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The effect of knee-flexion angle on wheelchair turning

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Abstract

The increasingly popular hyperflexed knee-flexion angle was evaluated to determine its effects on wheelchair turning. Twenty able-bodied subjects were tested comparing the effect of full knee extension and full knee flexion on a number of parameters. We empirically measured the angular velocity of subjects spinning 720° in place, subjects' perceived ease of wheelchair turning, the overall length of the wheelchair, the anteroposterior position of the center of mass (COM), rolling resistance, turning resistance and rear-wheel traction. The combined moment of inertia of the wheelchair and system was modeled. We found that, in comparison with full extension, fully flexing the knees increased angular velocity by 40% and was perceived to be 66% easier by subjects. Overall length decreased by 39%, COM moved rearward 38%, rolling and turning resistance decreased by 21% and 17% respectively, rear-wheel traction increased by 12% and moment of inertia decreased by 42%. All empirically tested parameters were statistically significant ($p<0.007$). We conclude that the knee-flexion angle has a significant effect on wheelchair turning. The implications of these findings for wheelchair design and prescription will need to be validated on actual wheelchair users and for smaller increments in knee-flexion range. © 2001 IPEM. Published by Elsevier Science Ltd. All rights reserved.

Keywords: Wheelchair turning; Knee position; Center of mass; Rolling resistance; Turning resistance; Traction; Moment of inertia

1. Introduction

Wheelchair positioning has been found to affect pressure distribution, stability, and propulsive efficiency [1–6]. However, limited research has been conducted on the effects of body positioning on wheelchair turning. Most wheelchair users have a 90–100° seat-to-backrest angle and 60–90° of knee flexion [7], with 0° being full extension [8]. Many lightweight and special-purpose wheelchairs allow for greater than 90° of knee flexion. Surprisingly, no research has been published on the effects of this increasingly popular hyperflexed knee-flexion angle on wheelchair performance measures, including wheelchair turning.

Brubaker has speculated that the ease of turning about the yaw (vertical) axis is enhanced by decreasing the horizontal distance from the center of mass (COM) to the rear wheel axle [5]. In addition, if wheelchair footrests are adjusted so that the knees are flexed greater than 90°, the overall length of the wheelchair should be reduced. This would enable the user to complete tighter turns, move closer to objects, protect the feet, and transport the unoccupied wheelchair more easily. Turns about the yaw axis should also be faster and more easily performed due to the probable effect of knee flexion on the moment of inertia [1]. Because the moment of inertia directly affects the force required to maneuver a wheelchair [9], a reduced moment of inertia should also help to decrease upper-extremity overuse injuries commonly seen in wheelchair users [10].

Varying body postures in a wheelchair can alter the vertical and horizontal position of the COM with respect to the wheelchair's rear wheel axles, thereby affecting a number of performance variables. One such variable is rolling resistance, the resistance to propulsion resulting from ground and wheel interactions. Movement of the COM rearward, closer to the rear wheel axles has been

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found to decrease rolling resistance [5], presumably due to the increased proportion of the weight on the larger rear wheels. Because rolling resistance is inversely proportional to wheel diameter, it can be inferred that knee-flexion angle should affect rolling resistance of rear-wheel-drive wheelchairs, to the extent that it alters the fore–aft position of the COM.

A direct linear relationship has been found to exist between caster load and turning resistance, that is, the force required to initiate wheelchair turning [11]. Therefore, if the knee-flexion angle increases (moving the COM rearward), less weight is placed on the caster wheels and turning resistance should decrease. With more weight on the rear wheels, rear-wheel traction should increase because friction between the rear wheels and the floor is proportional to the load on the rear wheels.

The purpose of this study was to determine whether knee-flexion angle affects wheelchair turning on level surfaces and, if so, what the extent of this effect is and why it occurs. Our primary hypothesis was that wheelchair turning is easier if the knees are flexed rather than extended. Secondary hypotheses were also investigated as probable reasons for the increased ease of turning due to the increased knee-flexion angle. The secondary hypotheses were that, with the knees flexed, overall wheelchair-user length decreases, the COM moves rearward, rolling and turning resistance decrease, traction increases and the moment of inertia decreases.

2. Methods

2.1. Subjects

We studied a sample of convenience consisting of 10 male and 10 female able-bodied subjects. The appropriate sample size was not estimated using a power analysis because the variance of the data was unknown. Able-bodied subjects were studied rather than wheelchair users because many wheelchair users would not have the range of motion necessary to achieve the knee-angle positions, or the muscular strength needed to maintain these positions. The anthropometric make-up of actual wheelchair users can vary widely, with the mass of the lower legs sometimes increased (e.g., due to edema) and sometimes decreased (e.g., due to muscular atrophy or amputation). However, it is common to use the anthropometric make-up of able-bodied subjects to provide general indications about wheelchairs (e.g., using ISO test dummies for fatigue tests).

Eligible subjects were between the ages of 18 and 50 years, were 152–188 cm tall and had a mass less than 90 kg (to fit the wheelchair used), and did not use a wheelchair for everyday mobility. Subjects were actively recruited to participate in this study. Ethical approval for this study was obtained from the Research Ethics Committee of the Queen Elizabeth II Health Sciences Centre, and informed consent was obtained from all subjects prior to data collection.

Each subject's age, gender, height, and weight were recorded. Height was measured to the nearest 0.01 m and weight was recorded to the nearest 0.1 kg using a balance scale (Detecto-Medic Scale, Cardinal Scales Manufacturing Company, P.O. Box 151, Webb City, Missouri, 64870). The knee range of each subject (at full flexion and full extension) was measured with a goniometer to the nearest degree with the subject sitting in the wheelchair. The greater trochanter, lateral femoral epicondyle and lateral malleolus were used as anatomical landmarks. All subjects were normally clothed and wore shoes.

2.2. Subject position

Unless otherwise stated for specific outcome measures, the following subject position was maintained throughout data collection. Each subject sat in the wheelchair with the wheelchair seatbelt fastened. The subject kept the upper and lower back against the backrest at all times to prevent variations in COM position resulting from trunk flexion rather than knee-flexion angle. The forearms were placed on the armrests (keeping the elbow directly below the shoulder), and the thighs were kept together on the seat. Foam cushioning (400 g) was placed between the clothing guards of the wheelchair and the subject's thighs to maintain thigh positioning and to limit mediolateral leg motion during rapid turning.

2.3. Wheelchair

The wheelchair (Quickie LXI wheelchair, Sunrise Medical Canada, 237 Romina Dr., Unit 3, Concord ON, L4K 4V3) that we used for this study was a lightweight manually propelled model, with rear-wheel drive and swivel casters. The frame was 46 cm wide, the seat was 41 cm deep, and the backrest was 37 cm high. The caster wheels were 13 cm in diameter with low-profile polyurethane tires. The armrests were adult sized, mounted with a single post, with desk-length pads (25 cm), that were adjusted to a height of 27.5 cm above the seat. The axles were quick-release and the axle plates were adjusted to the second from the furthest back position and the middle of the three vertical possibilities. The rear wheels were 61 cm in diameter with metal spokes, plastic coated handrims, and pneumatic treaded tires with airless inserts. There were also push-to-lock brakes.

To test the two extremes of full knee extension and knee flexion, the front rigging of the wheelchair was removed because there was no commercially available front rigging that could be adjusted to the two extreme knee-flexion angles tested. Removing the front rigging can be expected to result in slightly different results than would be expected with the front rigging still in place, in that the mass of the front rigging (1.4 kg) was absent. However, this was expected to have a conservative effect that would tend to underestimate the effects under study.

2.4. Procedure

Subjects completed four trials in total for each parameter, twice with full knee extension (~0°) and twice at full knee flexion (~120°). The two extreme knee-flexion angles were chosen for this study to provide some initial evidence as to the effect that knee-flexion angle has on wheelchair turning. When carrying out a preliminary study such as this, where there is no existing literature to guide one's estimates of effect size, it is a common strategy to look at extreme situations (as long as they are clinically plausible). If there had been no significant findings at the extremes, then there would be little point in looking at finer increments. We considered full knee extension to be a reasonable extreme in that this position is commonly used when elevating legrests are prescribed.

To eliminate any bias due to the order of testing, the flexion and extension trials were randomly balanced for each test. The two trials at each position were to allow the reliability of the measures to be determined.

2.5. Floor preparation

The floor surface was flat painted concrete. To keep variability due to dust or dirt to a minimum, the testing floor was washed with detergent prior to data collection.

2.6. Angular velocity

Warm-up and training for the maximum angular velocity test consisted of the subject twice propelling forward 5 m followed by reversing back to the start. The subject then maneuvered through a figure-eight pattern, using two pylons 2 m apart. After a 60-second rest period, the experimenter positioned the wheelchair in the center of a 1.6 m diameter circle. The subject then performed 2 practice trials of the angular velocity test described below. Between the practice trials and the recorded trials, there was a two-minute rest.

The maximum angular velocity test consisted of timing the subject as he/she attempted to rotate the wheelchair 900° about the yaw axis at maximum speed. All four wheels had to remain within the 1.6 m-diameter circle. This diameter was chosen because, during pilot work, it kept the yaw axis approximately between the rear wheel tires. The subject started in the wheelchair with the casters trailing backwards (as though wheeling forward) in the center of the circle, facing the experimenter, hands placed top dead center on the push-rims of the rear wheels. The subject was then asked to rotate the wheelchair 2.5 times as quickly as possible in the direction of the non-dominant hand. The extra 180° was to avoid having subjects stop at 720°. If the subject moved outside the circle before completing 720° the attempt was considered a mistrial and was repeated. The time needed to turn the wheelchair 720° was timed to the nearest 0.01 s using a stopwatch. To calculate angular velocity (degrees/s), 720° was divided by time. Trials were also videotaped using a video camera (Zenith VM 7170, Zenith Electronics Corporation, 1000 Milwaukee Ave., Gelnview IL, 60025) positioned 3 m from the center of the turning circle to evaluate mistrials.

2.7. Subjects' perception

Immediately following the angular velocity trials, each subject was asked to quantify his/her subjective impression of the ease of turning by using visual analog scales (VAS) [12]. Each VAS was labeled "extremely easy" at 0 mm and "extremely hard" at 100 mm. VAS have been widely used to quantify subjective impressions. Subjects marked along the 100 mm line to indicate how they perceived the ease of turning with the knees fully flexed and fully extended.

2.8. Overall length

The overall length of the subject and the wheelchair was measured to the nearest 0.001 m to reflect the magnitude of the turning circle. The wheelchair was placed with the rear-wheels against a wall and the distances from the wall to the end of the foot (for knee extension), and to the end of the patella (for knee flexion) were measured with a measuring tape.

2.9. COM position

Using a roll-on scale (HR50, Howe Richardson Inc., 214 Brunswick Blvd., Pointe Claire PQ, H9R 1A6), the reaction board method was used to determine how horizontal COM position changed as knee-flexion angle changed. The total mass of the subject and the wheelchair was measured to the nearest 0.1 kg. The mass distribution on the front wheels (trailing backwards) was then measured by rolling the rear wheels off the scale leaving the caster wheels 20 cm from the edge of the scale (to avoid edge-of-scale effects). Wheel brakes were applied. The wheelbase (horizontal distance between the front caster and rear wheel ground contact points) was measured to the nearest 0.001 m with the casters trailing backwards. These measured values were entered into the following equation:

$$d_1 = \frac{F_2 \times d_2}{F_1} \quad (1)$$

where d_1 is the unknown horizontal distance from the rear axles to the gravity force line of the COM, F_1 is the total weight of the wheelchair and the subject, d_2 is the wheelbase, and F_2 is the force on the front wheels.

2.10. Rolling resistance

To measure rolling resistance, we used a variation on the coast-down method [13]. The experimenter positioned subjects on a 5° aluminum ramp, leaving the caster wheels (trailing backwards) on the floor touching the end of the ramp. The wheelchair was then released and the horizontal distance traveled from the center of the edge of the ramp to the lead caster wheel axle (rolling distance) was measured to the nearest 0.01 m. Because the wheelchair could coast slightly to the left or the right rather than in a perfectly straight line, a range of 30° was marked on the floor with masking tape from the center of the edge of the ramp. This created a fan-like area within which the wheelchair moved. Trials resulting in a finishing position with the wheels outside this area were declared mistrials and were repeated. The method described above was chosen for this study after piloting a number of alternative possibilities including the treadmill method [14], and measuring the force needed to pull the wheelchair over level ground. Results from pilot work indicated that the coast-down method was the most reproducible and the easiest to execute.

2.11. Turning resistance

Turning resistance, the initial resistance to rotation, was measured using a dynamometer (Chatillon CSD 200 Strength Dynamometer, Ametek/Chatillon Test and Calibration Instruments, Division 8600, Somerset Dr., Largo FL, 33773). A yaw force was applied to the left side of the front of the wheelchair frame 41 cm above the ground. Care was taken to ensure that the force was horizontal and perpendicular to the sagittal plane of the wheelchair. The caster wheels were initially in rear-trailing positions. The experimenter gradually applied force to cause the wheelchair to rotate counterclockwise until the ground contact point of the left caster wheel crossed a line 2.5 cm from the initial starting position. Brakes were not applied. The axis of rotation of the wheelchair was through the contact point between the left rear wheel and the floor. The peak-applied force was recorded to the nearest Newton.

2.12. Rear-wheel traction

Sliding friction was measured to evaluate how traction changed as knee positioning was altered. Static friction is the product of the coefficient of sliding friction and the object's normal force [15]. Due to the rearward movement of the COM with increased knee flexion, the normal force on the rear wheels (and thus static friction) was expected to increase. With increased static friction, there should be an increase in rear-wheel traction (degree to which the rear-wheels maintain contact with the floor) which should facilitate more efficient turning.

To test the effect of knee positioning on traction, both rear-wheel brakes were applied. The rear wheels were also tied to the wheelchair frame to prevent any movement of the wheels that could have occurred due to brake slippage. The casters were free to roll. Rope was placed around the front of the wheelchair frame (40 cm above the floor) and attached to the dynamometer. The experimenter applied tension to the rope–dynamometer system gradually (avoiding a jerking motion) and the peak force needed to initiate movement (~5 cm) of the wheelchair was recorded. Care was taken to keep the dynamometer level and parallel to the sagittal plane. The peak force required to move the wheelchair forward was recorded to the nearest Newton.

2.13. Statistical analysis

Test–retest reliability was calculated with Pearson's correlation coefficients and matched-pairs *t*-tests using the results from the two trials at each knee flexion angle for all outcome measures except for overall length, which was only measured once. Then the two trials for each knee position were averaged and the difference between the two knee-flexion angles were compared using one-tailed matched-pairs *t*-tests. The definition of statistical significance was $p<0.007$, having used a Bonferroni adjustment to the $p<0.05$ level to allow for having made seven comparisons.

2.14. Modelling

To determine the moment of inertia of the wheelchair user, the magnitude and location of the center of mass (COM) of each body segment was needed. We used Dempster's cadaveric data and anthropometric data for the 2.75th and 97.5th percentile male and female populations to include the widest possible range of body size [16] (pp. 51–74), [17] (pp. 4–18). Using able-bodied anthropometric data allowed us to compare our modeled results on the moment of inertia with the empirically tested parameters.

Although the seat and seat–backrest angles vary among wheelchairs and the wheelchair user's posture is free to vary during activities [4], for the purposes of the modelling, we assumed that the seat angle was 0° (horizontal) and the seat–backrest angle was 90°. The wheelchair user was modeled in an erect sitting posture with the back against the wheelchair backrest. The hips and elbows were flexed 90°, the wrists and ankles were neutral, but the knee-flexion angle was varied. We assumed, on the basis of pilot work, that the vertical or yaw axis of rotation was located between the rear wheel axles when the wheelchair turned in place.

To determine the total-body moment of inertia, the moment of inertia about each segment's COM was first calculated. The radius of gyration (r) and mass of each segment were obtained from the literature [18] (pp. 142–172), and entered into the following equation [9]:

$$I=\sum mr^2 \tag{2}$$

Radius of gyration values are given as a percentage of segment length. Therefore, to determine the moment of inertia about each segment's COM, each segment's length was multiplied by the appropriate radius of gyration value (about the proper axis of rotation described below), to provide the r value for Eq. (2).

The radius of gyration for each segment was determined based on its orientation relative to the yaw axis about which the system was rotating. Radius of gyration values are given for the standing anatomical position in the literature, therefore, some adjustments were necessary to model the seated wheelchair position described above. In the assumed model position, the head, trunk, and upper arm were orientated in the anatomical position and thus the radius of gyration values about the yaw axis were used. In the seated position, the forearm, hand and thigh segments had rotated 90° from their anatomical positions. As a result, radius of gyration values about the saggital axis were used to calculate the moment of inertia about the COM of these segments and the lower leg and foot.

The parallel axis theorem was then used to determine the moment of inertia of each segment rotating about the yaw axis (in this case between the rear wheel axles). The parallel axis theorem allows for the moment of inertia about an object's COM to be translated about another center of rotation located any distance away from the COM, if the axis of the new center of rotation is parallel to the axis about which I_{com} was calculated. The formula is as follows:

$$I_{\mathrm{new}}=I_{\mathrm{com}}+mx^2 \tag{3}$$

where I_{new} is the moment of inertia about the new axis, and x is the distance (in m) between the COM and the center of rotation [16].

The distance between the COM and the center of rotation (x) is affected by changes in knee-flexion angle, and is the variable responsible for changes in moment of inertia. The distances from the COM of all body segments to the axis of rotation were determined using anthropometric data and basic geometry in the case of the COM of the lower leg and foot. The distance between the wheelchair's COM and the axis of rotation was measured directly. Once these variables were determined, the total-body moment of inertia was calculated by summing all of the segmental values together (after accounting for bilateral segments).

Total body moment of inertia was calculated for knee-flexion angles from 0° (full extension) to 120° in ten-degree increments for both the male and female models. This was done to allow for insight into the effect of a range of knee angles that would have been difficult to test empirically.

To determine the moment of inertia of the unoccupied lightweight wheelchair used (the same one used for empirical testing) the fore–aft position of the COM was determined using the reaction board method with the front rigging removed (for the reasons noted earlier). The COM of the wheelchair was located at 26% of the wheelbase (with the casters trailing backward) from back to front. Midline location was assumed.

The torsional vibration method was used to determine the period of vibration as the wheelchair oscillated about its COM [19]. This involved hanging the chair with three strings of equal length orientated equidistant from the wheelchair's COM. The wheelchair was level, as if propelling on a flat surface, and all moveable parts (i.e., casters, and rear wheels) were stabilized (Fig. 1). Then the wheelchair was rotated about the yaw axis approximately 10° to one side and was allowed to oscillate freely. The period was then calculated (number of cycles over a five-minute period) and averaged over three trials. The period was then entered into the following equation, calculating the wheelchair's moment of inertia:

$$I_{\mathrm{com}}=\frac{mgr^2T^2}{4\pi^2L} \tag{4}$$

where I_{com} is the moment of inertia about the object's COM, m is the mass of the wheelchair (kg), g is the gravitational constant (9.81 m/s^2), r is the distance from the cords to the wheelchair COM (m), T is the period of vibration (cycles/s), and L is the lengths of the cords (m) [19].

3. Results

The subjects' mean (SD) age, height, weight, knee flexion and extension ranges were 21 (0) yrs, 1.75 (0.08) m, 68 (12.2) kg, 135 (6)° and 9 (5)°. Correlation coefficients for each parameter, as a reflection of the test–retest reliability are shown in Table 1. Although correlations were high for most parameters ($r>0.71$, $p<0.007$), the turning resistance trials were not highly correlated. On matched-pairs t-tests, there were no significant differences between trials 1 and 2 for any of the parameters.

The flexion and extension data are presented in Table 2 and Fig. 2. Significant differences were found between full knee flexion and full knee extension for all parameters empirically tested ($p<0.007$). Effect sizes were

Fig. 1. Set-up for the determination of wheelchair moment of inertia. Strings located on the push handles and front seat of the wheelchair are equidistant from the wheelchair COM that is illustrated by the plumb bob.

Table 1
Test–retest reliability for each parameter[a]

Parameters	Reliability Extension	Flexion
Angular velocity	0.81*	0.94*
Center of mass	0.99*	0.97*
Rolling resistance	0.96*	0.72*
Turning resistance	0.45	0.26
Traction	0.89*	0.78*

[a] Values shown are correlation coefficients. *Indicates r values that were statistically significant ($p<0.007$).

calculated by relating the knee-flexion values to those for knee extension. With the knees flexed, angular velocity was 40% faster, subjects' perceived exertion decreased by 66%, overall length was reduced by 39%, COM was 38% closer to the rear wheel axles, rolling resistance was 21% lower, turning resistance decreased by 17% and rear-wheel traction increased by 12%.

Plots of the moment of inertia against knee-flexion angle for all models are presented in Fig. 3. The relationships were all non-linear ones, the regression equations for which are shown in Fig. 3. For the small female model (2.75 percentile), the total wheelchair-user system moment of inertia changed from 4.7 kg m/s^2 at 0° of knee flexion to 2.8 kg m/s^2 at 120° of knee flexion (a 41.0% decrease), whereas the large female model's (97.5 percentile) moment of inertia decreased from 7.5 kg m/s^2 to 4.2 kg m/s^2 (a 43.3% decrease). In the small male model, the wheelchair-user system's moment of inertia decreased from 5.9 kg m/s^2 to 3.4 kg m/s^2 from 0° to 120° of knee flexion (a 43.5% decrease), whereas the large male model's moment of inertia changed from 13.1 kg m/s^2 to 8.1 kg m/s^2 (a 38% decrease).

4. Discussion

All empirically tested hypotheses were corroborated. Increasing knee-flexion angle from full extension to full flexion increased the ease of turning to a clinically significant extent. Both the angular velocity and the perceived exertion (VAS) associated with the increased knee-flexion angle were affected. Several factors appeared to contribute to the increased ease of turning.

The largest contributing effect (38%) was seen in the horizontal position of the COM relative to the rear wheel axles. The COM effect most likely caused the decreased moment of inertia, rolling resistance, turning resistance, and increased traction [5,9,14]. In the method used to determine COM position, it should be noted that the wheelbase (d_2) was measured with the casters in rear trailing positions. Turning the wheelchair orientates the caster wheels in the direction of rotation, effectively lengthening the wheelbase by the extent of caster trail (7.5 cm for the wheelchair that we studied). However, in pilot work, where one subject was tested with the casters in both rear-trailing and side-trailing positions, in the latter position there was only a 3% (1 cm) smaller change in COM position due to knee flexion.

Rolling resistance was found to have the second greatest effect size (21%). It was measured indirectly by measuring the rolling distance, with a longer distance representing a decrease in rolling resistance. The significant difference found was most likely due to the decreased mass distributed on the casters, as rolling resistance is inversely proportional to wheel diameter [5].

Turning resistance had the next greatest effect size (17%). Although turning resistance appeared to be a significant contributing factor to the ease of turning, the low reliability of the method used decreased our confidence in the significance of the effect. This low reliability may have been due to experimenter error, or the fact that the sensitivity of the dynamometer was only to the nearest Newton. A more reliable method of

Table 2
Effect of knee-flexion angle on wheelchair turning[a]

Parameters	Knee extension		Knee flexion		Difference	
	Mean	SD	Mean	SD	Mean	SD
Angular velocity (deg/s)	106	14	148	23	42*	17
Perceived exertion (%)	64	16	22	13	42*	21
Overall length (cm)	127.5	7.5	78.2	4.5	49.3*	5.4
Center of mass (%)	34.4	4.1	21.5	4.0	13.0*	3.6
Rolling resistance (cm)	248.0	34.8	299.1	24.3	51.1*	17.1
Turning resistance (N)	27.2	5.5	22.6	4.8	4.6*	3.0
Traction (N)	184	41	206	41	21*	23

[a] Values are the mean of two trials. *Indicates statistically significant differences ($p<0.007$). Center of mass values are presented as a percentage of total wheelbase from back to front.

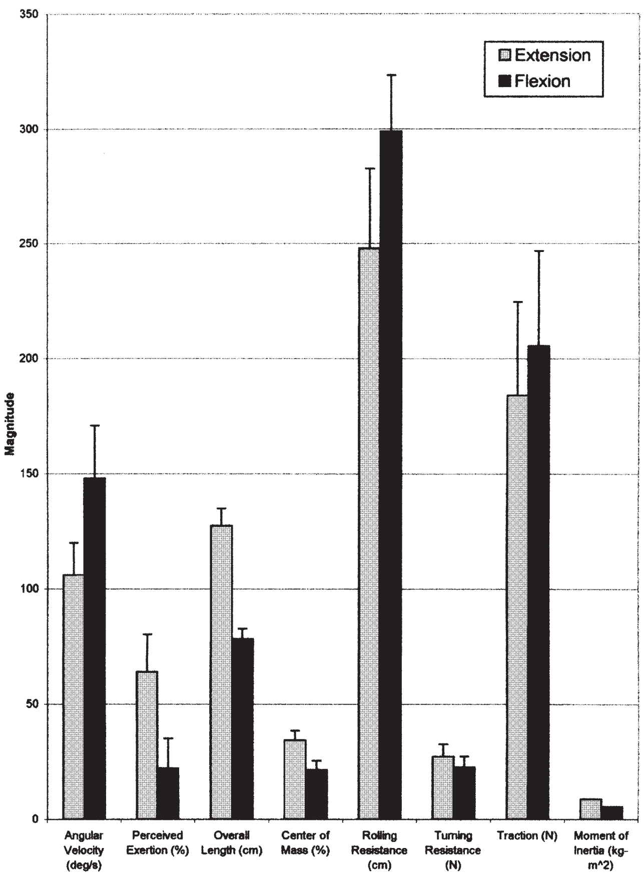

Fig. 2. Comparison of full knee extension and full knee flexion on wheelchair turning parameters. The moment of inertia values shown are the average proportional changes of the 97.5th percentile male and the 2.75th percentile female from 0° to 120° of knee-flexion.

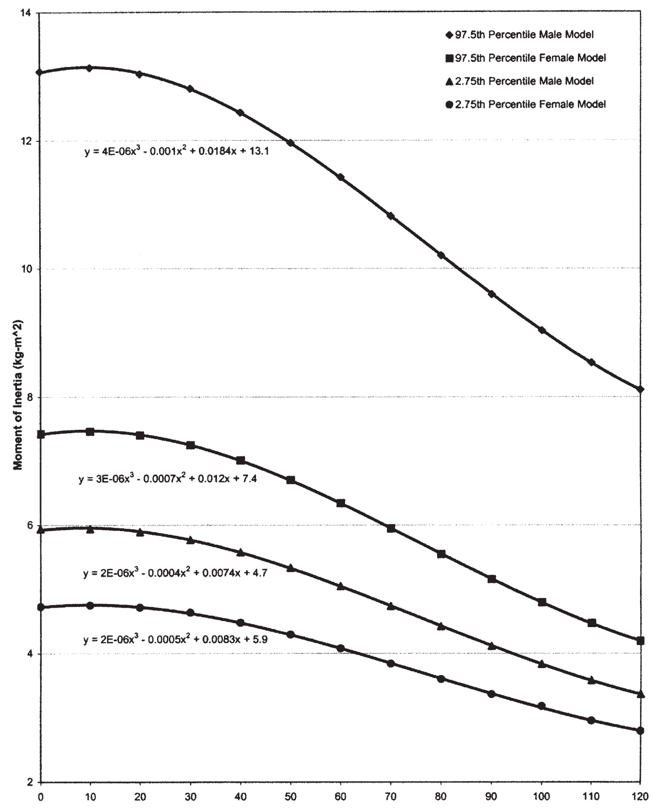

Fig. 3. Moment of inertia for large and small (97.5 and 2.75th percentiles) male and female models from 0–120° of knee flexion in 10° increments. Full knee extension was 0° [8]. The regression equation is given for each curve, where y is the moment of inertia and x is the knee-flexion angle. The R^2 values of all equations were 0.99.

determining turning resistance should be used in future studies.

Although traction was found to have the smallest effect size (12%), the effect of altered knee-flexion position was nonetheless significant. Qualitative analysis of the videotaped angular velocity trials supported this finding, in that rear-wheel slippage was more common when the knees were extended.

The modeling results support the hypothesis that increased knee-flexion angles decrease the wheelchair-user system's moment of inertia about the yaw axis. The relationship between full knee extension and full knee flexion is a non-linear one. The shape of the curves, however, were not exactly as expected. We predicted that the slope of the curve would be minimal initially because early knee flexion from a horizontal position would result in more inferior than posterior movement of the lower leg and foot COM. We then expected the slope to increase progressively until 90° where rearward movement of the segments' COM would be maximal. After 90° of flexion, we predicted that the slope would decrease as further knee flexion increasingly would lead to more vertical rather than horizontal displacement of the segments' COM. Instead, the models showed that the moment of inertia slightly increased initially for three of the models. The unexpected increase in the moment of inertia from 0° to 10° may be explained by the fact that the distance between the COM of the feet and the axis of rotation increases when flexing from 0° to 10°. In this initial stage of flexion, it seems as though the effect of the foot is great enough to exceed or meet the predicted effect of the lower leg.

Although the absolute modeled values are clearly related to the anthropometric characteristics of the wheelchair user, the proportional changes in the total moment of inertia of the wheelchair and user do not seem to be related independently to body size or gender. The slight differences in the proportional changes, ranging from 38%–43.5%, are not clinically significant. We thought that greater proportional changes might be seen with larger body size and the male gender as these models would have larger lower extremities (both in mass and in segment length) relative to the wheelchair. However, this did not appear to be the case.

Limitations of this study included the size and type of sample used. Although the sample was small, it proved to be adequate. We studied young able-bodied individuals primarily because many wheelchair users would not have the range of motion necessary to assume the two knee-flexion positions, or the necessary strength to maintain them. Although the able-bodied subjects received limited manual wheelchair training and would not have been as familiar with wheelchair use as actual wheelchair users, the only active test was spinning in place, minimizing the impact of this limitation. Results using wheelchair users as subjects would be expected to vary more because of differences in lower and upper extremity muscle mass. Only one wheelchair user-position was modeled, and anthropometric data were of able-bodied males and females. The anthropometric data would probably be different from a wheelchair-user population where there may be muscle atrophy in the lower extremities and muscle hypertrophy in the upper extremities.

In addition, only one lightweight wheelchair was used for this study. An area of further research could examine how knee-flexion angle affects turning in various types of wheelchairs, especially sport wheelchairs where rapid turning may be more common. The lighter the wheelchair, the greater the effect should be. Removal of the front rigging also limits the generalizability of the results. However, removal of the front rigging would only have a conservative effect, tending to slightly underestimate the effects.

Finally, the extreme positions tested limit the clinical application of these results. Full flexion and full extension are not as common as angles in between. The moment of inertia models provide some insight into the possible effect of these intermediate angles suggesting that the relationship is a non-linear one. The regression equations allow one to estimate the effect of smaller changes in knee-flexion angle. For instance, changing the knee-flexion angle from 60° to 120° reduces the moment of inertia by 32% in the 97.5th percentile male model. The generalizability of the modeling results are also somewhat limited because the data presented is based on static rather than dynamic models. Further studies need to look specifically at the wheelchair-user population, model different wheelchairs (with the front rigging in place) and model the user in varying positions.

Knee-flexion position may affect other parameters that we did not study, such as the forces applied to the push-rims and the resulting moments, which would have required instrumentation that we did not have.

Knee flexion should also affect downhill turning tendency (DTT), or side-slope effect, the tendency that wheelchairs have to turn downhill due to gravity when placed on a side slope, so that the chair faces forward to roll down the incline. Due to the fact that many outdoor surfaces (such as sidewalks) are engineered with one to two degrees slope for drainage, this problem is a common one [9]. As was the case for the other parameters, knee-flexion angle should affect DTT to the extent that the fore–aft position of the COM is affected.

Similarly, the performance of wheelies should be affected by altering body positioning. Brubaker suggested that popping a wheelie should be easier when the horizontal distance from the COM to the rear wheel axles is decreased [5]. Kauzlarich and Thacker reported that it is harder to pop wheelies in heavier wheelchairs, that have the COM located further forward. They also reported that removing the footrests made it easier to pop a wheelie [20]. This may have been due to the rearward displacement of the COM (due to the removed anterior mass) or because the knees were able to flex. Increased knee flexion probably reduces the rear static and dynamic stability of the wheelchair (although the forward stability probably increases) [21].

Another area of interest would be to see how wearing lower-limb prostheses would affect wheelchair turning by wheelchair users with amputations. Removal of prostheses should reduce the total mass of the wheelchair user, and shift the COM rearward, thereby increasing the ease of wheelchair turning.

However, there are also potential negative effects of increasing the knee-flexion angle. These include the possibility of the casters striking the feet when the casters swivel, increased risk of knee-flexion contractures, decreased venous return, and the development of pressure sores (due to the excessive pressure created where the posterior thighs hang over the anterior seat edge) [1]. An answer to whether the advantages of the hyperflexed knee position generally outweigh the disadvantages will require further study, and any conclusion regarding individual wheelchair users should take into consideration the clinical circumstances.

Despite the study limitations and the need for further study, our findings may have clinical implications. The effect of knee-flexion angle on the ease of turning has not been previously documented in the literature. Knowing that knee position affects the ease of turning suggests that wheelchair prescription could be altered to increase the wheelchair user's maneuverability. This could aid in reducing the prevalence of upper-extremity overuse injuries commonly seen in wheelchair users because the effort needed to turn should be reduced. Our findings support the evolution in wheelchair design towards wheelchairs with front rigging allowing for more than 90° of knee flexion.

5. Conclusion

The ease of wheelchair turning increases as the knee-flexion angle increases. This effect is contributed to by a decreased overall length of the wheelchair-user unit, rearward displacement of the COM position, a decrease in rolling and turning resistance, an increase in traction and a decrease in the moment of inertia. The implications of these findings for wheelchair design and prescription will need to be validated on actual wheelchair users and for smaller increments in knee-flexion range.

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