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A kinetic analysis of manual wheelchair propulsion during start-up on select indoor and outdoor surfaces

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Abstract—The objective of this study was to conduct a kinetic analysis of manual wheelchair propulsion during start-up on select indoor and outdoor surfaces. Eleven manual wheelchairs were fitted with a SMART$^{\text{Wheel}}$ and their users were asked to push on a course consisting of high- and low-pile carpet, indoor tile, interlocking concrete pavers, smooth level concrete, grass, hardwood flooring, and a sidewalk with a 5-degree grade. Peak resultant force, wheel torque, mechanical effective force, and maximum resultant force rate of rise were analyzed during start-up for each surface and normalized relative to their steady-state values on the smooth level concrete. Additional variables included peak velocity, distance traveled, and number of strokes in the first 5 s of the trial. We compared biomechanical data between surfaces using repeated-measures mixed models and paired comparisons with a Bonferroni adjustment. Applied resultant force ($p = 0.0154$), wheel torque ($p < 0.0001$), and mechanical effective force ($p = 0.0047$) were significantly different between surfaces. The kinetic values for grass, interlocking pavers, and ramp ascent were typically higher compared with tile, wood, smooth level concrete, and high- and low-pile carpet. Users were found to travel shorter distances up the ramp and across grass ($p < 0.0025$) and had a higher stroke count on the ramp ($p = 0.0124$). While peak velocity was not statistically different, average velocity was slower for the ramp and grass, which indicates greater wheelchair/user deceleration between strokes. The differences noted between surfaces highlight the importance of evaluating wheelchair propulsion ability over a range of surfaces.

Key words: access, biomechanics, community access, driving surfaces, manual wheelchair, propulsion forces, ramps, rolling resistance, sidewalks, standards, surface resistances, wheelchair propulsion.

Abbreviations: ANOVA = analysis of variance, MEF = mechanical effective force, NVWG = National Veterans Wheelchair Games, SCI = spinal cord injury, 3-D = three dimensional, VA = Department of Veterans Affairs.

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INTRODUCTION

Achievement of the highest degree of independence in a manual wheelchair often depends on the user's ability to negotiate a range of environments and overcome indoor and outdoor obstacles. Surfaces that may pose difficulty for manual wheelchair users include gravel, sand, mud, grass, uneven ground, ramps/hills, carpet, and surfaces that are wet or snow covered [1]. Curbs, bumps, cracks in the pavement, and door thresholds can also be troublesome for wheelchair users [2]. In a majority of wheelchair propulsion studies, a wheelchair ergometer [3–5], dynamometer [6], roller system [7], or treadmill [8–9] are used to simulate wheelchair motion. A stationary system is ideal when certain measurements are needed, such as electromyography of arm musculature, positions of the upper body, and/or physiological data; however, this type of system does not represent the surfaces that a wheelchair user would encounter on a daily basis.

Wheel-based measurement systems allow for the collection of propulsion kinetics and wheelchair kinematics (e.g., position, velocity, and acceleration) in a variety of settings. These systems also require little to no accommodation period. Among others, Sabick et al.* used a custom-developed wheelchair wheel with an instrumented pushrim, a load cell assembly, and a data logging device to collect kinetic data during wheelchair propulsion up a ramp at four different grades (level, 20:1; 12:1, and 8:1). They found that tangential and radial forces increased with increasing grade with no significant change in peak velocity.

The SMART^{Wheel}, a commercial force- and torque-sensing pushrim (Three Rivers Holdings, Inc., Mesa, Arizona), has been used in several studies to examine three-dimensional (3-D) propulsion forces, moments, and/or temporal characteristics over different surfaces and inclines. The SMART^{Wheel} contains an on-board optical encoder that determines the rotational angle of the wheel. Newsam et al. [10] used a SMART^{Wheel} to determine average velocities, distances traveled per stroke, and cadence during propulsion at self-selected and fast-paced speeds over level tile and carpeted floors. Persons with tetraplegia propelled at significantly slower speeds and over shorter distances than persons with paraplegia. Both groups propelled slower and had shorter cycle distances on carpet compared with tile; whereas, stroke cadence remained similar on both surfaces.

Chesney and Axelson [11] used the SMART^{Wheel} to collect objective measurements of surface firmness on playground surfaces (e.g., rubber, sand, pea gravel), carpet, plywood (ramp surface), and hard-packed decomposed granite. The study demonstrated the feasibility, reliability, and limitations of the test methods for determining surface firmness and did not specifically focus on user performance.

Another study used the SMART^{Wheel} for determination of propulsion force and amount of work required to traverse bumpy tile, sloped tile (1% grade), flat tile, and carpet at slow, medium, and fast self-determined speeds [12]. The study considered only steady-state strokes in the analysis. Although the amount of work required was significantly different between surfaces (bumpy versus sloped tile, sloped versus flat tile, bumpy tile versus carpet), the results were based on only one subject who had no disability and who pushed at very slow (0.39 m/s) and moderate (0.69 m/s) average speeds. Data for the fast self-determined speed were not presented.

The high prevalence of upper-limb pain and injury reported among individuals with spinal cord injury (SCI) has been attributed to everyday wheelchair propulsion along with transfers, overhead activities, and weight relief [13–14]. Propulsion forces, rate of force application, and cycle cadence have all been associated with the development of median nerve injury [15–17]. Median nerve injury is the underlying cause of carpal tunnel syndrome. These findings were determined from an analysis of propulsion kinetics at slow and moderate constant speeds on a wheelchair dynamometer that simulated a smooth level tile surface. Specific environmental conditions that may lead to overuse injuries have not yet been identified. The mechanical stresses from wheeling over rough and nonlevel surfaces in a wheelchair user's environment may be even more detrimental to the upper limbs. In addition, little information is available regarding how much force is necessary to start a wheelchair moving on different surfaces.

This study investigated propulsion kinetics over select indoor and outdoor surfaces and focused on the initial start-up of wheelchair motion; prior studies investigated steady-state responses to changes in surface resistance [10,12]. Knowledge of the interaction between

*Sabick MB, Wu HW, Su FC, An K-N. Handrim force increases with increasing ramp grade in wheelchair propulsion [abstract]. Proceedings of the North American Congress on Biomechanics; 1988, August 14–18; Waterloo, Ontario, Canada. Abstract available from: http://asb-biomech.org/onlineabs/NACOB98/81 the wheelchair user and the environment may aid clinicians and consumers in determining the best form of mobility (manual or powered) or the best type of wheelchair for reducing external stresses on the upper limbs. The information gained in this study may also be useful in discussions of environmental barriers to mobility.

METHODS

This study, which took place at the National Veterans Wheelchair Games (NVWG) in Cleveland, Ohio, July 9–13, 2002, was approved by the Department of Veterans Affairs (VA) National Special Events Committee, the local Pittsburgh VA Research and Development Committee, the VA Human Studies Subcommittee, and the University of Pittsburgh's Institutional Review Board.

Study Participants

The inclusion criteria for potential subjects were as follows: (1) use a manual wheelchair as their primary mode of mobility, (2) be between the ages of 18 and 65 years, and (3) use a wheelchair that has quick-release axles. Subjects were excluded from participating in the study if they reported a history of trauma to either upper limb or had a heart or cardiovascular condition that may be exacerbated by pushing a wheelchair. Also, to be eligible for participation in the NVWG, all participants underwent a medical examination and obtained clearance from a physician. Participants at the NVWG have a broad range of disabilities and athletic training. Events include billiards, shooting, bowling, and wheelchair racing. Our past experience has been that the population at the NVWG is similar to a standard veteran wheelchair population [18].

Ten men and one woman volunteered for the study. All subjects provided written informed consent before they participated in this study. **Table 1** provides characteristics of the subjects. Nine of the eleven subjects had an SCI that ranged from L5/S1 to C6/7. One male subject had a unilateral transfemoral amputation, and the female subject had multiple sclerosis. All subjects used ultralightweight manual wheelchairs (K0005), with the exception of one subject (6), who used a high-strength lightweight wheelchair (K0004).

Kinetic Measurement System

Propulsion kinetics were obtained with the use of a SMART$^{\text{Wheel}}$, a 3-D force and torque-sensing pushrim (Three Rivers Holdings, Mesa, Arizona). Details about the system components, percent linearity, and precision of the device have been documented previously [19–20]. The SMART$^{\text{Wheel}}$'s coordinate system follows the right-hand rule, with positive "*x*" forward, positive "*y*" up, and positive "*z*" pointing out of the wheel along the axle. Kinetic data were collected via an infrared wireless transmitter at 240 Hz and then filtered with an 8th-order Butterworth low-pass, a zero-lag, and a 20 Hz cutoff frequency filter [21].

Table 1.
Subject characteristics.

Subject	Gender	Age (yr)	Weight (kg)	Height (cm)	Years with Disability	Disability	SCI Level
1	M	49	88.4	179.1	24	SCI	L4
2	M	54	80.7	182.9	19	SCI	C6/7
3	M	54	77.8	166.4	34	SCI	L5/S1
4	M	62	86.2	175.3	2	SCI	T10
5	M	55	99.8	177.8	11	SCI	T6
6	M	57	79.4	175.3	34	SCI	L3
7	M	53	68.4	170.2	31	SCI	T11
8	M	43	59.0	182.9	21	SCI	T8/9
9	M	27	81.6	188.0	9	SCI	T6
10	F	—	71.2	172.7	—	MS	—
11	M	49	70.3	177.8	8	AMP	—
Mean ± SD	—	50.3 ± 9.7	78.5 ± 11.1	177.1 ± 6.2	19.3 ± 11.5	—	—

M = male; SCI = spinal cord injury; AMP = unilateral transfemoral amputation
F = female; MS = multiple sclerosis; SD = standard deviation

Mobility Course

The mobility course consisted of eight surfaces located in and around the perimeter of the Cleveland Conference Center where the NVWG were held. The targeted surfaces included high-pile carpet, tile, smooth level concrete, low-pile carpet, hardwood flooring, grass, and interlocking concrete pavers (**Figure 1(a)–(g)**). All the surfaces were level except for the interlocking pavers, which varied between 0° and 1.5° (nonuniformly) over the distance traveled. The tiles were 23.5 cm^2, with a 2.23 cm joint between each tile. The interlocking pavers were 15.2 cm^2, with a 0.64 cm joint between each paver. Outside the conference center, subjects pushed on a sidewalk that had a 5° grade. Sections of each surface were marked with tape and ranged in length from 6.1 to 18.3 m (20 to 60 ft) depending on space and availability. **Figure 1** shows the test conditions, and **Table 2** provides the exact lengths of each section. All subjects completed the mobility course without difficulty, with the exception of one subject (subject 5), who had some trouble pushing up the ramp but did not require assistance. The course was completed in the same fixed order for each of the subjects so that wheeling far distances between surfaces would be minimized and the test time would be kept within 2 hours. The subjects started indoors; they pushed first on the high-pile carpet, next on the low-pile carpet, and then on the smooth level concrete. The subjects then proceeded outdoors and pushed up the ramped sidewalk, next on the interlocking pavers, and then on the grass. Afterward, the subjects entered the conference center and pushed on the tile and, lastly, the hardwood flooring.

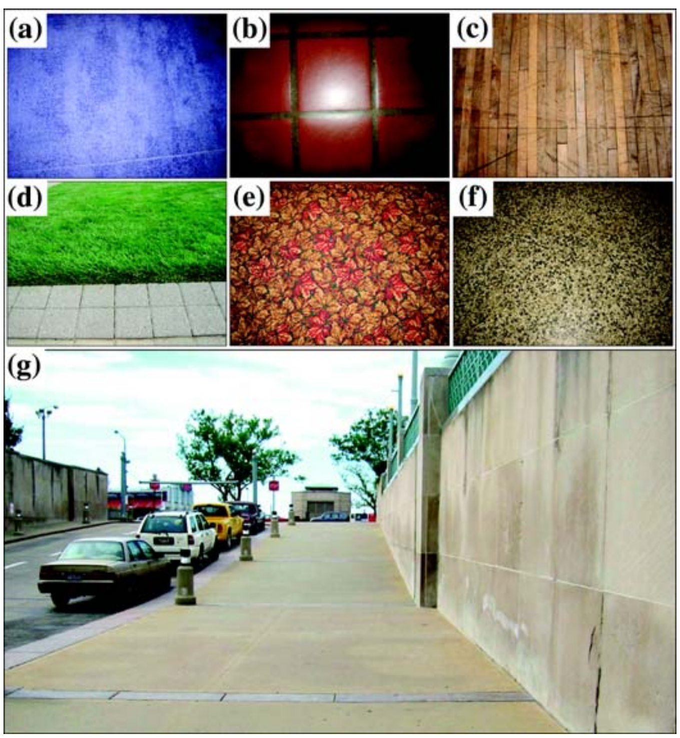

Figure 1.
Surfaces tested were **(a)** high-pile carpet, **(b)** indoor tile, **(c)** wood flooring, **(d)** grass, **(e)** low-pile carpet, **(f)** smooth level concrete, and **(g)** ramp ascent.

Data Collection

The SMART^{Wheel} was secured to each subject's own manual wheelchair on the right side since previous studies have found high correlations between right- and left-side propulsion kinetics [16,22]. Each trial began with the subject positioned at a designated starting line for each surface tested. Subjects were instructed to start propelling their wheelchair from a resting position up to a comfortable pace, pushing in a straight line. They were asked to maintain this pace until they reached the designated finish line at the end of the section. Each trial lasted less than 2 min, with a rest period of at least 5 min in between each section. We recorded the three components of applied pushrim force (F_x, F_y, and F_z), moment about the hub, and wheel position data from the SMART^{Wheel} for the duration of each trial.

Data Analysis

The selected biomechanical variables analyzed for each section of the course were resultant pushrim force, rate of rise of resultant force, mechanical effective force, wheel torque (moment generated about the rear hub), number of strokes, speed, and distance traveled.

Table 2.
Length of each section for all surface conditions.

Condition	Distance (m)
High-Pile Carpet	7.6
Low-Pile Carpet	18.3
Ramp Ascent	7.6
Interlocking Pavers	15.2
Smooth Level Concrete (Reference Surface)	15.2
Grass	6.1
Tile	15.2
Wood Flooring	15.2

Forces and Wheel Torque

We determined the resultant force (F_R) by calculating the vector sum of the SMART^{Wheel} components (F_x, F_y, F_z). We computed the rate of rise of F_R by taking the derivative of F_R with respect to time and then determining the maximum value during the first third of the stroke to capture initial impact load on the pushrim. We chose to analyze F_R and rate of rise of force because these variables have been related to upper-limb injuries among wheelchair users [15]. Mechanical effective force (MEF) is the proportion of force at the pushrim that contributes to forward motion and is defined as F_t^2/F_R^2, where F_t is the tangential force obtained by dividing the measured wheel torque by the radius of the pushrim (0.257 m). This definition of F_t assumes that the hand moment is negligible [23]. We calculated all kinetic parameters over the push phase of the stroke only, which we determined by visually inspecting the wheel torque curves. Because of occasional technical difficulties with the new SMART^{Wheel} instrumentation (interference in data transmission from direct sunlight and accidental user-inflicted damage to the wheel), we could not analyze some trials.

Peak F_R, peak wheel torque, maximum rate of rise of force, and average MEF were determined for the first seven strokes on the smooth level concrete. We chose seven strokes for the analysis since all subjects had performed at least seven strokes before reaching the finish line. We performed separate repeated-measures analysis-of-variance (ANOVA) tests (on each biomechanical parameter) and pairwise comparisons to differentiate start-up biomechanics from steady-state biomechanics. The pairwise comparisons revealed that steady state was achieved after the fourth stroke; that is, strokes five through seven were statistically similar to each other but statistically different from strokes one through four. The biomechanics data for strokes five through seven were averaged and used to normalize the data from the first four strokes for each surface condition. The smooth level concrete was chosen as the reference surface for normalizing all the other surface data because we considered it the easiest surface over which users could push a wheelchair. Since wheelchair users are frequently starting and stopping their wheelchairs throughout the day, we chose the maximum of the start-up strokes for the statistical analysis for all variables except MEF, which was averaged over all four strokes.

Distance, Velocity, and Stroke Count

Linear distance and velocity were determined from the angular wheel position data. For velocity, we calculated a 30-point moving average of the angular distance data before determining the instantaneous velocity for each stroke. This allowed for smoothing of the velocity curve. The total distance traveled in the first 5 s of the trial and the peak velocity of the first four strokes were used in the statistical analysis. We determined stroke count by visually inspecting plots of the wheel-torque curves for the first 6 s and then manually counting the number of cycles and partial cycles (1/4, 1/2, or 3/4) completed in the first 5 s. Postprocessing of all variables in the study was done in MATLAB® (Mathworks, Inc., Natick, Massachusetts).

Statistical Analysis

We compared biomechanical data across each section of the mobility course using separate repeated-measures mixed models ($\alpha < 0.05$) for each variable. The subjects were entered as the random factor, and the surface type was the fixed factor. The statistical program SAS (SAS Institute Inc., Cary, North Carolina) was used for the analysis. A mixed-model test (PROC MIXED) retains all subject data for cases when missing data for a trial are present. This test is different from a traditional repeated-measures ANOVA test (PROC GLM*), which omits all of the subject's data if he/she does not have complete data for every level of the within-subjects independent variable, in this case, surface type. In summary, there were missing trials from subjects: 1 (wood and interlocking pavers), 5 (ramp and grass), 6 (ramp), 7 (interlocking pavers), 8 (high-pile carpet), and 10 (wood). The mixed-model test is valid only if the data were missing at random. No systematic reasons existed for missing data in this study; therefore, this assumption was met. If a significant difference was found, we performed paired comparisons with a Bonferroni adjustment.

RESULTS

Table 3 shows the force and torque data for all seven strokes on the smooth level concrete. Wheel torque and F_R were highest for the first stroke, followed by the second

*General linear model.

Table 3.
Right-side group mean ± standard deviation of propulsion kinetics for first seven strokes on smooth level concrete flooring (reference surface). Last column contains steady-state values (average of last three strokes).

Biomechanics Variable	Strokes							
	1	2	3	4	5	6	7	Average 5–7
Peak Wheel Torque (N•m)	25.2 ± 6.7	22.6 ± 7.0	20.6 ± 8.8	17.5 ± 7.9	13.4 ± 6.3	14.1 ± 5.4	12.5 ± 6.0	13.32 ± 0.8
Peak Resultant Force (N)	103.2 ± 24.4	101.8 ± 30.7	90.4 ± 31.5	74.3 ± 25.8	63.0 ± 29.7	66.7 ± 24.8	61.1 ± 20.2	63.6 ± 2.9
Mechanical Effective Force	1.11 ± 0.49	0.57 ± 0.24	0.76 ± 0.47	0.66 ± 0.59	0.59 ± 0.53	0.52 ± 0.32	0.53 ± 0.36	0.55 ± 0.04
Peak Rate of Rise of Resultant Force (N/s)	406.4 ± 309.7	671.3 ± 195.3	1085.6 ± 1100.8	796.3 ± 644.8	491.4 ± 156.9	579.0 ± 233.0	753.6 ± 654.9	608.0 ± 133.5

and then the third strokes. Wheel torque continued to decrease with each additional stroke, but strokes five through seven were statistically similar. Force rate of rise increased in the first three strokes and then decreased and tapered off for the remainder of the strokes. MEF decreased after the first stroke and gradually leveled off. **Figure 2** shows sample plots of the moments generated about the hub for subject 4 for each section of the mobility course. Considerably higher values were seen for grass and the ramp ascent.

Table 4 shows the maximum F_R, torque, force rate of rise, and MEF for each surface normalized to their steady-state values on smooth level concrete. A mixed-model analysis resulted in significant differences among the surfaces in the normalized F_R ($p = 0.0154$), wheel torque ($p < 0.0001$), and MEF ($p = 0.0047$). F_R rate of rise approached significance with a p value of 0.0680. Results of the paired comparison tests are shown in **Table 5**. Normalized resultant forces at start-up on each surface were 1.8 to 3.5 times higher than the forces applied during steady-state propulsion on smooth level concrete (the reference surface). Forces were lowest for tile and low-pile carpeting and highest for the ramp ascent. After the Bonferroni correction was made, only two surfaces were different from one another in terms of force: the ramp ascent and the reference surface.

Wheel torque was 2.0 to 3.5 times higher for each surface compared with that measured during steady-state propulsion on the reference surface. Again, lower torque was needed for tile and low-pile carpet and higher torque was needed for the ramp ascent. The post hoc comparison on the torque data revealed that high-pile carpet, low-pile carpet, the reference surface, tile, and wood flooring all required less torque than the ramp, interlocking pavers, and grass.

Subjects used a higher proportion of tangentially directed forces during start-up for all surfaces compared with steady-state propulsion over the reference surface. The MEF during start-up was 1.6 to 3.1 times higher than at steady state, with a smaller MEF for the tile and a higher MEF for grass.

Distance, velocity, and stroke count data are shown in **Table 6**. A mixed-model analysis resulted in significant differences between surfaces for total distance traveled ($p < 0.0001$) and stroke count ($p = 0.0068$). Distance traveled in 5 s ranged from 2.5 to 4.8 m. Subjects traveled further on the tile, reference, and interlocking paver surfaces than on the grass or ramp ascent (**Table 5**). We found no statistical difference in the peak velocity between surfaces. Stroke count ranged from 3.7 to 4.8 in the first 5 s of the trial. We found a statistical difference between the ramp and reference surface, with a higher stroke count on the ramp.

DISCUSSION

As expected, our data confirmed that, regardless of surface type, greater propulsion force and torque is needed for users to start pushing the wheelchair from a dead stop compared with maintaining a constant self-chosen pace. Force and torque variables were highest when subjects propelled on surfaces that imposed greater resistance to propulsion (ramp, interlocking pavers, and grass). Because of the differences found in propulsion kinetics between surfaces, assistive technology practitioners must evaluate their client's ability to start up and propel a manual wheelchair on surfaces other than thelevel floors commonly found inside clinic and hospital settings unless the wheelchair will only be used indoors in accessible living accommodations, such as a nursing home or assisted-living center. In addition, wheelchair users should be advised to avoid rapid starts and stops to limit excessive forces on the pushrim.

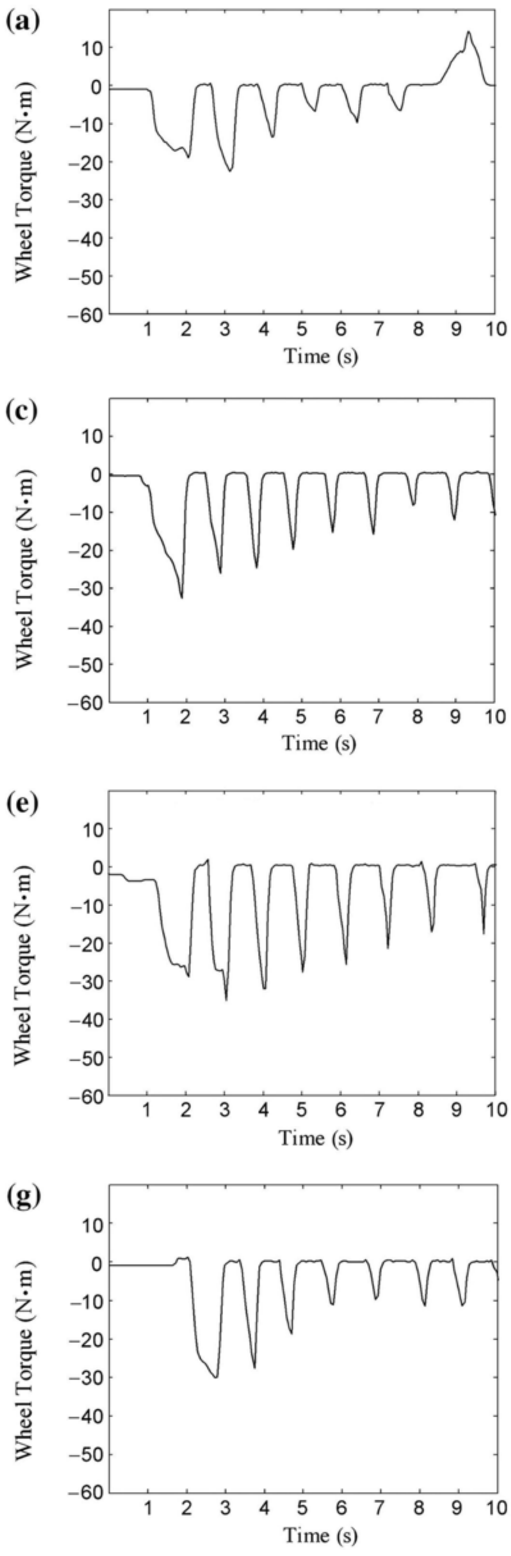

Figure 2.
Sample wheel torque curves for **(a)** high-pile carpet, **(b)** low-pile carpet, **(c)** smooth level concrete, **(d)** ramp ascent, **(e)** interlocking concrete pavers, **(f)** grass, **(g)** tile, and **(h)** wood flooring (subject 4). Wheel torque was about *z*-axis. Moments that cause wheelchair to move forward are negative, and moments that act in opposite direction are positive. Data was truncated at 10 s for graphs only.

Table 4.
Right-side propulsion kinetics for each section of mobility course normalized to steady-state values on smooth level concrete during start-up.

Biomechanics Variable	High-Pile Carpet	Low-Pile Carpet	Reference	Ramp Ascent	Interlocking Pavers	Grass	Tile	Wood	Mean ± SD
	(*n* = 10)	(*N* = 11)	(*N* = 11)	(*n* = 9)	(*n* = 9)	(*n* = 10)	(*N* = 11)	(*n* = 9)	
Peak Wheel Torque	2.04 ± 0.96	2.01 ± 0.77	2.17 ± 0.58	3.51 ± 1.40	2.59 ± 1.20	3.11 ± 1.62	2.00 ± 1.03	2.06 ± 0.79	2.44 ± 0.58
Peak Resultant Force	1.83 ± 0.51	1.79 ± 0.54	1.92 ± 0.53	3.54 ± 1.40	2.38 ± 0.80	2.44 ± 0.68	1.79 ± 0.50	1.87 ± 0.59	2.19 ± 0.61
Peak Rate of Rise of Resultant Force	1.25 ± 0.36	1.22 ± 0.44	2.11 ± 1.73	2.07 ± 0.88	1.68 ± 0.22	2.46 ± 1.89	1.65 ± 1.25	1.25 ± 0.33	1.71 ± 0.47
Mechanical Effective Force	1.31 ± 0.88	1.62 ± 1.01	1.77 ± 1.10	2.09 ± 1.98	1.80 ± 1.37	3.11 ± 3.55	1.60 ± 1.23	1.77 ± 1.02	1.88 ± 0.54

SD = standard deviation

Table 5.
Results of post hoc analysis for wheelchair propulsion variables as well as distance traveled and stroke count.

Biomechanics Variable	Surface 1	Surface 2	Mean* Difference	Adjusted *p*†
Resultant Force (N)	Ramp	Reference	1.62	0.0142
Wheel Torque (N•m)	Ramp	High-Pile Carpet	1.47	0.0052
	Ramp	Low-Pile Carpet	1.50	0.0037
	Ramp	Reference	1.34	0.0002
	Ramp	Tile	1.51	0.0199
	Ramp	Wood	1.45	0.0047
	Interlocking Pavers	Low-Pile Carpet	0.58	0.0315
	Interlocking Pavers	High-Pile Carpet	0.55	0.0221
	Interlocking Pavers	Tile	0.59	0.0204
	Interlocking Pavers	Wood	0.53	0.0072
	Interlocking Pavers	Reference	0.42	0.0216
	Grass	Low-Pile Carpet	1.10	0.0187
	Grass	Reference	0.94	0.0195
	Grass	High-Pile Carpet	1.07	0.011
	Grass	Tile	1.11	0.0004
	Grass	Wood	1.05	0.0006
Mechanical Effective Force	Grass	Tile	1.51	0.0047
Distance (m)	Reference	Ramp	1.46	<0.0001
	Low-Pile Carpet	Grass	1.98	0.0006
	Reference	Grass	2.28	0.0001
	Interlocking Pavers	Grass	3.02	0.0017
	Indoor Tile	Grass	1.69	<0.0001
	Wood	Grass	1.85	0.0025
Stroke Count	Ramp	Reference	0.51	0.0124

*Surface 1 means were larger than surface 2 means in each case.
†The adjusted *p* is the *p*-value after the Bonferroni correction.

Table 6.
Peak velocity during start-up, distance traveled, and number of strokes completed in first 5 s.

Biomechanics Variable	High-Pile Carpet (n = 10)	Low-Pile Carpet (N = 11)	Reference (N = 11)	Ramp Ascent (n = 9)	Interlocking Pavers (n = 9)	Grass (n = 10)	Tile (N = 11)	Wood (n = 9)	Mean ± SD
Distance (m)	3.75 ± 0.66	4.51 ± 0.91	4.81 ± 0.83	3.35 ± 0.96	5.55 ± 1.97	2.53 ± 0.76	4.22 ± 0.86	4.38 ± 0.70	4.14 ± 0.92
Peak Velocity (m/s)	1.10 ± 0.18	1.26 ± 0.14	1.38 ± 0.17	1.10 ± 0.28	1.28 ± 0.17	0.91 ± 0.25	1.20 ± 0.17	0.88 ± 0.14	0.83 ± 0.18
Stroke Count	3.65 ± 0.67	3.73 ± 0.67	3.82 ± 0.98	4.33 ± 0.66	4.22 ± 0.75	4.75 ± 1.62	3.95 ± 1.11	4.00 ± 0.87	4.06 ± 0.36

SD = standard deviation

Data recorded during dynamometer testing that emulated propulsion on a smooth level surface at low to moderate speeds [15,17] have linked peak force and rate of rise of force to wrist injuries. These studies and others have found that wheelchair users routinely exert forces over 70 N and moments over 9 N·m. This study found that propulsion forces and moments are considerably higher when wheelchair users are starting up, propelling uphill, or pushing over grass and interlocking pavers. Thus, frequent starting and stopping and regularly propelling on outdoor and inclined surfaces may further increase the likelihood of users developing upper-limb injuries because of increased and cumulative loading on the arms. For this reason, as a way to reduce stress on their arms and preserve upper-limb function, individuals who work outdoors (e.g., on a farm) or spend much leisure time outdoors may want to consider alternative mobility devices, such as crank- or lever-driven wheelchairs [24], power-assist hubs [25], add-on power units, scooters, or fully powered wheelchairs.

Peak velocity did not significantly differ between surfaces; however, differences existed in average velocity during start-up. The average velocity for each surface, calculated from the distance traveled over 5 s, indicates that subjects pushed at a slower average speed on the ramp than on the reference surface and slower on the grass than on the low-pile carpet, interlocking pavers, tile, wood flooring, or reference surface. The absence of differences in peak velocity indicates greater between-stroke deceleration on the more challenging surfaces. Users adjusted their applied force and torque to compensate for this deceleration. On the ramp, more strokes were also required to keep the wheelchair moving forward.

Comparing our findings with those of other studies [10,26] is difficult since the authors reported only average speeds of propulsion over different surfaces and not start-up velocities. The average speeds during start-up in this study ranged from 0.51 m/s (grass) to 1.11 m/s (interlocking pavers), with an overall average speed across surfaces of 0.82 m/s. In Newsam et al.'s study, the average self-chosen speed over tile was 1.31 m/s for the entire group of 70 users with paraplegia and tetraplegia [10]. In an earlier study by Wolfe et al., subjects with paraplegia pushed at a self-chosen speed of 1.36 m/s over level concrete and 1.07 m/s over carpeting "of the same type used in hospital and nursing homes" [26]. Not too surprisingly, the average speeds during start-up in this study are lower than those reported during steady-state on similar surfaces in other studies; wheelchair users generally do not go faster at start-up and then slow down.

We applied inferential statistics in this study to assess group differences between surfaces, but perhaps of greater interest to the clinician is whether a certain patient is capable of performing at a functional level in his or her wheelchair on different surfaces. For instance, two subjects (1 and 5) had peak velocities that were generally lower than the mean on most surfaces. Factors that could influence mobility performance in general may include functional capacity (e.g., strength, endurance, spasticity), wheelchair type and characteristics (e.g., weight of the system, frame material properties, suspension, caster size and type) [27], wheelchair maintenance (e.g., frame and caster alignment, air pressure), wheelchair setup (e.g., camber and horizontal and vertical axle position) [28], and user characteristics (e.g., experience, activity level, disability type, gender, age, weight, upper limb pain) [10,15,29]. Investigating these factors can help the clinician understand reasons why their patient is not performing at a functional level.

The measured moment at the hub is a combination of hand moment and the product of tangential force and wheel radius. A kinematic measurement system is best for separating the hand moments from the product of tangential force and wheel radius. This type of system was not available for this study; thus, this procedure was not performed. Therefore, the measured moment was used in the calculation of MEF. Prior studies have found MEF or the fraction of effective force during steady-state propulsion to range from 0.26 to 0.72 [15,30–31]. The maximum value for MEF is “1,” which would indicate that all the forces applied to the pushrim were used to generate wheel rotation. However, this assumption is not realistic since some nontangential force is necessary for grasp and for maintaining contact with the pushrim. The traditional calculation for MEF assumes that hand moments are negligible; however, several studies have shown that hand moments are not negligible at low speeds and during propulsion uphill [32–34]. Our data indicate that for a user to propel a wheelchair from a dead stop also requires significant hand moments since the MEF exceeded a value of “1” for the first stroke on the reference surface. In addition, the MEF was 1.3 to 3.1 times higher during start-up than at steady state (0.73 on high-pile carpet to 1.71 on grass). How hand moments assist with start-up propulsion across rougher terrain and inclines would be an interesting topic for future study.

Although differences were noted in some kinetic variables, they may not have been statistically significant because of the small sample size and insufficient power of the statistical tests. While using a diverse sample adds strength to the study, the variability in the measures may be higher because of differences in user characteristics. These differences could be examined in more detail with a larger sample of diverse users who could be separated into groups and compared.

The types of surfaces tested were limited to those available in and around the Cleveland Conference Center. Future studies should incorporate a broader range of surface types, ramp grades, and surfaces exposed to inclement weather conditions. Objective information regarding surface roughness and stability [11] could have provided a greater understanding of the differences found and the opportunity for study replication. For example, the differences seen for the interlocking pavers could have been attributed to surface roughness due to the beveled edges and joints.

Since the order of the surfaces was not randomized, the subjects could have experienced a “warm-up” period or performed the later trials differently because they were tired or were anxious to finish the testing. All the wheelchair users in this study used the SMART^{Wheel} on their everyday wheelchairs, which is unique compared with the aforementioned studies of wheelchair propulsion in different environments. Only one SMART^{Wheel} was available at the time of the study. A bilateral analysis of propulsion technique may provide insight into side-to-side asymmetries present during “free” propulsion over different obstacles and surfaces.

CONCLUSION

Users in this study adapted to changes in surface resistance at start-up by modifying their applied forces, torque, and number of times they struck the pushrim, depending on surface type. While peak start-up velocity was similar across surfaces, the average velocity for the ramp and grass was slower than for other surfaces, which indicates greater deceleration of the wheelchair/user on these surfaces. Forces and wheel torque during start-up for all surfaces tested in this study were considerably higher compared with steady-state propulsion on a smooth level surface. As a result, in assessing wheelchair mobility, clinicians should consider propulsion ability on a range of surfaces and conditions and not only level-ground steady-state ambulation. The results of this study may help indicate possible wheeling conditions that are responsible for overuse of the upper limb in wheelchair users and thus help in discerning mechanisms responsible for injuries.

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