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WHEELCHAIR PROPULSION TECHNIQUE AT DIFFERENT SPEEDS

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ABSTRACT. To study wheelchair propulsion technique at different speeds, five well-trained subjects propelled a wheelchair on a treadmill. Measurements were made at four belt speeds of 0.56–1.39 m/s and against slopes of 2 and 3 degrees. Cardiorespiratory data were collected. Three consecutive strokes were filmed. Using markers on subject, wheelchair and treadmill frame a kinematic analysis was performed. Considerable inter-individual differences in propulsion style were found, but also general changes relative to speed occurred in the group as a whole. Cycle time decreased with speed, predominantly as the result of a shorter push time while push angle remained constant and the movement ranges of trunk and arms shifted with speed. It is concluded that despite different propulsion styles, general and continuous adaptations to speed changes occurred, mainly by flexion of the trunk and arms.

Key words: wheelchairs, wheelchair athletes, treadmill, propulsion technique, speed.

The cyclic pattern of movement in conventional (hand rim) wheelchair propulsion can be divided into two phases together forming one complete cycle: (*a*) the push or contact phase, during which the person's hands are in contact with the rim and supposedly exert force to the time and (*b*) the recovery phase during which the arms are brought back to the position in which a new propulsive phase can begin. Stroke frequency is defined as the number of those cycles per unit time.

Changes of speed of driving are related to changes in stroke frequency and in the interrelation of push and recovery time (1, 5, 6, 9). It is expected that changes in the movement pattern related to speed changes may occur also (6). Sanderson & Sommer (4) in their study on the propulsion techniques of three wheelchair athletes concluded that between subjects considerable freedom appeared to exist in choosing a style of wheelchair propulsion. The variation in style within subjects was found to be small, but the 'circular' propulsion technique in which the driving hands stay close to the rim during the recovery phase, which was seen in some of these fast driving athletes, was suggested to be favorable (4). Since adaptation to speed changes may well be related to different propulsion styles, it may be difficult to make general statements on adptation to speed, irrespective of propulsion technique.

This study describes some variations of the pattern of movement during wheelchair driving with different propulsion speeds at two different slopes on a motor driven treadmill. Starting from a description of some inter-individual differences in technique, an attempt is made to describe changes of selected propulsion cycle parameters related to belt speed, for the group of subjects as a whole. These results were compared to those of mechanical efficiency, which was calculated by measuring energy expenditure and external power.

METHODS

Five male wheelchair athletes participated in an exercise test on a motor driven treadmill (Enraf Nonius model 3446, belt width 1.25 m, length 3.0 m). Physical characteristics of the subjects are presented in Table I. Subjects were either marathon racers and/or basketball players and had no upper extremity problems. Subject KB, who is able-bodied, participates in wheelchair marathon racing. The maximum work capacity (watts $\times$ $kg_{body\ mass^{-1}}$) as attained by the subjects during the present or previous experiments is shown in Table I as well. Exercise tests were performed using a basketball wheelchair. Power output during wheelchair propulsion was assessed with a drag test (8). In two sessions, with the treadmill at a two or three degree slope, belt speed increased every three minutes from 2 to 3, 4 and 5 km/h (0.56, 0.83, 1.11, and 1.39 m/s). During the third minute of every speed step, expired gasses were collected in Douglas bags. Gasses were

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Abbreviations: ME = mechanical efficiency, SA = start angle, EA = end angle, PA = push angle PA = EA − SA, CT = cycle time CT = PT + RT, PT = push time, %PT = percentage push time %PT = PT/CT × 100%, RT = recovery time, ANOVA = analysis of variance, WS = work per stroke WS = external power output × CT, EP = end of push.

Table I. *Subject data*

Subject	Age (yrs)	Body weight (kg)	Sitting height (m)	Max. power output (W kg^{-1})	Impairment
KB	30	70.1	0.942	1.29	Able-bodied
KW	39	77.7	0.947	1.42	Ankle ankylosis
PW	45	70	0.900	1.0	Spinal lesion L4/L5
MB	35	82.8	0.934	1.02	Spinal lesion T11/T12
RO	31	85.7	0.957	1.02	Spina bifida
Mean	36	77.25	0.936	1.15	

analysed for oxygen and carbon dioxide (Mijnhardt UG61, paramagnetic; Mijnhardt UG51, infrared). Gross mechanical efficiency (ME) was calculated as the ratio of external power output and energetic cost (×100%).

During the same period at least three complete strokes were filmed (DBM-55, Teledyne camera systems, 60 f/s, Kodak 4XR film). Markers on film were analysed on a Summagraphics Supergrid digitizer (accuracy 1/40 mm). Data were then filtered using a second order Butterworth filter (cut-off frequency 5 Hz) and differentiated in order to obtain velocity and acceleration (2).

Markers were placed on the treadmill frame, the wheel axle and on the subject (acromion, lateral humeral epicondyle, radio-carpal joint and the distal end of the third metacarpal).

For each block of data, linear displacement and velocity of the wheelchair relative to a fixed origin were calculated. Start angle (SA) and end angle (EA) of the push range were calculated relative to the horizontal through the rear wheel axle and were defined as the vector from the rear axle to the marker on the hand. Push angle (PA) is defined as the difference between SA and EA.

Timing parameters cycle time (CT), push time (PT) and recovery time (RT) were estimated on basis of the number of film frames and film speed. Work per stroke (WS) was calculated as the product of external power output and cycle time.

Both the angle of the trunk-to-wheel axle vector as indicator for trunk and shoulder positions, and the angles of the upper arms projected in the sagittal plane were calculated relative to the vertical. Elbow angles were calculated three-dimensionally, based on the estimation of the displacement of upper arm and forearm markers in the frontal plane through the ratio segment length/projected length (3) and subsequent calculation of the scalar vector product of the relative coordinates of wrist, elbow and upper arm.

Minimum and maximum values of the parameters were statistically analysed by a two way analysis of variance (ANOVA) repeated measures procedure ($N=5$ subjects, 4 speeds versus 2 slopes).

RESULTS

The subjects participating in this study worked at a mechanical efficiency (ME) of around 10% (mean 10%, SD 1.1%). ME increased significantly with belt speed from 9.4% at the lowest speed up to 10.5% at a speed of 1.39 m/s.

The propulsion styles of subjects showed considerable differences. Fig. 1 shows an example of the movements of segment markers of trunk, elbows, wrists and hands for two subjects in a two-dimension-

Table II. *Results of a two-factor ANOVA with repeated measures ($N=5$)*

	Slope (1, 4)	Speed (3, 12)	Slope× speed (3, 12)
Parameter (df)			
ME	2.4	4.4*	0.7
PT	0.0	77.0**	0.0
RT	48.0**	6.0**	0.0
CT	21.3**	31.8**	0.8
%PT	7.8*	33.8**	0.5
SA	7.0	4.8*	1.0
EA	2.2	4.4*	0.6
PA	1.0	1.2	0.0
WS	53.3**	259.0**	39.7**
Angles			
min. trunk	16.0*	44.0**	0.0
max. trunk	60.0**	17.6**	1.3
min. upperarm	3.0	10.4**	0.0
max. upperarm	7.8*	6.7**	1.7
min. elbow	0.2	0.0	2.0
Angular velocities			
max. trunk	12.3*	7.1**	0.7
max. upperarm	0.0	60.1**	3.3
max. elbow	0.9	22.8**	3.3
ang. vel. hand-on-rim	5.4	149.0**	0.6
ang. acc. hand-on-rim	0.1	48.2**	1.5

* $p<0.05$. ** $p<0.01$.

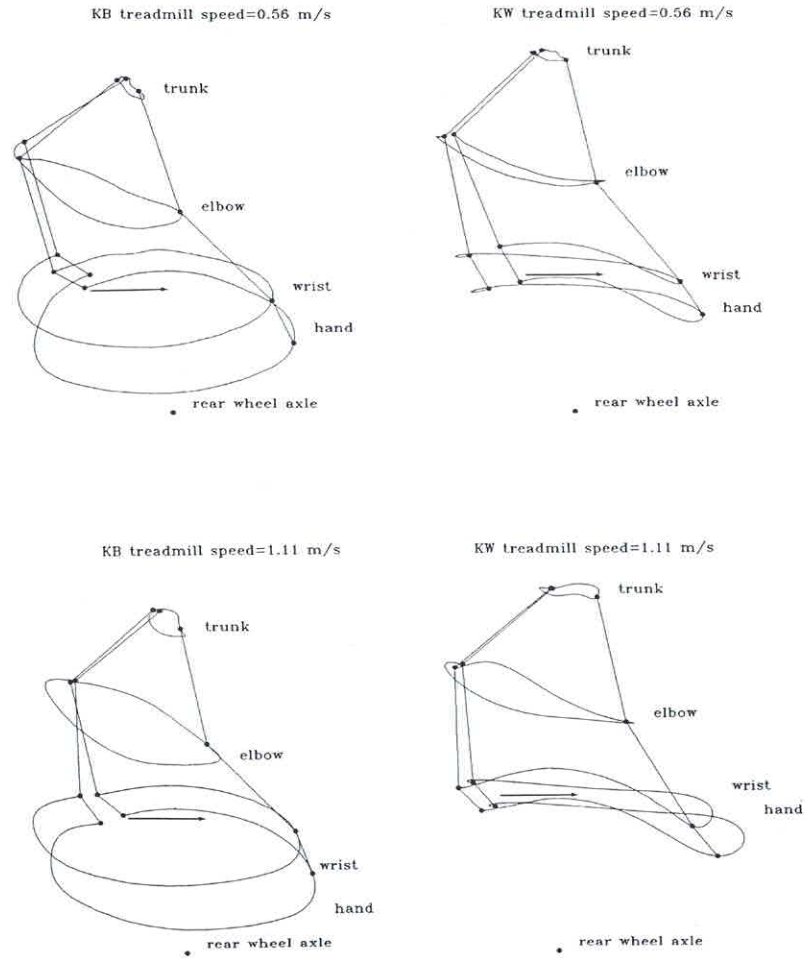

Fig. 1. Two-dimensional trajectories of the markers of acromion, elbows, wrists and hands for subjects K. B. and K. W. The positions of the markers at the start of the push, the end of the push and the start of the following push phase are drawn. The diagrams are presented for belt speeds of 0.56 and 1.11 meters/second at a two degree slope.

al projection. These subjects were selected on basis of differences between their stroke patterns; 'circular' (KB) versus 'pumping' (KW). The results show that within one condition the cycles are not completely identical: the position of the upper limb at the beginning and end of the cycle is not always exactly the same. However, the variation of style within these subjects is small. Despite the differing propulsion styles of both subjects, general features are also apparent: the push angle PA and segment angles differ only marginally.

Linear displacement and velocity of the wheelchair while driving on the treadmill was calculated for all conditions. Fig. 2. shows a typical example for the horizontal velocity of the rear wheel axle and the trunk at 1.11 m/s and a 2 degrees slope relative to a reference marker on the treadmill frame. The recovery phase is considerbly longer than the push phase. Moreover, during the first three quarters of the push phase the horizontal velocity of the axle is lower than the belt velocity. The velocity of the axle is higher than the belt speed during the last part of the push phase and during most of the recovery phase. The position of the center of gravity of the wheelchair-user combination with respect to the belt remained fairly constant as the trunk is moved in the opposite direction as the axle during the second half of the push phase and most of the recovery phase

ANOVA performed on timing- and stroke parameters did indeed not show any dependence on speed for PA. A small change with speed was found for SA and EA (Tables II and III). PT decreased strongly with speed while RT also decreased but not as strong as PT. Since CT is the sum of both, the decrease in CT is mainly caused by PT. This results in a decrease in PT expressed as a percentage of CT with increasing belt

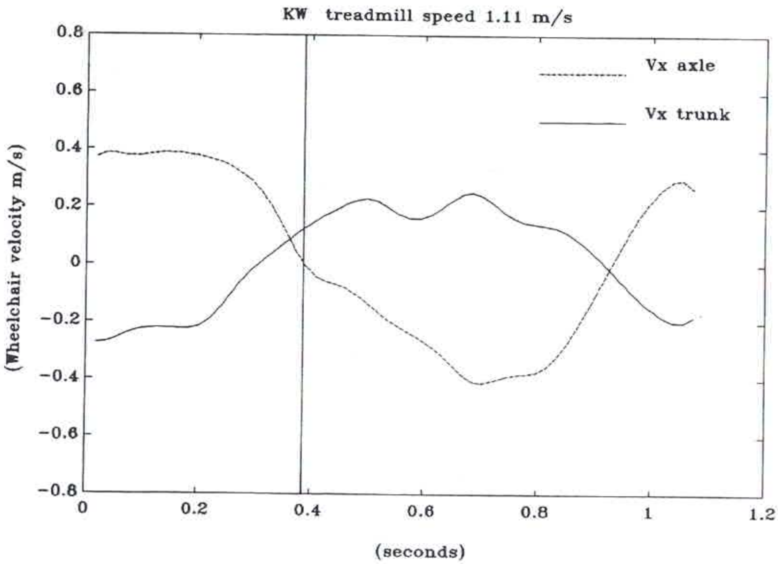

Fig. 2. Horizontal velocity of the wheelchair axle relative to a fixed marker on the treadmill frame. Plotted are the horizontal velocities of the rear axle (dotted line) and the trunk marker (solid line). The vertical line indicates the end of the push phase.

speed. Work per stroke (WS) increased significantly with belt speed. During the push phase, increasing peak angular accelerations of the hands with speed were found.

A comparison between results obtained at the slopes used showed analogous speed dependent changes for both sessions. Significant differences were found for both CT and RT: while with respect to the different slopes CT and RT were shorter for steeper slopes at the same speed, PT did not change with slope.

The segment marker projections shown in Fig. 1 indicated speed dependent changes of segment excursions for both subjects. Fig. 3 shows some of the angles and angular velocities during the push phase for the same subjects as in Fig. 1. Despite inter-individual variations in angles and angular velocities, ANOVA ($N=5$) indicated significant speed related differences for the minimum and maximum values during the push phase. The minimum and maximum angles of the trunk and the upper arms changed significantly with belt speed, but the elbow angles did not (Tables 2 and 3). The trunk angles increased with speed, the upper arm angles decreased: The trunk is flexed more forward and the arms are brought more backwards. The sectors of motion of trunk and arms

Table III. *Mean results for parameters push time (PT), recovery time (RT), cycle time (CT), start angle (SA), end angle (EA) and push angle (PA)*

Speed (km/h)	Parameter	2° slope (s)	3° slope (s)	Parameter	2° slope (rad)	3° slope (rad)
2	Push time	0.67	0.63	Start angle	1.1	1.14
3		0.49	0.5		1.05	1.13
4		0.36	0.34		1.1	1.22
5		0.30	0.29		1.13	1.25
2	Recovery time	0.78	0.65	End angle	2.41	2.43
3		0.71	0.6		2.44	2.53
4		0.65	0.55		2.49	2.58
5		0.62	0.51		2.51	2.6
2	Cycle time	1.45	1.28	Push angle	1.31	1.29
3		1.20	1.1		1.39	1.4
4		1.01	0.89		1.39	1.36
5		0.92	0.81		1.38	1.35

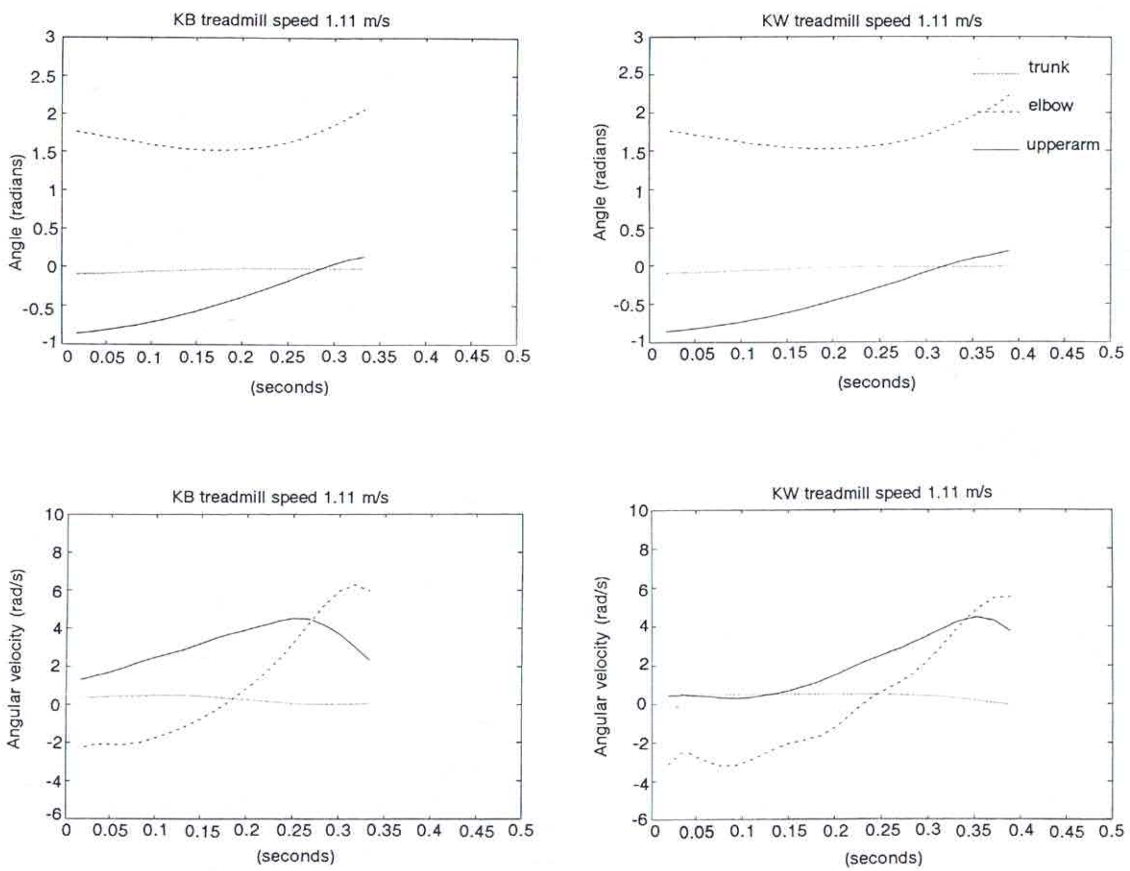

Fig. 3. Angles and angular velocities of upper arms and trunk relative to the vertical and three-dimensional elbow angle and angular velocity. Time series describe the push phase for subjects K. B. and K. W. at a belt speed of 1.11 m/s and a two degree slope.

did not appear to change but were shifted toward flexion (trunk) and extension (upper arms).

All maximum angular velocities increased with belt speed. The increase of angular velocity with slope was only significant for the trunk (Fig. 4).

DISCUSSION

The subjects of this study were all well trained wheelchair sportsmen. However, the group was not homogeneous with respect to impairment or body mass (Table I). Differences in propulsion styles between subjects were considerable. Movement patterns were classified as predominantly 'circular' in subject K. B. to a 'pumping' technique in subject K. W. (Fig. 1). These clear inter-subject but small within-subject variations agree with the findings of Sanderson & Sommer (4). The movement pattern of subject K. W. can be compared with those described by Brauer (1) for wheelchair propulsion on an ergometer. Inspection of speed related changes in style was not indicative of incomparability of those changes in our subjects. Sanderson & Sommer suggest that the circular pushing style is superior to the other styles some of their subjects adopted. In the present study subject K. B., using a circular propulsion technique showed a significantly higher mechanical efficiency than subject K. W. (mean values 10.7% versus 9.5%, $p<0.05$). Conclusions on basis of this study with respect to a causal relationship between style and ME would be premature since other differences were found as well: mean power output for subject KB was 37.1 watts versus 44.4 watts for subject K. W. ($p<0.05$).

Differences of belt speed were found to be accompanied by changes of cycle time and push time but not by differences in push angle. The consequence of a

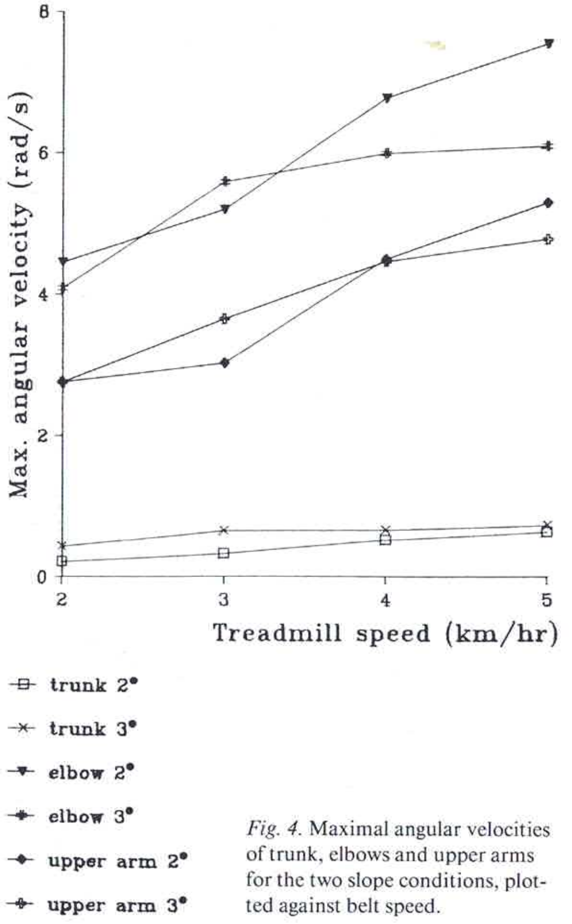

Fig. 4. Maximal angular velocities of trunk, elbows and upper arms for the two slope conditions, plotted against belt speed.

shortened CT during increasing power output at higher belt velocities is a strongly increased work per stroke. Since belt speeds employed in this study did only increase up to 1.39 m/s, it can be argued that the maximum speed in this experiment may have been too low for changes in push angle to occur. The present results regarding cycle time however do agree with those of an earlier study using a racing wheelchair at speeds up to 4.17 m/s (9).

The decrease of cycle time with speed also agrees with results of previous studies (1, 4, 6, 7). However, exact comparisons are difficult to make since CT is also dependent on other factors affecting work load. A higher work load at the same speed, for instance as a result of a higher body mass, leads to a higher stroke frequency (1, 5). Higher work loads due to a steeper slope at equal speeds are also related to a lower CT.

Increasing belt speed not only influenced work load, but also led to considerably higher tangential velocities of the rims. This higher speed of travel must have led to a shorter push time. Since recovery time decreased considerably less than push time, the percentage PT of the total cycle time also diminished. The decrease of PT could theoretically partly have been compensated by an extended push angle, but in the present study PA was not found to change. Above finding was supported by the results for a racing wheelchair experiment in which the push angle was found not to change significantly with speed (9). The decrease of PT must thus be the inevitable result of higher speeds. Because at a higher speed also more power must be produced against a higher tangential velocity, the conditions are triple unfavorable: due to the higher work load more power has to be produced in less time against a higher hand rim velocity. It would have been possible for subjects to improve the conditions by either extending their push angle and thus counteracting the decrease in push time, or by increasing their stroke frequency while lowering the recovery time and thus enhancing the percentage push time. However, this adaptation did not occur as it does in adaptation to steeper slopes, in which case the extra power output needed for negotiating the steeper slope seems mainly to be related to a shorter RT. Results of a study by Woude et al. (10) of the effect of different cycle frequencies on efficiency showed that when forced into a different frequency, subjects tended to change RT far stronger than PT.

The linear displacement of the wheel axle (Fig. 2) during the recovery phase is likely to be the result of flexion and extension of the trunk, such that the combined center of gravity of the chair and user is in fact slowed down according to the losses in energy to rolling resistance, internal resistance and gravity effects related to treadmill slope. The decrease in velocity over the complete recovery phase indeed coincides with the theoretical decrease calculated on basis of average energy losses. In reality a change of position of the center of gravity also has consequences for losses due to rolling resistance. A rearward displacement will lower these losses since more mass will be supported by the rear wheels which have a lower rolling resistance than the front casters.

Trunk flexion as well as the flexion of the upper arms are shown to be strongly related to changes of belt speed. These changes of segment sectors of movement cannot be caused by an increasing push angle since PA does not change. It is suggested that these changes are the result of the higher power output demands of the task itself and thus of propulsive impact needed to produce more power at a higher angular velocity of the rims and a constant push angle. The fact that minimum elbow angles do not change with speed is probably due to the small amount of change in trunk flexion, not sufficient to cause a significant change in elbow angle.

Since at a treadmill slope of 3 degrees where power output losses are higher than at a 2 degree slope, a stronger retroflexion of the upper arms followed by a larger trunk flexion was found, it is suggested that most power will be delivered around the shoulder joint. Further studies on this will be performed with a wheelchair ergometer facilitating the measurement of forces applied on the rims and on the seat.

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