Limitations of Kinematics in the Assessment of Wheelchair Propulsion in Adults and Children With Spinal Cord Injury
Janet H Bednarczuk and David J Sanderson

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Limitations of Kinematics in the Assessment of Wheelchair Propulsion in Adults and Children With Spinal Cord Injury

Background and Purpose. Recently, there has been a trend for designers to reduce the weight of wheelchairs. Wheelchair performance is frequently evaluated in clinical as well as laboratory settings by kinematic motion analysis. The purpose of this study was to examine the effect of weight on the kinematics of wheelchair propulsion in nonathletic adults and children with spinal cord injury.

Subjects and Methods. The weight of identical new low-weight test chairs (9.3 kg) was manipulated by adding weight (5 and 10 kg) in two matched groups (n=10) of adults and children with spinal cord injury. The three-dimensional coordinates of reflective markers were obtained as the subjects performed level wheeling at a speed of 2 m/s. Results. The pediatric group was found to have significantly lower wheeling speeds than the adult group. The addition of weight, however, did not alter the wheeling speeds in either group. Neither the proportions of the wheeling cycle spent in propulsion (24%) nor the angular (shoulder flexion-extension, elbow flexion-extension, shoulder abduction, and trunk flexion-extension) kinematics of wheeling changed with additions of weight in either group. The angular kinematics of the pediatric group, however, were different than those of the adult group. Conclusion and Discussion. These results indicate that adding weight in the range of 5 to 10 kg did not affect wheeling style under the level-wheeling, low-speed conditions of the study. It is possible that performance in wheelchair propulsion may be more appropriately determined by kinetic and energetic outcome measures than by kinematic measures. [Bednarczyk JH, Sanderson DJ. Limitations of kinematics in the assessment of wheelchair propulsion in adults and children with spinal cord injury. Phys Ther. 1995;75:281–289.]

Key Words: Kinematics, Spinal cord injury, Wheelchair propulsion.

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David J Sanderson

The clinical assessment of wheelchair propulsion is usually based on visual observation of the motion of the user and the chair. Typically, clinical decisions are made regarding wheelchair design and adjustments are based on short exposures of the user to the chair. There is no information available to the clinician to support the validity of this approach to matching the user to the chair. In the laboratory environment, kinematic measurements have been shown to be important.
A kinematic evaluation of the wheelchair propulsion system will affect drag and therefore the energetics of wheeling.6 For a discussion of the differences between mass and weight and the use of kilograms to estimate weight, see Rodgers and Cavanagh.5 There is little experimental evidence, however, to support the belief that low-weight chairs could have an effect that would be evident in routine, everyday wheelchair use. The weight of the user-chair system will affect drag and therefore the energetics of wheeling,7 and a large increase in system (user and/or chair) weight will undoubtedly have an effect on wheeling. It is unclear, however, whether the reduction in the weight of wheelchairs currently available in the marketplace is large enough to have an effect on the wheeling style of the user. It is also theoretically possible that an increase in user weight could have an effect of wheeling performance. Once again, it is unclear at what point a gain in user weight becomes important in routine wheeling performance.

A change in the weight of the chair should theoretically have the greatest effect on the pediatric wheelchair user because, for many such users, their own weight is relatively small compared with that of the chair. Despite their smaller size, muscle power, and weight, pediatric wheelchair users must propel chairs of relatively large weight because most pediatric wheelchairs have comparable weight to wheelchairs for adults.3 There is little information on what effect this might have on the pediatric wheelchair user because much of the published work on wheelchair propulsion has selected for athletic, adult populations of persons with paraplegia.9-11 The purpose of this study was to examine the effect of weight on the kinematics of wheelchair propulsion in nonathletic adults and children with spinal cord injury. The hypothesis was that the addition of weight in the range of 5 to 10 kg would affect the kinematics of wheelchair propulsion by changing both the timing and angular characteristics of wheelchair propulsion. The study design was chosen to reflect real-life wheeling situations, which primarily involve low-speed, level, overground wheeling. Weight additions of 5 and 10 kg were selected because they reflect the range of current commercially available wheelchairs.

**Methods**

**Subjects**

Ten adult subjects with traumatic spinal cord injury (T6-L2 inclusive) were recruited from the Vancouver (British Columbia, Canada) population. They were all experienced wheelchair users who were nonambulatory. They had both complete and incomplete spinal cord injuries. Ten pediatric subjects were recruited through the British Columbia Children’s Hospital meningo(myelo)cele clinic. The pediatric subjects all had spinal cord injuries, complete as well as incomplete, that were the result of neural tube defect (T6-L2 inclusive) at birth and were experienced wheelchair users. Some of the pediatric subjects were able to walk for short distances. All pediatric subjects spent the majority of their day in a wheelchair and had done so for at least 1 year prior to the study. Pediatric subjects with upper-extremity involvement or cognitive deficits resulting from uncontrolled hydrocephalus or other concomitant birth defects were excluded from the study. Children with meningo(myelo)cele were selected because they represent a large group of children with spinal cord injury and because it was not possible to obtain sufficient numbers of children with traumatic spinal cord injuries for the study. Although the neural tube defect that results in meningo(myelo)cele is a multisystem disorder, functionally there are many children who have an isolated paraplegia. These were the subjects who were selected for this study. The characteristics of the two study groups are summarized in Table 1.

<table>
<thead>
<tr>
<th>Group</th>
<th>Age (y)</th>
<th>Weight (kg)</th>
<th>ASIA Scale Score</th>
<th>Arm Length (cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pediatric</td>
<td>n=10</td>
<td>11.3*</td>
<td>37.4*</td>
<td>60.1*</td>
</tr>
<tr>
<td></td>
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<td>2.2</td>
<td>9.9</td>
<td>4.9</td>
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<td></td>
<td>5-17</td>
<td>23.9-54.5</td>
<td>40-60</td>
</tr>
<tr>
<td>Adult</td>
<td>n=10</td>
<td>33.5*</td>
<td>68.5*</td>
<td>56.8*</td>
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<tr>
<td></td>
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<td>8.9</td>
<td>8.7</td>
<td>3.5</td>
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<tr>
<td></td>
<td></td>
<td>22-52</td>
<td>53.7-84.7</td>
<td>52-72</td>
</tr>
</tbody>
</table>

* Asterisk (*) indicates differences between pediatric and adult groups (age and weight, P=.001; arm length, P=.0002).

**Arm Length (cm)**

<table>
<thead>
<tr>
<th>Group</th>
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<td>22-52</td>
<td>53.7-84.7</td>
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</tr>
</tbody>
</table>

**ASIA = American Spinal Injury Association.**

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Table 1. Subject Characteristics

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Note: Table 1 provides subject characteristics including age, weight, ASIA scale score, and arm length for both pediatric and adult groups. Differences are indicated with asterisks (*) for age and weight (P=.001) and arm length (P=.0002).
culation was based on the expectation that similar angular changes would be seen in the present study involving children as have been described in previous studies involving adult subjects.

**Neurological Assessment**

All subjects were assessed by an experienced physical therapist, according to the American Spinal Injury Association (ASIA) scale. This scale has been shown to be rapid, sensitive, and accurate in classifying disability. Each subject was given an ASIA score between 0 and 100 based on an accumulation of selected manual muscle tests of key muscles (C5-L5) from both the right and left sides of the body.

**Matching**

Subjects were matched according to their ASIA scores. There was a need for neurological matching to ensure equivalency in propulsion ability in the two subject groups. Adult and pediatric subjects were considered matched if their ASIA score fell within 4 ASIA scale points of each other. Gender was also included in the matching criteria, with equal numbers of male and female subjects present in each group. All subjects demonstrated symmetry in manual muscle test scores, although this factor was not specifically considered in the matching process. Other factors, such as age, anthropometry, and weight were clearly different in the two groups. Subjects were recruited, assessed, and matched until 10 pairs of subjects were obtained.

**Wheelchair Selection**

A wide variety of manual wheelchairs are commercially available. They have a wide variety of characteristics. With respect to chair weight, however, a local survey of over 100 wheelchairs currently available in North America revealed that the majority of wheelchairs range from 10 to 20 kg in weight, with pediatric wheelchairs having the same weight as wheelchairs for adults. It was also apparent from this survey that the majority of wheel sizes for pediatric wheelchairs were identical to those of wheelchairs for adults. Thus, in this study, the Kuschall Champion 3000 was selected as the test wheelchair because it is a low-weight chair (9.3 kg) that is available in identical styles but in a range of sizes for both adult and pediatric users. The rationale for the addition of 5 and 10 kg to the test chairs was that these weight additions would replicate the range of change in system weight most likely to be experienced by a particular user (either through weight gain or loss to the system). This decision was based on a combination of clinical experience and a survey of the weights of commercially available manual wheelchairs.

A total of seven new Kuschall Champion 3000 wheelchairs, individually matched to the subjects, were provided for use in the study. The dimensions of the test chairs were matched as closely as possible to the subjects' own chairs. The test chairs and subjects' own chairs had a considerable range in seat dimensions from 30.5X30.5 cm (12X12 in) to 38.1X45.7 cm (15X18 in). Foam cushions (5.1 cm [2 in] in depth) were used on all test chairs. Identical wheel types, push-rim sizes, and wheel camber settings were used in all test chairs. Seat height and fore-aft position relative to the wheels were maintained as constant as subject safety would permit, because these factors have previously been shown to affect wheeling performance.

**Weight Additions**

The two weight additions were placed such that they were equally distributed over the contact surface area of the seat of the chair. This was accomplished by use of several straps, which supported either a 5- or 10-kg expandable-size, steel weight placed just beneath the chair seat. This selection of weight placement was made because pilot data indicated that other methods of adding weight to the system produced an increase in the rolling resistance of the chair.

**Laboratory Setup**

A temporary laboratory was set up in a large, wheelchair-accessible gymnasium. Two Panasonic digital video cameras were positioned at an approximate distance of 4 m and at 90 degrees to each other. They were electronically gen-locked in order to provide a synchronous measurement, recorded at 60 Hz of the wheeling movement. (The term "gen-locked" describes the situation in which the shutter-controlling pulse from one camera is used to control the shutter of the other camera, thus ensuring that both cameras take pictures exactly at the same time.) The net error term of three-dimensional motion analysis of a previously calibrated 5X1.5X1.5-m volume of the runway was determined to be 7 mm using direct linear transformation and Peak Performance Technologies software (version 4.0). Two photoelectric cells were positioned at the beginning of this calibrated volume. When a subject passed through the beam between the photo-cells, an event light was triggered, which was visible in both cameras. This event light was used to synchronize the wheeling cycle for analysis. There was sufficient space on the test runway for the subjects to complete three to four wheeling cycles before and after they entered into the calibrated volume. This was necessary to permit the subjects to achieve and maintain a steady wheeling speed through the test space.

The events of the hand contact (grab) on the wheel rim and hand off (re-
lease) were determined by the use of a custom-made hand switch, which was mounted on the distal phalanx of each subject's right thumb. The switch, which responded to pressure by producing an electrical signal, was linked to small light-emitting diodes (LEDs), which were placed at the upper edge of both camera lenses. Whenever contact was made between the subject's thumb and the wheel, the lights of the LEDs were visible in both cameras. In this way, the moment of contact with the wheel (and release from the wheel) was determined by analyzing the videotape. A number of switches were constructed of varying sizes and shapes to provide the best match between the size and shape of a particular user's thumb and the switch. The placement of markers and hand switch are further illustrated in Figure 1.

Data Collection

The subjects were assessed in the laboratory after consent, neurological testing, matching, and wheelchair fitting had been completed. (Refer to earlier sections on neurological assessment and matching for details of methodology.) They were asked to propel their test wheelchairs across the practice runway at a constant speed of 2 m/s. Subjects were given several practice trials at wheeling across the runway. They were given visual and auditory feedback as to their wheeling speed based on readouts from a cycle speedometer mounted on the left wheel of the test chairs. Reflective markers were placed over the right side of each subject's neck (spinous process of C-7); the joint centers of the shoulder, elbow, wrist, hip, knee, and ankle; and the right tire and wheel center (refer to photograph in Fig. 1). The reliability of the hand-switch recordings at every hand contact and release and each subject's comfort level with a particular switch were determined during practice trials. Data collection did not proceed until the subject indicated complete satisfaction with the switch and every contact between thumb and wheel evident to the eye resulted in a signal that was evident in the LEDs placed on the camera lens.

After warm-ups and practice trials, each condition (no added weight, +5 kg, and +10 kg) was presented in a previously determined counterbalanced order. The subjects were asked to complete five pushing trials for each condition. The subjects were given feedback as to their wheeling speed and were closely supervised during each wheeling trial. The first author (JHB) jogged behind each subject during each trial while supporting the hand-switch cables. This procedure appeared to be especially important for the pediatric subjects, who reported relief that they were constantly accompanied during the testing procedure.

Data Analysis

The key features of wheelchair propulsion, as measured kinematically, have been shown to be the angular displacements of the upper extremities and the timing of the propulsive and recovery phases of the wheeling cycle. These features, therefore, were selected for analysis in this study. The propulsive phase of the wheeling cycle is the portion of wheeling when the hands are in contact with the wheel rims and force is being applied by the user to propel the chair forward. The recovery phase of wheeling is when the hands are not in contact with the wheel rims and are returned to the initial wheel contact position.

The first complete cycle after the subject entered the calibrated volume was selected for analysis. The wheeling cycle was defined by one hand contact to the subsequent hand contact, thus consisting of both the propulsive and recovery phases of wheeling. A complete wheeling cycle was defined as 100% of a cycle, and this complete cycle was subdivided into 21 sections at 5% increments.

The positions of the light-reflective markers were digitized from both camera angles and converted to real spatial units by direct linear transformation (Peak Performance Technologies software, version 4.0) and then were smoothed with a fourth-order, low-pass Butterworth digital filter (6 Hz). The angular range of motion data (elbow flexion-extension, shoulder flexion-extension, shoulder abduction, and trunk flexion-extension) were then computed based on link-segment modeling of the three-dimensional marker coordinates derived from the definitions of angles described in Table 2. Each file was time-normalized after the various body joint angles had been computed. Time normalization is a process whereby data arrays from different subjects and trials are linearly interpolated to be the same length. This ensures that each data array has the same number of points between successive hand grabs, which permits the averaging of arrays across trials and subjects. The time-normalized angular files were then within-subject ensemble averaged for each condition.

The subjects wheeled at a target wheeling speed of 2 m/s. The achieved wheeling speeds were deter-
Table 2. Definitions of Angles and Segments

<table>
<thead>
<tr>
<th>Segment</th>
<th>Markers That Define Segments</th>
<th>Ranges of Motion</th>
</tr>
</thead>
<tbody>
<tr>
<td>Upper arm</td>
<td>Elbow-shoulder</td>
<td>0°=full flexion</td>
</tr>
<tr>
<td>Lower arm</td>
<td>Elbow-wrist</td>
<td>180°=full extension</td>
</tr>
<tr>
<td>Trunk</td>
<td>Shoulder-hip</td>
<td>0°=trunk upright</td>
</tr>
<tr>
<td></td>
<td></td>
<td>90°=full flexion</td>
</tr>
<tr>
<td>Abduction</td>
<td>Upper-arm-sagittal plane</td>
<td>0°=arm at side</td>
</tr>
<tr>
<td></td>
<td></td>
<td>90°=full flexion</td>
</tr>
</tbody>
</table>

The three-dimensional linear speed files of the wheel markers were used to determine the achieved, average resultant speeds for each of the trials. This average speed of the wheel marker was then used to determine average wheeling speeds for each subject, for each condition. Ensemble averages of the average wheeling speeds were taken to determine the average wheeling speeds of the two subject groups (adults and children).

The timing of wheel contact and release was used from the events (grab and release) recorded from the LEDs on the videotapes (generated by the hand switch). The percentage of propulsion was determined by taking the percentage of a raw wheeling cycle (defined as one grab to the next consecutive grab) in which the hand was in contact with the wheel (grab to release). This percentage was determined for all five trials for each subject and each condition. Averages across subjects and across conditions were then computed as previously described.

**Statistical Analysis**

The analysis of variance (ANOVA) of the data was performed using the BMDP (P2V) biomedical computer program. Only the propulsive phase of the wheeling cycle, consisting of six increments of the total cycle (0%, 5%, 10%, 15%, 20%, and 25%), was used for the statistical analysis because this was the portion of the cycle of greatest clinical interest. A 2 (groups)×6 (propulsive phase of wheeling cycle)×4 (conditions: own chair, test chair, and test chair with 5 and 10 kg of added weight) repeated-measures analysis of covariance (ANCOVA) was performed of the angular data, with wheeling speed as the covariate. The angular data analyses were performed as previously described. No differences in the outcomes of the significance testing. In all cases, the Greenhouse-Geisser adjusted probability values were used to compensate for any violations of assumptions of sphericity because the epsilon values were low for some of the angular data. For analysis of the subject characteristic data comparing age, weight, ASIA score, and arm length between the adult and pediatric groups, a two-tailed, paired t test was used. Significance was defined by adjusted probability values (P<.05).

**Results**

The weight, age, and anthropometry of the two groups were found to be significantly different (Tab. 1). The ASIA scores were not significantly different between the two groups. There were three female and seven male subjects in each of the two groups. Thus, the selection and matching criteria of the experimental design were met.

The actual wheeling speeds are shown in Figure 2 for the two groups wheeling over the four conditions (subjects' own chair, test chair alone, and test chair with 5 and 10 kg of added weight). Error bars refer to the standard deviations of the average values.
propulsion for the adult and pediatric groups under the four wheeling conditions. Each value is the average of five trials per condition for 10 subjects in each group. "Own" refers to the subjects' own wheelchair. Other conditions refer to the test chair condition with 0, 5, or 10 kg of added weight. Error bars refer to the standard deviations of the average values.

Figure 3. Average percentage of propulsion for the adult and pediatric groups under the four wheeling conditions. Each value is the average of five trials per condition for 10 subjects in each group. "Own" refers to the subjects' own wheelchair. Other conditions refer to the test chair condition with 0, 5, or 10 kg of added weight. Error bars refer to the standard deviations of the average values.

The effect of adding weight on the angular patterns (elbow, shoulder flexion-extension, trunk flexion-extension, and shoulder abduction, in degrees) in the subjects were able to achieve the weight). It is apparent that all of the subjects were able to achieve the nominal wheeling speed of 2 m/s over all of the test conditions. The averaged group wheeling speeds (±SD) were significantly lower for the pediatric group than for the adult group (2.3±0.4 m/s versus 2.4±0.3 m/s). The ANOVA revealed a significant main effect (P=.04) and a nonsignificant interaction effect (P=.12), indicating that although the pediatric group subjects where wheeling at lower speeds than the adult group subjects on average, the two groups responded in a similar fashion in terms of wheeling speeds to the test conditions of weight additions.

The percentage of the wheeling cycle that the subjects spent in contact with the wheel (percentage of propulsion) is shown in Figure 3 for both groups under the four wheeling conditions. The two groups spent comparable proportions of the wheeling cycle in propulsion (pediatric group = 24.5%±7.3%, adult group = 24.4%±7.6%). A repeated-measures ANCOVA of these data, with wheeling speed as the covariate, showed that neither the main effects nor the interaction effects were significant (group, P=.99; condition, P=.54; group×condition interaction, P=.72). Interpretations of these results were that the portion of the time spent in the propulsive and recovery phases was similar in both groups and that both groups showed no alteration in the relative timing of grab and release in response to the test chair or weight additions.

The data in Figure 4 demonstrate the effect of the two weight additions (5 and 10 kg) on the angular kinematic data ensemble averaged across the 10 subjects in the pediatric group. The angular data show very little change in any of the four angular variables between the weight addition conditions. Similar data are presented in Figure 5 for the adult group. The graphs shown in Figures 5 and 6 demonstrate very little difference, regardless of weight addition conditions. An ANCOVA of all of these data was performed over the propulsive phase of wheeling cycle, using wheeling speed as the covariate to correct for the observed differences in wheeling speed. The results confirmed the impression given from the graphical data. The effects were nonsignificant for all of the four angular variables (elbow, P=.21; shoulder flexion-extension, P=.30; trunk flexion-extension, P=.91; shoulder abduction, P=.28) over the weight additions. There was, however, a significant group effect for three of the four variables (elbow, P=.003; shoulder flexion-extension, P=.0007; shoulder abduction, P=.0003). Only the trunk angular data showed a nonsignificant group effect (P=.53), which was perhaps due to the small absolute value and large variability of this variable.

These data also indicate that there were significant differences in the angular kinematics between the two groups. These differences between the angular kinematics of the adult and pediatric groups are summarized in Figure 6. This is the condition of 10 kg of added weight in which differences...
between the wheeling style of adults and children would be expected to be greatest. The pediatric group subjects showed less shoulder abduction and more shoulder and elbow extension than the adult group subjects in the wheeling cycle. Despite these group differences, it is of interest that the group × time × condition effects were nonsignificant for all of the angular variables (elbow, \( P = .62 \); shoulder flexion-extension, \( P = .50 \); trunk flexion-extension, \( P = .58 \); shoulder abduction, \( P = .50 \)), in all of the test conditions, indicating fundamental similarities in the wheeling cycle in both groups.

**Discussion**

This study showed that there was no change in the percentage of the cycle spent in propulsion with 10-kg weight additions in either the adult group or the pediatric group. This finding is surprising given the large differences in weight, age, and anthropometry between the adult and pediatric groups. One possible explanation for these results is that the invariance of the wheeling motion might be indicative of motor control of the upper limbs. This invariance of the wheeling motion is supported by data collection in our laboratory for the same subjects over a 3-year period, as well as by data from other studies. Van der Woude et al. manipulated hand-rim diameter in an adult group and found that there was no effect on cycle time, nor on its subdivisions (push time and recovery time) or the push angle. Rodgers et al. studied 11 male subjects with paraplegia (T5-T11) wheeling to the point of fatigue on a laboratory-instrumented wheelchair. Their study showed no change in the temporal factors of wheeling with fatigue. A second possible explanation for this lack of change of upper-extremity movement might be that the motion is limited by the physical constraints of the chair and the wheels themselves.

Our study showed a mean percentage of propulsion of approximately 25% for both the adult and pediatric groups wheeling overground. These values are slightly lower than other reported values for adult athletic subjects wheeling on an ergometer. Higgins reported a percentage of propulsion of 33.8% for eight elite track and field athletes. Masse et al. reported a mean percentage of propulsion of 33.4% for five elite athletes over a variety of seat positions. Van der Woude et al. reported percentages of propulsion of 30% to 45%, depending on cycle frequency. Sanderson and Sommer reported percentages of propulsion of 43.3%, 34.7%, and 34.3% for three world-class track and field athletes. Cooper reported that approximately 33% of the time was spent in propulsion in a study of five elite athletes. The study by Rodgers et al. showed percentages of push time of 62% to 64%. The differences in push times between those reported by Rodgers et al. and those in our study may be due to the differences in study populations (athletic versus nonathletic) or to the differences in push-rim sizes. We used standard 50.8-cm (20-in) push rims, and Rodgers et al. used 38.1-cm (15-in) push rims. Another possible explanation for the differences in results could be on the basis of differences in wheeling environments. Wheeling speed was 2.5 m/s in our study as opposed to 3.5 m/s in the study by Rodgers et al., and overground wheeling was used in our study as compared with ergometer wheeling.

The purpose of our study was to examine the effect of weight on the kinematics of wheelchair propulsion in nonathletic adults and children with spinal cord injury. The hypothesis was that weight additions (in the range of 5 to 10 kg) would affect the kinematics of wheelchair propulsion by changing both the timing and angular characteristics of wheelchair propulsion. This hypothesis was not supported by the data. There are several explanations for these results. One explanation might be that the effect of weight on wheelchair propulsion could not be measured with the chosen experimental conditions.
The kinematics of wheelchair propulsion might have changed in response to weight, but these effects would only be apparent over more demanding wheeling conditions such as longer time periods and more varied wheeling surfaces. The chosen experimental design was not sufficient to demonstrate the effects of fatigue as demonstrated by Rodgers et al. An other possibility is that weight additions in the range of 5 to 10 kg (reflective of the commercial range of chair weights) were not large enough to cause a measurable change in the wheelchair kinematics. The kinematic assessment of wheelchair propulsion used in this study may not have been the most appropriate method for evaluating wheeling performance. Kinet- ics only describes the final pattern of movement of body parts. It does not reveal anything about what elements may have caused the movements, elements such as work load or power production. Van der Woude et al. describe the factors that influence power production from a mechanical perspective. In the level wheeling situation, it is acceleration of the wheelchair system that will most affect power production. The effect of weight might be more apparent in the first accelerating push from a stopped position, as described by Tupling et al., rather than in the steady wheeling speed condition reported in this article. Thus, it is possible that an assessment of wheeling performance by way of the mechanical and metabolic costs of propulsion would have been more appropriate.

Figure 6. Comparison of the adult and pediatric group angular patterns (elbow, shoulder flexion-extension, trunk flexion-extension, and shoulder abduction, in degrees) for the condition of the test chair with 10 kg of added weight. Mean values are shown for each group (n=10) over a normalized wheeling cycle (five trials per condition). Note that the scale for elbow angle begins at 60 degrees. The entire wheeling cycle from first contact with the wheel (grab) to the subsequent grab (includes both the propulsive and recovery phases of propulsion) is shown. Statistical analysis was performed only on the propulsive phase of the wheeling cycle.

Despite the lack, however, of statistically significant differences of the selected variables of performance in response to the weight additions, differences were found between the adult and pediatric group subjects' wheeling style with respect to the angular kinematics. The pediatric group subjects were required to stretch the generally shorter arms to reach the same size wheels as did their adult counterparts. Thus, it was not surprising to see more shoulder and elbow extension in the pediatric group than in the adult group. The fact that the pediatric group showed less upper-arm abduction than the adult group may reflect differences in pediatric muscle strength rather than anthropometry. Because this study did not include the direct measurement of force on the wheel, it is not possible to do more than speculate on this result.

Clinical Implications

An implication of this work for clinical practice is that a change of 10 kg in system weight as a result of either user or chair weight changes will probably not affect the wheeling motion in short-distance, level wheeling. Clinicians also need to consider that a kinematic analysis of wheeling style under short-distance, level wheeling conditions may not be appropriate in supporting clinical decision making. It may be necessary for subjects to perform fatigue cycling over a variety of wheeling surfaces to demonstrate visible changes in wheeling style. It is also possible that performance in wheelchair propulsion may be more appropriately determined by kinetic measures and measures of energy expenditure than by kinematic measures.

Conclusions

The data presented in this report showed no change in the selected kinematics of wheeling in response to weight additions of 5 and 10 kg in either the adult group or the pediatric group under constant-speed, level wheeling conditions. This finding led us to conclude that weight changes in the range of 5 to 10 kg did not appear to have an effect on the style of wheelchair propulsion. The responses of both groups to the experimental conditions and over the wheeling cycle were not different, indicating fundamental similarities in the kinematics of propulsion in both groups.
Acknowledgments

We acknowledge assistance for the statistical analysis by Dr Han Joo Eom and support from the meningomyelocele team and Dr William Arnold, the director of the meningomyelocele clinic at British Columbia Children’s Hospital. New test wheelchairs were donated for this study by Motion 2000 Inc (Canada) with the assistance of Mr Marco Ferrara. We also acknowledge the contribution of Dr Dirkjan Veeger to the revision of the manuscript.

References

8 Robbins S. The 9th annual survey of the lightweights or how to solve the puzzle and find the right chair for you. Sports ’N Spokes. 1991:23–50.
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