at the beginning of the push phase (Robertson et al., 1996; Cooper et al., 1997). There was a significant association between curve type and surface as evidenced by a chi-squared test ($\chi^2 = 9.489$, $p = .008$). Bimodal was most common on the dynamometer whereas unimodal was most common on tile (Figure 3).

Four subjects with bimodal and six subjects with flat distributions on the dynamometer switched to a unimodal Mz curve on tile. As a result, the number of subjects on tile with unimodal Mz curves increased from 5 to 15.

Discussion

This is the first study to perform quantitative and qualitative analyses comparing biomechanics of manual wheelchair propulsion on a simulator and over a natural surface. The higher magnitude mean differences we observed in the kinetic variables and reduction in self-selected velocity for the dynamometer condition are likely a result of a rolling friction inherent in the system that is greater than that experienced over a smooth level surface (DiGiovine et al., 2001; Koontz et al., 2007). However, from the correlation analyses we learned that subjects who pushed with higher forces and moments and larger push angles on the dynamometer tended to do so overground as suggested by the high, positive correlation coefficients for these parameters between the two conditions. Subjects were also consistent, although to a lesser degree, with the timing of their propulsion cycles. Moreover, our results suggest that knowledge of the peak propulsion forces on the dynamometer and body weight provide a good estimate of what the peak propulsion forces are overground as 83% of the variability in overground peak forces was explained by these two variables alone.

The rate at which subjects applied force and when they reached their peak FR and Mz was consistent regardless of condition. Compared with Hurd et al. (2008a) time to peak moment in this study was more similar between conditions. Hurd et al. (2008a) found that peak propulsion moment occurred at 61% and 48% of the propulsion cycle for tile and

dynamometer conditions respectively. Comparatively, our study found peak moments occurring at 62% and 65% of the push phase. As both studies tested samples with similar subject characteristics (e.g., experienced manual wheelchair users with SCI using their own wheelchairs with the SmartWheels attached), differences in time to peak moment found across studies for the dynamometer conditions may be attributed to differences in dynamometer design or setup.

The qualitative analysis of curve type revealed that approximately 60% of subjects (14 of 24) generated similar moment patterns regardless of condition. No subject demonstrated a bimodal or flat curve on the dynamometer unless they also used this curve type on tile. Qualitative findings are strengthened by the cross-correlation (e.g., objective) analysis which indicated excellent agreement in the torque curves between the two conditions. Four subjects identified in the qualitative analysis altered their moment pattern in response to increased resistance to a potentially more injurious pattern (e.g., those who converted from unimodal overground to bimodal on the dynamometer). A bimodal distribution is thought to result from a mismatch of hand and wheel velocity at the start of push phase causing a more forceful contact of the rim (Robertson et al., 1996). The changes in patterns observed in general across the two conditions highlight the importance of evaluating wheelchair users on different types of surfaces. For example, curve shapes may deviate more for persons who have difficulty pushing against a higher resistance, for example, due to muscle weakness, pain, or increased trunk and upper arm impairment. A strength and weakness of this study respectively is the diversity of the sample. However, the sample size was too small to investigate the influence of specific demographic characteristics like disability type and level of injury on the kinetic patterns. Future studies are needed to understand the effects of subject-specific factors on the torque patterns generated during propulsion.

Clinical Relevance

Ideally wheelchair propulsion techniques should be evaluated on natural surfaces. Rehabilitation professionals should take into account that different surfaces may elicit different results. For those who have force-sensing wheels available, quantitative and qualitative information of the propulsion kinetic curves combined may provide further insight into potentially injurious propulsion techniques. Overall, this study demonstrated that experienced wheelchair users are consistent in how much force and torque they use, the direction of force application, rate, and timing of peak force and moment, push angle and stroke frequency when propelling on a dynamometer when compared with propelling on level tile.

Study Limitations

Subjects propelled at a self-selected speed, which differed across conditions. Having included a constant speed condition for both may have led to an increased understanding of the biomechanical differences. However, it should be noted that while a dynamometer can provide visual speed feedback for velocity control, it is more challenging to control speed overground. We have found that pacing a power wheelchair traveling at a fixed speed led to higher stroke-to-stroke variability than observed during a self-chosen speed during propulsion overground (Timko et al., 2009). The dynamometer was not matched to the

rolling resistance of the surface and to the subject's inertial characteristics. As a result, we were unable to separate differences due to inertial characteristics between the two conditions from the environmental factors (e.g., lack of visual cues in dynamometer propulsion) that may have impacted the findings. Despite this limitation we found marked consistencies in key biomechanical parameters of propulsion across the two conditions. A limited number of kinetic and temporal variables were included in this study. Future studies incorporating kinematics, inverse dynamics, electromyography, and metabolic variables may provide a more comprehensive understanding of surface differences.

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Figure 1.

Moment about the hub for dynamometer (solid line) and tile (dashed line) for various correlation coefficients. The highest correlations are seen for curves with similar curve types and magnitudes (A). Panel (B) shows curves with similar slopes but different magnitudes. The lowest correlation is seen between different curve types (C); however, it can be noted that the slope of the final decrease in moment is similar between surfaces.

Figure 2. Representative curves for bimodal, unimodal, and flat curve types. Distinct peaks can be seen in the bimodal and unimodal curves. A characteristic level region can also be seen with the flat curve type.

Figure 3.
Distribution of moment about the hub curve types across conditions.

Table 1

Group means (*SD*) and correlation coefficients for each biomechanical variable for the dynamometer and tile conditions

	Dynamometer	Tile	Correlation Coefficient (<i>r</i>)
max FR (N)	87.1 (22.2)	69.3 (22.1)	0.714 *
max Fz (N)	19.1 (12.2)	14.8 (12.4)	0.317
max Ft (N)	68.4 (17.8)	52.7 (16.0)	0.720 *
max Fr (N)	60.2 (20.3)	52.0 (18.1)	0.514 *
max Mz (N·m)	18.2 (4.75)	14.0 (4.28)	0.720 *
mef	0.58 (0.26)	0.51 (0.32)	0.830 *
Velocity (m/s)	1.20 (0.27)	1.41 (0.18)	0.356
Push angle (deg)	91.4 (15.0)	95.7 (13.9)	0.738 *
Stroke frequency (s ⁻¹)	1.02 (0.20)	0.94 (0.17)	0.406 *

* Indicates significance of $p < 0.05$ for Pearson or Spearman's rho correlation. Abbreviations are defined as follows: FR (resultant force), Fz (medial-lateral force), Ft (tangential force), Fr (radial force), Mz (moment about the hub), and mef (mechanical effective force).

Group mean (*SD*) of time to maximum FR and time to maximum Mz (normalized to a percentage of the push phase) and maximum slopes at the start and end of the weight-normalized FR and Mz curves for the dynamometer and tile conditions. Also shown are the correlation coefficients.

Table 2

	Time max FR (% push phase)	Time max Mz (% push phase)	Slope FR start	Slope Mz start	Slope FR end	Slope Mz end
Dynamometer	58.6 (18.3)	64.5 (15.1)	4.5 (1.8)	0.79 (0.37)	-4.0 (2.4)	-0.95 (0.52)
Tile	49.6 (22.6)	62.4 (15.2)	2.7 (1.4)	0.45 (0.19)	-2.8 (1.6)	-0.66 (0.34)
Correlation coefficient (<i>r</i>)	0.461 *	0.492 *	0.402 *	0.491 *	0.660 *	0.590 *

* Indicates significance at $p < 0.05$ for Pearson correlation.