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ORIGINAL RESEARCH

Kinematics and pushrim kinetics in adolescents propelling high-strength lightweight and ultra-lightweight manual wheelchairs

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ABSTRACT

Purpose: The purpose of this study is to describe and compare pushrim forces, propulsive work cost, and upper body kinematics in adolescents propelling (1) a standard high strength lightweight wheelchair, and (2) an ultra-lightweight wheelchair with adjustable main axle positioning, on a level tiled floor ("Tile"), ascending a ramp ("Ramp"), and across a foam mat ("Mat").

Methods: A within-subjects repeated measures study design was used. Eight adolescent manual wheelchair users propelled the standard and ultra-lightweight wheelchairs across the three conditions. Average pushrim tangential force, propulsive power and work per unit distance travelled, as well as upper body kinematic angles, were analyzed.

Results: Average pushrim tangential force (1.80 ± 0.7 N, $p = .042$) and propulsive work per unit distance travelled (8.3 ± 1.7 J·m^{-1}, $p = .002$) were higher for the standard lightweight wheelchair, whereas average speed was lower (0.12 ± 0.03 m/s, $p = .006$). Maximum shoulder ($9.2 \pm 2.0°$, $p = .003$) and elbow flexion ($8.0 \pm 2.2°$, $p = .009$) were higher for the ultra-lightweight wheelchair. Compared with Tile, propulsion on Mat and Ramp was associated with higher average tangential force, work per unit distance, power, and maximum flexion of the neck and trunk, whereas shoulder extension and average speed were lower for Mat and Ramp.

Conclusions: Compared with the standard lightweight wheelchair, ultra-lightweight wheelchair propulsion was associated with lower pushrim forces, lower energy costs, higher self-selected speeds, and increased shoulder and elbow flexion. These variables have been linked to injury risk and mobility efficiency, and the results provided evidence that differences in weight and configuration options are both contributors. Findings can inform decision-making in the prescription of manual wheelchairs for pre-adult populations.

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KEYWORDS
Adolescents; kinetics; kinematics; wheelchair; rehabilitation

➤ IMPLICATIONS FOR REHABILITATION

- A significant proportion of manual wheelchair users are children and adolescents, and due to the early onset of use they may be especially predisposed to the development of chronic overuse injuries.
- The study reports differences in energy costs, pushrim forces, and upper body kinematics measured when adolescents propelled standard and ultra-lightweight wheelchairs across three trial conditions.
- In the ultra-lightweight wheelchair, reduced energy cost is linked to more efficient mobility, and lower forces may be linked to lower risk of chronic injury.
- Significant differences in elbow and shoulder kinematics are also reported, and the findings support the importance of both weight and setup options in the selection of manual wheelchairs.

Introduction

The 2012 Americans with Disabilities Report [1] estimated that approximately 67,000 children under the age of 15 are manual wheelchair users. The high prevalence of upper limb injuries [2–4], as well as neck pain [5,6], has been linked to increased loading and overuse of joint structures resulting from long periods of manual wheelchair activity [5,7,8]. Due to early onset of manual wheelchair use, children and adolescents may be especially predisposed to chronic overuse and long-term injuries [2]. These injuries may severely diminish independence, function, and quality of life during development and later adult life. Therefore, it is important to understand how loading and overuse can be reduced, and how this might be influenced by developmental factors throughout life [9]. These can be affected by variations in kinematics and kinetics resulting from propulsion in different types of wheelchairs and conditions. Although wheelchair propulsion mechanics have been widely studied for adult populations [10,11], there have been fewer studies in children and adolescents who use manual wheelchairs. This leaves an important group of wheelchair users with specific limitations and challenges in their mobility that warrants further investigation.

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A wide range of wheelchairs is commercially available, and the selection of design and choice of materials can dramatically impact the interaction between the user and the wheelchair [12,13]. Wheelchair weight and axle positioning have been studied and shown to affect propulsive efficiency; for example, increasing wheelchair weight has been associated with reduced speed and increased pushrim forces [12,14]. Similarly, positioning the rear axle closer to the rear of the chair has been reported to increase pushrim forces [12], while low seat height has been shown to affect push time and push angle temporal variables [15]. Van der Woude et al. [16] reported that propulsive efficiency could be maximized by adjusting the wheelchair to set a specific elbow flexion angle as measured when participants were sitting at rest.

Current clinical practice guidelines recommend that manual wheelchair users be provided with a fully customizable manual wheelchair made of the lightest possible material [9]. In the United States, the Healthcare Common Procedure Coding System (HCPCS) [17] includes classifications for lightweight wheelchairs coded as K0004 ("High strength lightweight wheelchairs", "K4") and K0005 ("Ultra-lightweight manual wheelchairs", "K5"). These manual wheelchairs are commonly used as long-term mobility devices and have been previously investigated in terms of durability, strength, and stability [18,19]. K5 wheelchairs allow a range of adjustments for multiple features that include axle configuration, wheel camber, and seat and back angles, whereas K4 wheelchairs are heavier and offer fewer configuration options. Since the additional cost of K5 wheelchairs can present an obstacle to insurance reimbursement, understanding the differences in propulsive forces, energy cost, and movement patterns associated with these two wheelchairs is important to financial decision-making in the clinical care of paediatric manual wheelchair users.

Daily wheelchair propulsion can take place on many surface types with different propulsive demands that require biomechanical adaptations from the user [12]. Many previous studies of wheelchair propulsion biomechanics have been conducted only on level floor [15,20]; however, other authors have investigated wheelchair propulsion over different surfaces (e.g. tile, low carpet, high carpet, interlocking concrete paving stones, smooth level concrete, grass, hardwood flooring) [12] and slopes (3°, 6°, 5°, 8%) [12,21]. These studies provided important information about the force patterns involved in wheelchair propulsion on different surfaces and slopes; however, we found no prior reports of propulsion kinematics under such variations of propelling conditions.

The purpose of our study was to investigate upper body kinematics and pushrim forces in a paediatric population during propulsion of K4 and K5 manual wheelchairs across a range of trial conditions. Two challenging conditions – ascent of a ramp and crossing a floor surface intended to represent softer terrain – were studied, in addition to wheeling across level tiled floor. Propulsion forces were collected using an instrumented pushrim wheel, while synchronized data were collected using a wireless motion tracking system. The study's measures included pushrim forces, energy cost metrics, and maximum and minimum upper body kinematic angles. The overall hypotheses for pushrim forces and energy costs were that these measures would be higher when measured in the K4 wheelchair, and higher in the more challenging conditions than in the level tile condition. In addition, we hypothesized that kinematic measures would differ between wheelchair types and between trial conditions.

Methods

Participants

This study recruited participants between 7 and 18 years of age. Inclusion criteria included (1) manual wheelchair as primary means of mobility; (2) at least 1 year of wheelchair use; (3) one year post injury for spinal cord injury (SCI). Exclusion criteria included (1) incoordination impeding propulsion in a steady, straight line; (2) spinal orthosis; (3) past traumatic upper extremity injury affecting wheelchair use.

All research procedures were approved by the Institutional Review Board. Informed consent was obtained from the participant or parent/guardian per age-dependent regulations.

Wheelchairs

Rigid frame K5 (ZRA Wheelchair, Tilite Manual Wheelchairs, Pasco, WA) and folding frame K4 (9000XT Wheelchair, Invacare Corporation, Elyria, OH) wheelchairs in one of three sizes (seat width × depth = 14 in. × 16 in.; 15 in. × 17 in.; 18 in. × 18 in.) were individually set up by an assistive technology professional (author S. B.). Subject specific set up was performed by adjusting the following configuration features: backrest angle relative to the seat (through the backrest hardware), footrest height, and seat depth. K4 wheelchairs were heavier and had fixed, level seating, whereas the adjustable axle in the lighter K5 wheelchair provided a lower seating position overall (Table 1).

Instrumentation

An instrumented pushrim wheel (SmartWheel, Out-Front, Mesa, AZ) was used to record right side kinetics at 240 Hz. Kinematic data were collected using an Inertial Measurement Unit (IMU) system (MVN Biomech, Xsens Technologies BV, Enschede, The Netherlands) [22,23] at 60 Hz. Eleven sensors were placed on the head, sternum,

Table 1. Specifications for the wheelchairs tested.

Specification	K4	K5
Model	Invacare 9000XT	Tilite ZRA
Seat width × depth (in.); weight (kg)	14 × 15 in.; 18.5 kg 16 x 17 in.; 18.8 kg 18 x 18 in.; 18.9 kg	14 × 15 in.; 13.7 kg 16 × 17 in.; 14.1 kg 18 × 18 in.; 14.7 kg
Front seat to floor height (in.)	17.5	17.5
Horizontally and vertically adjustable wheel axle (rear seat height)	No	Yes
Adjustable seat back angle	Yes	Yes
Casters type/height (in.)	6 in. composite urethane 3 spoke	6 in. lite speed plastic wheel with soft roll casters
Front Frame Angle[a]	70	80
Rear wheel spacing (in.)	1.125	1.25
Camber (°)	0	4
Seat width taper	0	0
Footrest taper	0	0

[a]Front Frame Angle is measured between the descending front tube of the wheelchair frame and the floor, as configured for testing. The Front Frame Angle is less than 90° when the bottom of the frame is further forward than the top.

Figure 1. Participant and wheelchair setup, showing IMU sensors and instrumented pushrim. The supplied elastic straps, tight Lycra^{TM} suit top and gloves were used to fix the sensors according to the system manufacturer's protocol.

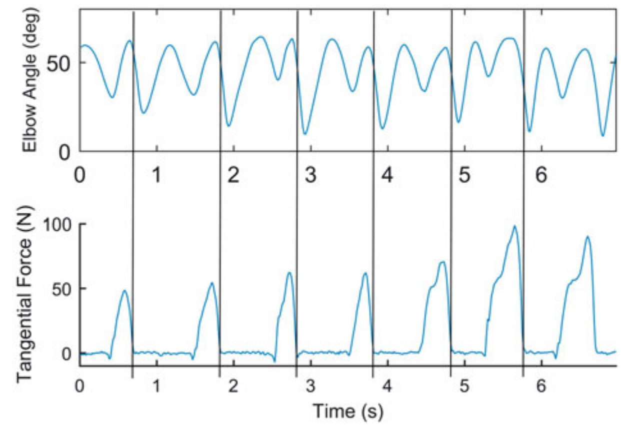

Figure 2. Sample synchronized force (FT) and kinematic (EL) data from a ramp ascent trial. Vertical lines show the start and end of each propulsion cycle, representing the moment of pushrim release as detected from the tangential force signal.

sacrum, scapulae, upper arms, forearms, and hands (Figure 1). A sensor ("WCIMU") was attached to the wheelchair frame to detect vertical acceleration and inclination angle. Synchronization of the systems was achieved by hardwired connection. Only right limb/side data are reported in the current study.

For kinematic calibration, participants were seated with arms abducted to shoulder height, with elbows extended and palms down. Some participants were provided active assistance under the guidance of a physical therapist (author S. B.), due to contractures or spasticity involving the elbows; any remaining elbow flexion was recorded.

Testing conditions and protocol

Testing order (wheelchair, condition) was randomized with trials completed in the first wheelchair before changing to the second. Participants were allowed to acclimate to the wheelchair, and were asked to propel at a steady, comfortable speed and perform 3 repetitions of each condition. All trials took place in a tiled paediatric hospital hallway. Participants initiated all trials from a stationary position: for "Tile", participants propelled for 10 m across level floor; for "Mat", they propelled the wheelchair for approximately 5 m before traversing a 2.2 m polyfoam mat representing outdoor grassy terrain; for "Ramp", participants propelled the wheelchair for approximately 10 m before ascending a 4.8 m long ramp. The ramp slope increased progressively until reaching a 1.2 m section of constant slope (5°) and reduced slope progressively thereafter.

Data analysis

Portions of the synchronized data were selected from each trial, and only complete propulsion cycles were analyzed. Cycle boundaries were defined at pushrim release by the tangential force dropping below 4 N (Figure 2). The selection periods were Tile – 7 m of steady travel following 3 m considered as initial acceleration; Mat – mat length (2.2 m) following contact with the mat detected from WCIMU vertical acceleration; Ramp – 3 m following the start of the constant slope portion detected from WCIMU vertical inclination. Ground distance was calculated by time integration of the pushrim-measured wheel speed. Ten kinetic and kinematic measures were analyzed (Table 2), with values taken as the mean across propulsion cycles and repetitions.

Two propulsive work rate measures were calculated. Work was calculated by time integrating propulsive power (product of pushrim measured torque and angular velocity). *Average power* (P_{avg}) and *propulsive work cost* (W_d) were calculated by dividing by elapsed time and distance travelled, respectively.

Statistical analysis

For each measure, a two-way repeated measures analysis of variance with significance of 0.05 tested for significant effects of wheelchair and trial condition. When significant effects were found, Bonferroni *post hoc* comparisons were used to determine pair-wise differences.

Results

Eight adolescents between 12 and 18 years of age with a range of diagnoses were enrolled (Table 3). Descriptive statistics and statistical test results are presented (Table 4, and Figures 3 and 4). Quantitative data on group values and differences (Δ) are reported below in the format "mean ± standard error".

Tangential pushrim force (FT_{avg}) for the K4 wheelchair was higher ($p = .042$, $\Delta = 1.80 \pm 0.7$ N). Within trial conditions, FT_{avg} showed a small contradictory finding for Tile (K5 > K4, $p = .016$). A main effect of trial condition ($p < .001$) was also recorded, with both Ramp ($p < .001$) and Mat ($p < .001$) average tangential forces higher than Tile.

Propulsive work cost (W_d) was higher for the K4 wheelchair ($p = .002$, $\Delta = 8.3 \pm 1.7\ J{\cdot}m^{-1}$). Similar to the contradictory finding for FT_{avg}, W_d was higher for the K5 on Tile ($p = .027$). A main effect of trial condition ($p < .001$) was also recorded for this measure, with W_d for both Ramp ($p < .001$) and Mat ($p < .001$) higher than Tile. W_d for Ramp was lower than Mat ($p = .029$).

The only main effect recorded for P_{avg} was of trial condition ($p = .017$). *Post hoc* tests revealed that average power on Tile was lower than on Ramp ($p = .003$). Average speed (v_{avg}) was higher for the K5 wheelchair ($p = .006$, $\Delta = 0.12 \pm 0.03$ m/s). Within trial conditions, speed was higher for the K5 on Ramp ($p = .023$) and Mat ($p = .003$). A main effect of trial condition ($p < .001$) was also recorded with both Ramp ($p < .001$) and Mat ($p < .001$) speeds slower than Tile, while Mat speed was significantly slower than Ramp ($p = .008$).

Table 2. Kinetic and kinematic output measures.

	Measure
V_{avg} (m·s^{-1})	Average speed, measured by the instrumented pushrim
Ft_{avg} (N)	Average tangential force, calculated by dividing the instrumented pushrim axial propulsion moment by the pushrim radius [10]
SL_{max}/SL_{min} (°)	Maximum/minimum right shoulder flexion, measured from the IMU system's biomechanical model
EL_{max}/EL_{min} (°)	Maximum/minimum right elbow flexion, measured from the IMU system's biomechanical model
NK_{max}/NK_{min} (°)	Maximum/minimum neck flexion, measured as the C1/head angle from the IMU system's biomechanical model
TK_{max}/TK_{min} (°)	Maximum/minimum trunk flexion, measured as the sum of the flexion angles for all spinal articulations from the inertia measurement unit's biomechanical model (L5-S1, L4–L3, L1–T12, T9–T8, T1–C7)

Maximum shoulder flexion (SL_{max}) was higher for the K5 wheelchair ($p = .003$, $\Delta = 9.2 \pm 2.0°$). Within trial conditions, SL_{max} was higher for the K5 wheelchair on Tile ($p = .002$), Ramp ($p = .034$), and Mat ($p = .001$). A main effect of wheelchair type on maximum elbow flexion (EL_{max}) was also recorded ($p = .009$). EL_{max} was higher for the K5 wheelchair ($\Delta = 8.0 \pm 2.2°$); within trial conditions, EL_{max} was significantly higher for the K5 wheelchair on Tile ($p = .29$), Ramp ($p = .006$), and Mat ($p = .009$).

A main effect of trial condition on minimum shoulder flexion (SL_{min}) was recorded ($p = .036$), with Ramp ($p = .156$) and Mat ($p = .186$) trending less negative than Tile. Since negative shoulder flexion represents shoulder extension, this indicated less shoulder extension in Ramp and Mat. A main effect of trial condition ($p = .022$) on maximum neck flexion (NK_{max}) was recorded. NK_{max} was higher on Mat than on Tile ($p = .046$). Also, a main effect of

Table 3. Characteristics of study sample.

Subject	Age (years)	Gender	Weight (kg)	Height (cm)	Diagnosis/ROM[a] limitations
1	17	M	79.3	175.2	CP
2	18	M	47.1	174.8	SCI (T6)
3	16	M	40.8	154.9	CP/right elbow contracture (10°)
4	16	F	42.6	154.9	CP/right elbow contracture (10°)
5	15	M	45.3	130.1	Spina Bifida
6	12	M	35.0	160.0	SCI (T5)
7	12	M	47.6	154.9	Spina Bifida
8	15	M	49.9	157.0	Friedrich's Ataxia
Mean ± SD	15.1 ± 2.2		48.5 ± 13.3	157.7 ± 14.1	

CP: cerebral palsy; SCI: spinal cord injury.
[a]Range of motion with assistance

Table 4. Descriptive statistics (mean ± SD) and statistical test results for the study's kinetic and kinematic measures.

		TILE		MAT		RAMP	
	Within-subjects effects	K4	K5	K4	K5	K4	K5
		b,c		a,c		a,b	
W_d (J·m^{-1})	*,†	15.3 ± 5.78§	19.7 ± 7.86§	105.6 ± 27.3	85.2 ± 20.1	82.4 ± 21.8	73.4 ± 23.6
		c				a	
P_{avg} (W)	*	18.5 ± 10.6	21.5 ± 9.27	32.4 ± 16.8	38.7 ± 24.1	37.9 ± 15.0	35.8 ± 12.8
		b,c		a,c		a,b	
V_{avg} (m·s^{-1})	*,†	1.20 ± 0.27	1.29 ± 0.17	0.44 ± 0.21§	0.61 ± 0.25§	0.71 ± 0.35§	0.82 ± 0.34§
		c, b		a		a	
Ft_{avg} (N)	*,†,‡	4.0 ± 1.47§	5.3 ± 1.87§	29.0 ± 6.94	25.4 ± 4.53	26.2 ± 5.43	23.2 ± 4.58
SL_{min} (°)	*	−41.7 ± 25.1	−36.7 ± 23.8	−32.9 ± 30.1	−33.0 ± 26.7	−34.6 ± 30.4	−31.6 ± 30.8
SL_{max} (°)	†	40.6 ± 12.0§	50.5 ± 11.6§	42.4 ± 14.3§	51.0 ± 13.3§	45.8 ± 12.7§	54.9 ± 12.1§
EL_{min} (°)		10.9 ± 12.6	11.0 ± 10.6	14.7 ± 12.8	30.8 ± 37.6	7.0 ± 12.4	8.6 ± 9.57
EL_{max} (°)	†	65.4 ± 21.0§	72.2 ± 17.4§	65.2 ± 22.0§	73.5 ± 19.4§	63.0 ± 21.9§	72.4 ± 19.6§
NK_{min} (°)		−5.4 ± 10.01	−5.3 ± 8.31	1.7 ± 10.81	−2.9 ± 10.98	−0.7 ± 13.09	−7.4 ± 7.26
		b		a			
NK_{max} (°)	*	4.4 ± 7.75	7.9 ± 8.94	12.7 ± 7.69	13.7 ± 6.6	9.6 ± 7.5	9.7 ± 5.2
TK_{min} (°)		20.7 ± 11.8	22.3 ± 8.9	21.6 ± 16.6	27.5 ± 14.6	24.6 ± 12.9	25.6 ± 14.8
		c, b		a		a	
TK_{max} (°)	*	32.2 ± 10.6	33.9 ± 12.5	41.2 ± 9.38	46.1 ± 17.9	42.8 ± 14.0	43.1 ± 17.3

*Statistically significant main effect of propelling condition ($p \le .05$).
†Statistically significant main effect of wheelchair type ($p \le .05$).
‡Statistically significant interaction condition × wheelchair ($p \le .05$).
[a]Statistically different from TILE ($p \le .05$).
[b]Statistically different from MAT ($p \le .05$).
[c]Statistically different from RAMP ($p \le .05$).
§Statistical differences between wheelchairs within propelling condition ($p \le .05$).

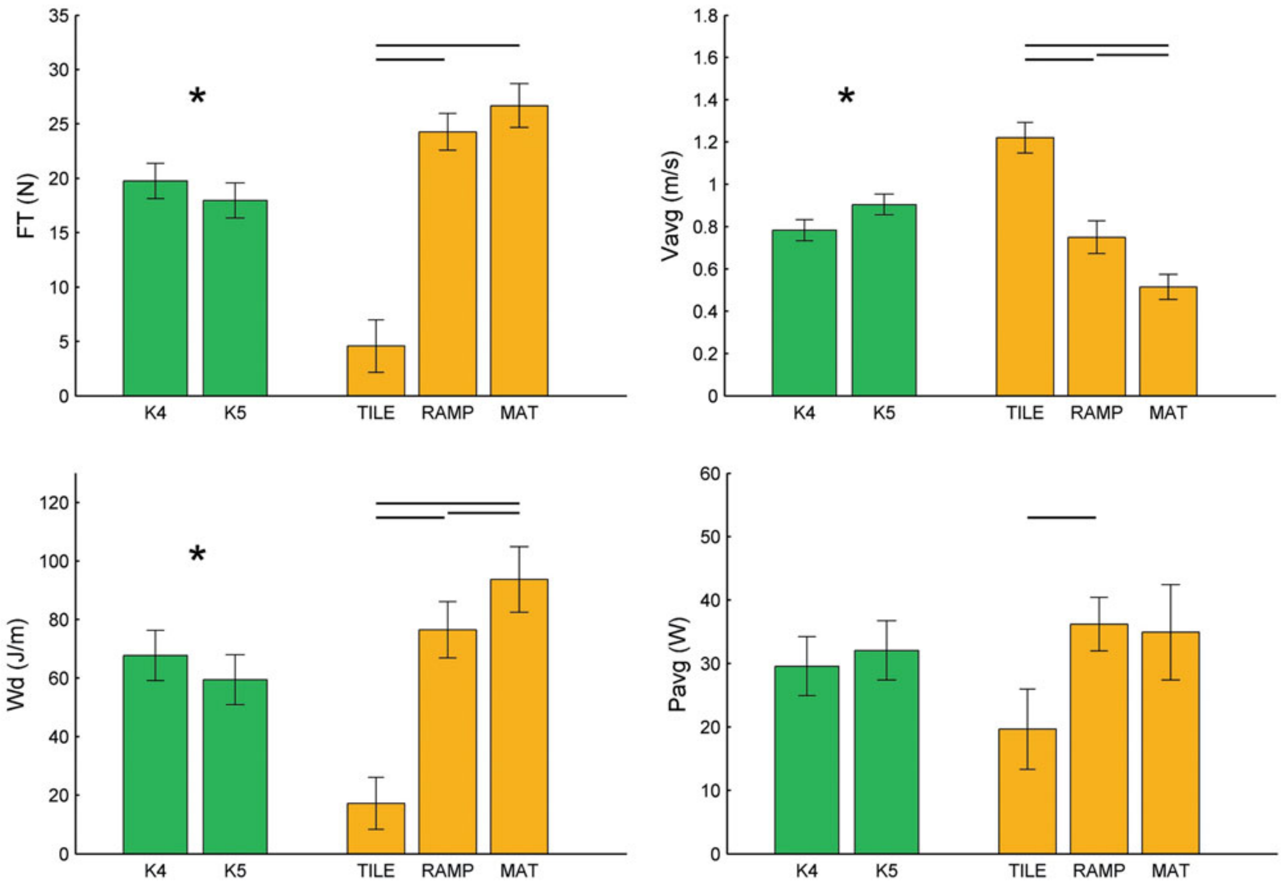

Figure 3. Marginal means for average tangential force (FT_{avg}), average speed (V_{avg}), propulsive work cost (W_d), and average power (P_{avg}) across wheelchair and propelling condition. Error bars represent within-subject 95% confidence intervals [28,29].

trial condition on maximum trunk flexion (TK_{max}) was recorded ($p = .004$), with *post hoc* tests revealing that maximum trunk flexion was significantly lower on Tile than on Ramp ($p = .018$) and Mat ($p = .026$).

Discussion

This study investigated standard lightweight (K4) and ultra-lightweight (K5) wheelchair propulsion in adolescents across a range of conditions. Lower forces and propulsive work cost were associated with the K5 wheelchair, and several kinematic differences were observed. The finding that propulsion forces were higher for the K4 wheelchair supported our hypothesis. Across all conditions, forces in the K4 wheelchair were an average of 10% higher, and excluding the Tile condition, K4 forces were an average of 13.7% higher. The K4 was the heavier wheelchair, and these findings were consistent with a previous report [12] that adding weight to a standard wheelchair increased forces. The same study also reported that anterior positioning of the rear axle was associated with lower forces, and while we did not study weight and configuration independently, the more anteriorly located rear axle of the K5 wheelchair in our study may have also contributed to the reduction in forces. Potential mechanisms for this are a lower overall rolling resistance associated with more posterior weight distribution and reduced loading of the front casters, and as discussed further below, alterations to propulsion kinematics associated with differences in seating position. Previous studies have demonstrated that higher pushrim forces require elevated muscle and joint loading [24–26], which has been suggested to increase risk of musculoskeletal injury [24].

Our study also found that the propulsive work cost was higher in the K4 wheelchair. Across all conditions, propulsive work cost in the K4 was an average of 13.9% higher, and excluding the Tile condition, was an average of 18.5% higher. These findings are consistent with the K4's higher weight and consideration of the test conditions: in Ramp, the work to increase potential energy increases with weight, whereas for Mat the loss of energy due to rolling resistance is increased with weight. As a ratio of cost (propulsive work cost) to benefit (distance travelled), results from this measure can be compared with studies on efficiency. Having found no reports of efficiency in adolescents, we reviewed studies in adults, acknowledging that this limits our ability to compare with the results and mechanisms described in the following. Beekman et al. [14] reported higher physiological work (oxygen cost per distance travelled) when adults with paraplegia propelled standard wheelchairs than when they propelled ultralight wheelchairs. Similarly to the contrast between the K4 and K5 wheelchairs we studied, the standard wheelchair had a folding frame and was heavier than the rigid frame ultralight wheelchair. In a study in able-bodied adults, Van der Woude et al. [16] reported the effects of seat height on gross mechanical efficiency (GME), defined as the ratio of energy expenditure to the external power output measured during propulsion on a treadmill. Lowering the seat height increased GME, and the lower seat height of the K5 wheelchair may have contributed to the difference we observed. Although we did not detect an overall difference between wheelchairs in average power, average speed was significantly higher in the K5, with a mean difference across conditions of 15.6%. An interpretation that accounts for the observed relationships in propulsive work cost, power, and speed is that participants tended to

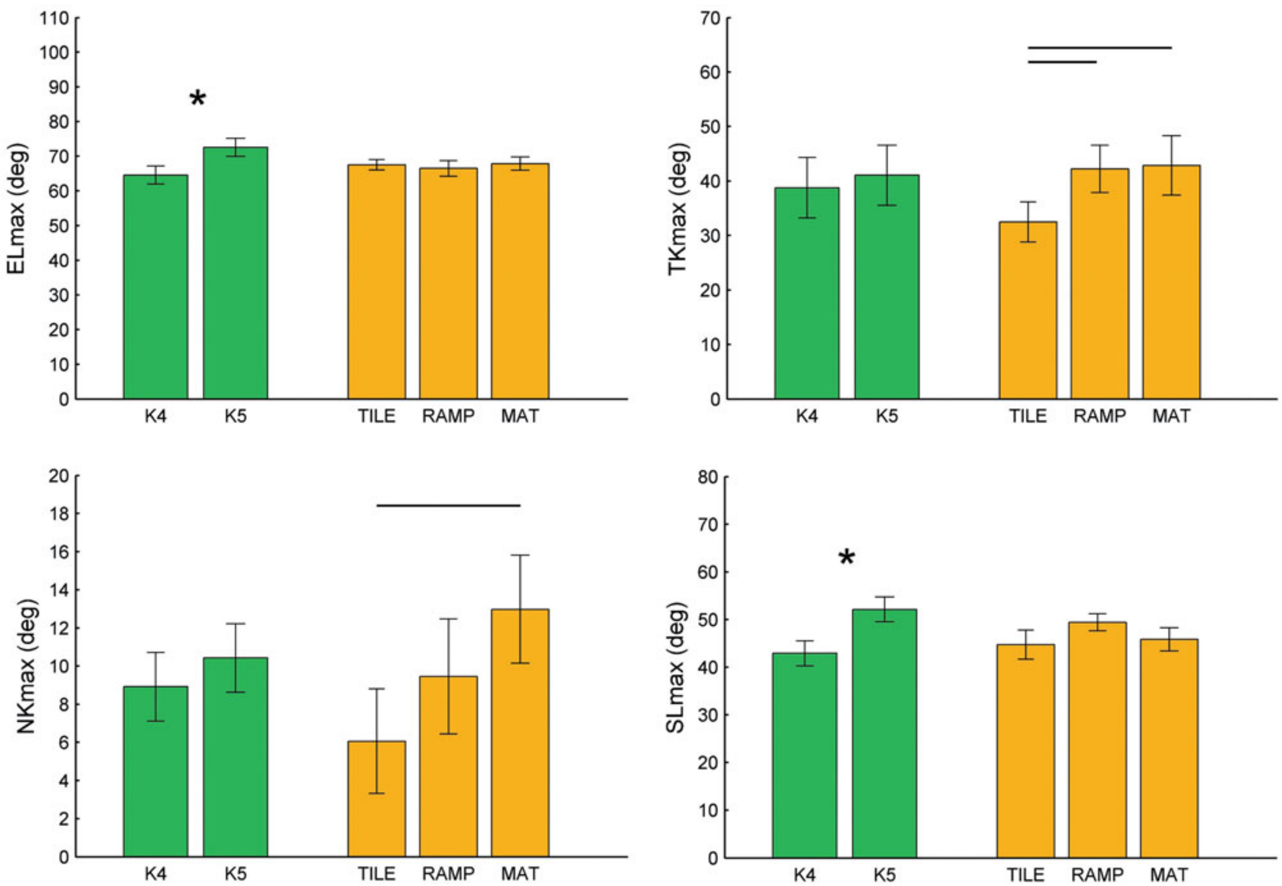

Figure 4. Marginal means for maximum elbow flexion (EL_{max}), maximum trunk flexion (TK_{max}), maximum neck flexion (NK_{max}), and maximum shoulder flexion (SL_{max}) across wheelchair and propelling condition. Error bars represent within-subject 95% confidence intervals [28,29].

maintain similar operating power levels across the two wheelchairs. Overall, our results provide evidence that the K5 wheelchair allows for increased mobility in terms of increased speed of travel and decreased energy cost.

As hypothesized, differences in upper extremity kinematic patterns were observed between the wheelchair types. Larger maximum shoulder flexion angles were observed in the K5 wheelchair, with a group difference across conditions of $9.2 \pm 2.0°$. Greater elbow flexion was also observed in the K5 wheelchair, with a group difference of $8.2 \pm 2.2°$. Since minimum flexion angles were not significantly different, this appears to represent greater functional range of motion with the K5 wheelchair. In able bodied adults, Van der Woude et al. [16] observed that lowering seat height was associated with a shift of functional range towards increased reach of the hands with respect to the rim during push phase, resulting in increased maximum elbow flexion and shoulder extension, and decreased maximum elbow extension and trunk flexion. This accompanied the increase in GME described above. Although we found no statistically significant effects of wheelchair type on neck and trunk flexion angles, ranges of motion tended to be more flexed with the K5. However, results suggest that more flexed positions were associated with higher effort activities, while the K4 wheelchair was associated with higher propulsive cost, as discussed above. Therefore, it seems likely that the shift in operating range of neck and trunk resulted more from setup differences than weight difference.

Also consistent with our hypotheses, marked differences between propulsion conditions were recorded. Propulsive force, propulsive work cost, and average power trended higher in the Mat and Ramp conditions, whereas Tile speeds were highest. Cowan et al. [12] studied wheelchair propulsion across a similar range of conditions and reported that self-selected speed was lowest on the ramp and highest on tile, whereas push forces increased in the opposite order (tile < carpet < ramp). Our study recorded larger maximum trunk and neck flexion for Ramp and Mat than for Tile. Increased flexion might reflect strategies to meet the higher demands of the Ramp and Mat conditions. Similarly, Julien et al. [27] observed increased upper trunk and neck flexion for individuals with tetraplegia when propulsive demands were increased by propelling at higher speeds. For our study, several factors might explain these kinematic changes: (1) when ascending a ramp, flexing the trunk might improve balance and stability during the push phase; (2) upper body momentum created from trunk flexion might assist in counteracting the push forces, perhaps allowing higher propulsive forces; (3) trunk flexion might allow prolonging of pushrim contact. Our study also recorded less maximum shoulder extension when propelling the wheelchair during demanding conditions. This might suggest difficulty in generating high propulsive forces in shoulder extension, as well as decreased stroke length and increased frequency to maintain momentum. Moreover, decreased stroke length and increased frequency during challenging conditions might provide the user with better wheelchair control and responsiveness for a better sense of safety. Our findings highlight the importance of testing wheelchair propulsion across conditions, and the increased functional ranges of motion in challenging conditions underscore the importance of avoiding movement restrictions in manual wheelchair design.

Our study has several limitations. First, although our recruitment age range was 7–18 years old, no participants younger than

12 were enrolled. This might limit the applicability of findings to the adolescent age group. Second, while we chose K4 and K5 wheelchairs to be typical of models prescribed for full-time paediatric use within the United States healthcare system, it is naturally important to consider individual differences in extrapolation of findings to other makes and models. Third, we allowed short-term practice periods in each wheelchair, whereas longer term acclimation might have impacted the study results. Based on observations during testing and discussion with participants, we believe that effects are driven primarily by differences between the wheelchairs and the test conditions, and that additional practice would not have critically changed the study's findings. Further, the IMU system zeroed the joint angle measurements in a standard calibration position that two participants could not fully adopt due to range of motion deficits. This is a source of systematic error; for instance, elbow flexion measurements could read lower because the participant's elbows could not be fully extended for calibration. While this did not affect our ability to test our within-subjects hypotheses, it should be taken into account when interpreting the quantitative values reported and comparing them with other studies. Finally, we note another methodological limitation: since we investigated only right side propulsion, we were unable to evaluate asymmetry such as might result from left/right differences in both limb dominance and clinical presentations such as motor weakness.

Today, manufacturers provide a wide range of manual wheelchairs from which clinicians can prescribe a specific model to meet the patient's functional needs, within any economic constraints that may be present. Our study has provided information on the differences in pushrim forces, energy costs, speeds, and upper body kinematics between representative standard lightweight and ultra-lightweight wheelchairs. The use of the ultralightweight wheelchair was associated with (1) lower pushrim forces that may reduce long-term risk of injury, (2) lower propulsive work costs in challenging terrain, and (3) higher self-selected speeds of propulsion. This study also added to the body of evidence that, in addition to lighter weight, differences in configuration options differentiate the effectiveness of manual wheelchairs. It is suggested that this evidence be used in the decision-making processes surrounding the prescription and purchase of manual wheelchairs for adolescent populations.

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