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WHEELCHAIR PROPULSION: FUNCTIONAL ABILITY DEPENDENT FACTORS IN WHEELCHAIR BASKETBALL PLAYERS

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ABSTRACT. The aim of this study was to examine the user-related parameters, 1) force generation 2) maximal aerobic power and 3) propulsion technique, in respect to functional ability level [ISMWSF] wheelchair basketball classification: groups 1, 2 and 3) of 40 elite wheelchair basketball players. Isometric (position on the handrims = −30, 0°, +30° and +60°) and dynamic force application (velocities = 0.56, 0.83 and 1.11 m.s^{-1}) on the handrims (test 1) was measured by means of a computerised wheelchair simulator, with the subjects sitting in a standardised position. Each subject performed a maximal exercise test (test 2) on a motor driven treadmill at 1.67 m.s^{-1} and four subsequent submaximal tests (test 3) at two exercise levels (60 and 80% of individual VO_2) and two velocities (1.11 and 2.22 m.s^{-1}) with constant power output. In tests 2 and 3, cardiorespiratory and kinematic data were recorded simultaneously. Although no significant differences between functional ability groups were found in relation to force application on the handrims, three different force application strategies were observed (test 1). Maximal aerobic capacity and power output (test 2) differed significantly ($p < 0.05$) between groups 1 and 2 and between groups 1 and 3. No differences in mechanical efficiency were observed between the three functional ability groups, irrespective of changes in wheelchair velocity and external load (test 3). Propulsion technique was not proven to be functional ability dependent although remarkable differences in movement pattern were observed, especially during the recovery phase.

Key words: muscle strength, aerobic power, wheelchair propulsion technique, functional classification, mechanical efficiency.

The ability of the wheelchair user is, among other factors, influenced by cardiorespiratory fitness, muscular strength and upper limb coordination. In practice, however, it is difficult to disentangle such factors from the influence of differences in habitual activity (training, psychological and social factors, etc.) and limitations of performance imposed by some primary disease process (15). The analysis of such factors also requires an integrated biomechanical and physiological approach. This approach has not often been adopted in manual wheelchair propulsion (5, 14), although interest is increasing (3, 6, 19, 20, 21, 25). The limited number of multidisciplinary studies contributing to the database on manual wheelchair propulsion often contains small samples (19, 20), groups with mixed disability levels (23), or subjects who are inexperienced in wheelchair propulsion (25).

In order to clarify the influence of disability level upon wheelchair propulsion performance, the subjects selected for this study were three populations of wheelchair users (International Stoke Mandeville Wheelchair Sports Federation [ISMWSF] classification [17]), differing in disability level; between group differences with respect to age, training intensity, sports discipline and duration of wheelchair confinement, were avoided. The user-related parameters, approached from a biomechanical and physiological viewpoint, were 1) force application on the handrims at different velocities on a wheelchair ergometer, 2) maximal aerobic power during wheelchair propulsion on a treadmill and 3) propulsion technique, i.e. the movement pattern driving the wheelchair at different velocities on a treadmill at constant submaximal power output. The efficiency of the wheelchair or the wheelchair-user-interface was not the subject of this study.

MATERIAL AND METHODS

Subjects

Forty highly trained male wheelchair athletes participated in the three tests. Subjects were classified in groups 1 to 3, according to the ISMWSF classification system for wheel-

Table I. *General characteristics of the subjects (mean & sd) categorized according to the International Stoke Mandeville Wheelchair Sports Federation (ISMWSF) classification system*

Pathology	ISMWSF	Age (years) mean & sd	Weight (kg) mean & sd	Number of years disabled mean & sd	Wheelchair confinement (years) mean & sd	Training (years) mean & sd	Training (hours/week) mean & sd
Group 1 ($n=13$)							
Spinal cord injury (lesion level T3–T12) ($n=12$)				11.1 (5.9)			
Polio ($n=1$)	I–I.5	29.6 (4.8)	65.5 (12.6)	33 (--)	12.92 (7.46)	7.07 (6.90)	4.5 (1.65)
Group 2 ($n=14$)							
Spinal cord injury (lesion level T9–T12) ($n=8$)				8.6 (4.8)			
Polio ($n=5$)				29.6 (7.1)			
Spina Bifida ($n=1$)	II–II.5	32.9 (8.4)	70.7 (12.4)	34 (--)	10.5 (7.16)	9.0 (6.51)	6.42 (3.41)
Group 3 ($n=13$)							
Spinal cord injury (lesion level L1–L5) ($n=2$)				6 (2.8)			
Polio ($n=7$)			30.4 (7.4)				
Spina Bifida ($n=1$)				18 (7.6)			
Amputation ($n=3$)	III–IV	32.8 (7.2)	67.9 (12.2)	30 (–)	no use in ADL	13.07 (8.6)	5.46 (1.71)

chair basketball (17). The general characteristics of the subjects including age, functional classification, disability level, as well as the duration of wheelchair confinement are presented in Table I. Consent was given by all subjects, in full knowledge of what the tests entailed.

Test 1: Force application on the handrims

Test protocol: Handrims (diameter: 53 cm, cross-sectional diameter: 22 mm) were mounted on an ergodyn device, i.e. a computerised device to evaluate joint functions. For an extensive description, see Spaepen et al. (16). A torque sensor, mounted at the produced part of the drive-axle, registered the torque generated at both handrims. The position of the handrims was measured by means of an electrogoniometer. Seat height of the chair was standardised at 90° elbow-angle and forward and backward positions were standardised with the acromion above the rear wheel axle. Wheel cambers were set at neutral position (0°).

For the isometric tests, subjects were asked to apply maximal propulsive force on the handrims at different hand contact positions (−30°, 0° = top of the handrim, +30° and +60°). For the dynamic tests, a maximal propulsive force at different handrim velocities (0.56, 0.83 and 1.11 $m.s^{-1}$) was requested. All tests were carried out twice in random sequence, with only the highest values being used for further analysis.

Kinematic data: Subjects' movement patterns were filmed by means of two video HI8 cameras. Markers were placed on the anatomical landmarks: acromion, epicondylus lateralis humeri, processus styloideus ulnaris and on the rear wheel axle. After automatic digitisation of the interfaced images (Kinemetrix Fs = 50 Hz), three-dimensional reconstruction was performed using the direct linear transformation method. Reconstruction accuracy was within 1.5 mm/marker. When marker positions were hidden due to segmental rotation, their positions were estimated by means of linear interpolation. No further processing of the three-dimensional data was necessary due to the high accuracy of the registration method. The following parameters were determined: start angle (SA), end angle (EA), push angle (PA) and the movement pattern of the arms and trunk (Fig. 1). Trunk inclination (TI) was calculated as the angle between a vertical line and the line between the acromion marker and the top of the handrim. The range of motion of the trunk (TR) was defined as the maximal range of the trunk angle within one propulsion cycle.

Test 2: Maximal cardiorespiratory capacity

Test protocol: Tests were performed in a lightweight basketball wheelchair (Quicki GPV, weight: 11.6 kg) on a motor driven treadmill (MDT). Seat height, forward and backward positions, and wheelchair configurations were standardised as in test 1. Rolling resistance of the wheelchair-user system was determined twice with the use of a drag test at 1.67 $m.s^{-1}$, the user being in an upright and inclined position. The subjects were asked to perform a discontinuous protocol on the MDT at 1.67 $m.s^{-1}$ and a 0° slope with the initial workload at the individual's rolling resistance. Loads were subsequently increased by a multiple of the individual's rolling resistance by means of a pulley system (Fig. 2) (21). Each stage lasted four minutes followed by a 2 minute active recuperation period at 1.67 $m.s^{-1}$ without supplementary load. Loads were increased until exhaustion. During the fourth minute of each stage of the maximal test, metabolic data were collected.

Metabolic and kinematic data: The composition of the expired air was continuously monitored with the use of an OXYMAT (Siemens) oxygen analyser. In the fourth minute of each bout of exercise, the following cardiorespiratory parameters were registered: minute ventilation (VE, $l.min^{-1}$, BTPS), oxygen uptake (VO_2, $l.min^{-1}$, STPD), carbon dioxide output (VCO_2, $l.min^{-1}$, STPD) and respiratory exchange ratio (RER). Heart rate was monitored electrocardiographically. Kinematic data were collected as in test 1. To identify hand contact (HC) and hand release (HR), the acceleration-curve of the wheelchair-user system was analysed. Visual identification of HC and HR was used as a control. The following time parameters were added to the kinematic data: Push Time (PT = time during which the hand is in contact with the handrim), Recovery Time (RT = time during which the hand, after releasing the handrim, is brought back to the starting position [start angle]), Cycle Time (CT = PT + RT) and Cycle Frequency (CF = number of propulsion cycles that the user performs per unit of time).

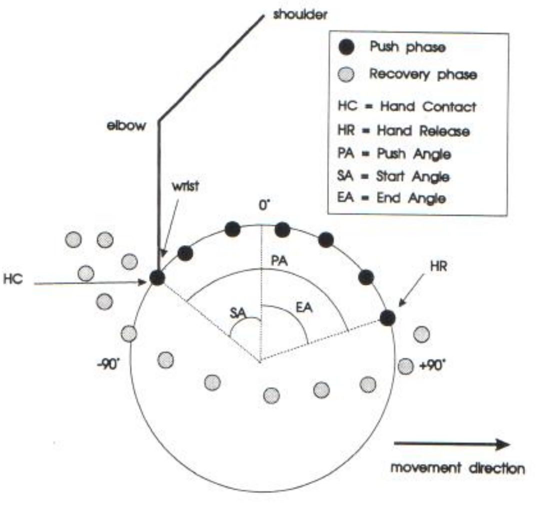

Fig. 1. Kinematic parameters used to analyse one propulsion cycle.

Test 3: Propulsion technique

Test protocol: Subsequent to the maximal test and after an interval of at least 1 hour, four submaximal tests, each of 6 minutes' duration, were performed in a random sequence at two velocities (1.11 & 2.22 $m.s^{-1}$), and two levels of power output (60 & 80% of the peak VO_2 obtained during the maximal test). Metabolic and kinematic data were recorded simultaneously during 8.2 seconds as described in test 2.

Statistics

Unless otherwise mentioned, a one-way Analysis of Variance (ANOVA) was used for the three tests, with 'functional ability' as the main factor, and the significance level as $p < 0.05$.

RESULTS

Test 1: Force application on the handrims

No intergroup differences in peak-isometric torque registered at the rear wheel axle could be demonstrated, except for the position $+60°$ (Fig. 3). The better performance of group 3 in this position is logical since an increase in force application at the front of the handrims implies the functional use of abdominal muscles. No significant intergroup differences in dynamic force application are apparent when calculating peak-torque, power output, total work and average work, although the results of group 3 were always slightly higher for each parameter (Table II). Further analysis, however, reveals totally different force application strategies between the three functional ability groups. Strategy 1 (Fig. 4) could be described as a single phase force application with a progressive increase in the torque curve until peak-torque is reached, after which a progressive decrease is demonstrated. Peak-torque is commonly localised in the middle of the push angle. This strategy is frequently used in combination with a small push angle and trunk inclination, commonly employed by subjects without control of abdominal and lower back muscles. Strategy 2 could be described as a two-phase force application, with peak-torque reached at the end of the push angle. Strategy 3 is also a two-phase force application, but peak-torque is localised in the first phase. Both strategies two and three, are accompanied by large push angles and trunk inclinations, which were demonstrated in functional ability groups 2 and 3. The significant difference between groups in push angle and trunk inclination are presented in Fig. 5 (group 1 and 2 [$p < 0.05$], group 2 and 3 [$p < 0.01$]).

Test 2: Maximal cardiorespiratory capacity

Between group differences (functional ability group 1 in comparison with groups 2 and 3) were found ($p < 0.05$) for peak oxygen consumption (peak-VO_2), peak pulmonary ventilation (peak-VE) and peak power output (peak-PO). Table III reveals high standard deviations for all parameters, elucidating high interindividual differences within the functional ability groups.

Gross mechanical efficiency (GME) was calculated as the ratio of external power output over energy expenditure (2), using metabolic data recorded during

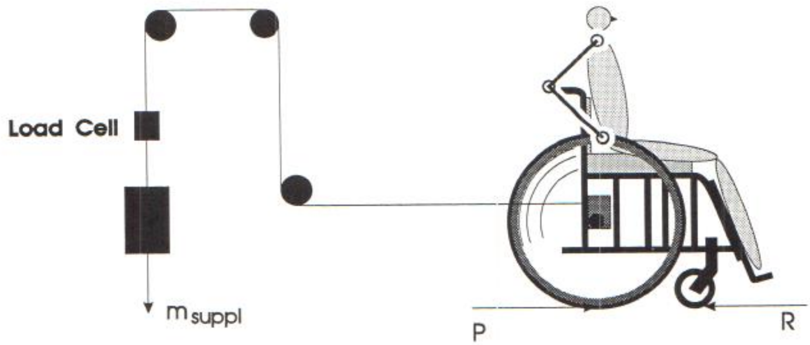

Fig. 2. Pulley system to increase workload (P: Propulsion force; R: Rolling resistance; m_{suppl}: supplementary weight).

the last minute of each exercise stage, excluding the last stage when exhaustion appeared. GME increased parabolically with power output, attaining maximal values of 13%. No significant intergroup differences were demonstrated when GME was calculated at an individual's submaximal level of effort (60 or 80% of individual peak-VO_2).

Test 3: Propulsion technique

GME values did not exceed 11.5% (=maximal value registered). No differences in GME were found between functional ability groups (Table IV). Nor were mean values dependent on exercise level, although all mean values were found to be slightly higher at the 80% exercise level. Even though all tests were carried out with constant power output, energy consumption increased remarkably with increased velocity from 1.67 to 2.22 $m.s^{-1}$ resulting in a significant decrease in efficiency ($p<0.01$). Some subjects also achieved their maximal oxygen consumption when driving the wheelchair at 2.22 $m.s^{-1}$ and 80% exercise level, because of the high velocity. As a steady state is out of the question at maximal exercise level, mean mechanical efficiency values for this test (velocity: 2.22 $m.s^{-1}$; level: 80%) are probably overestimated.

In increasing velocity from 1.11 to 2.22 $m.s^{-1}$, cycle time decreased progressively ($p<0.01$) due to a decrease in push time ($p<0.01$), while recovery time was kept almost constant (Fig. 6). A decrease in cycle time implies an increase in cycle frequency. These changes in time parameters were demonstrated for all three groups at different exercise levels (Table V). All subjects shifted start and end angles to the front of the handrim ($p<0.01$) with increasing velocity, without

Table II. *Peak-torque (Nm), power output (W), total work (J) and average power (W) for the three functional ability groups at different handrim velocities*

	Peak torque			Power output		
Speed	0.56 m/s mean & sd	0.83 m/s mean & sd	1.11 m/s mean & sd	0.56 m/s mean & sd	0.83 m/s mean & sd	1.11 m/s mean & sd
Group 1	132.5 (20.5)	124.2 (19.1)	117.9 (17.2)	445.4 (90.9)	607.2 (125.6)	699.4 (159.8)
Group 2	121.6 (14.4)	116.8 (17.9)	117.0 (17.3)	409.3 (71.9)	570.0 (116.9)	693.2 (162.5)
Group 3	140.7 (20.8)	126.1 (29.1)	124.3 (31.2)	471.7 (102.6)	614.4 (188.8)	736.7 (242.3)
	Total work			Average power		
Speed	0.56 m/s mean & sd	0.83 m/s mean & sd	1.11 m/s mean & sd	0.56 m/s mean & sd	0.83 m/s mean & sd	1.11 m/s mean & sd
Group 1	40.7 (12.5)	25.2 (6.9)	19.7 (5.7)	80.3 (15.8)	72.7 (14.1)	70.9 (14.6)
Group 2	43.2 (11.5)	26.6 (7.3)	19.5 (6.6)	84.5 (23.0)	75.2 (21.1)	69.5 (23.4)
Group 3	51.7 (13.6)	30.7 (13.0)	23.1 (10.9)	94.8 (21.1)	79.7 (30.3)	73.0 (31.2)

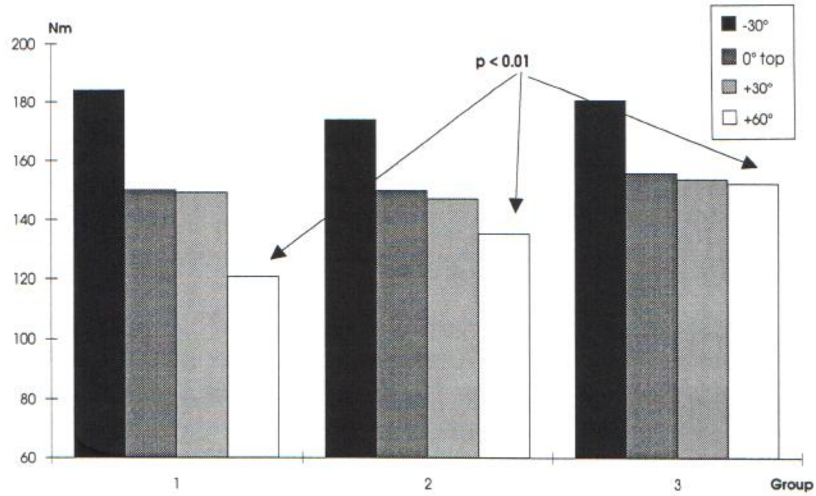

Fig. 3. Peak-isometric torque (Nm) on both handrims at different hand contact positions for the three functional ability groups.

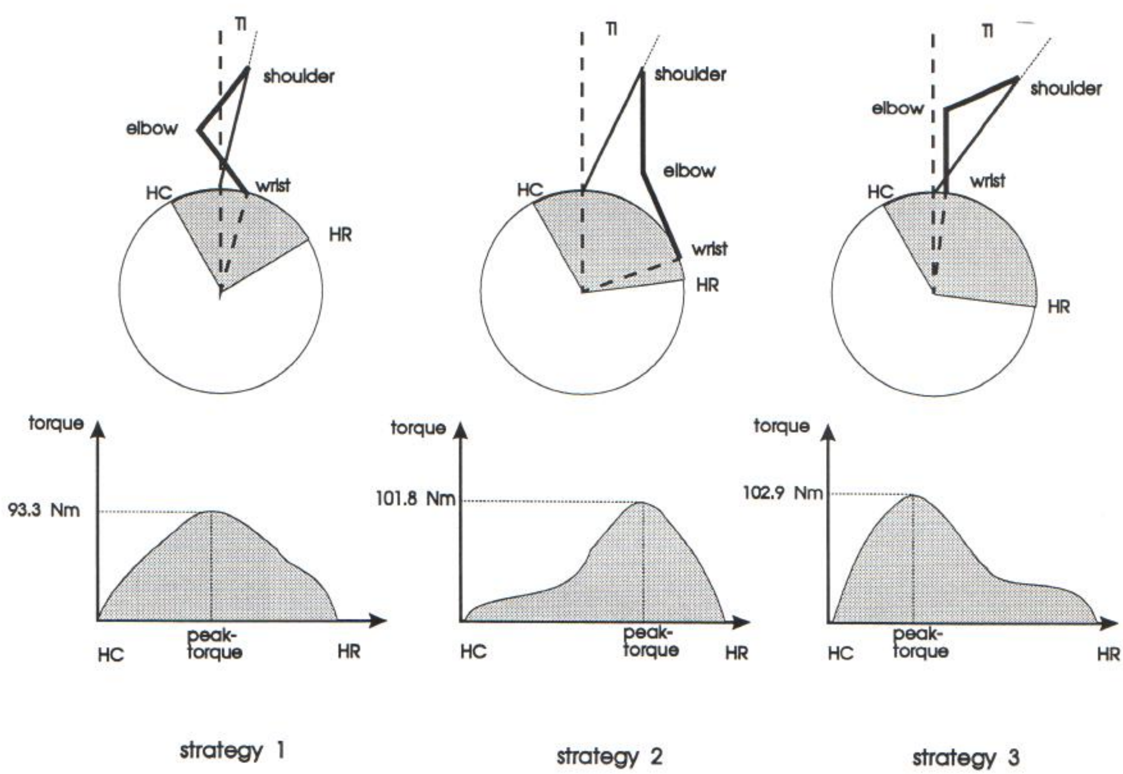

Fig. 4. Example of the different force application strategies used within the tested population. Above, the upper limb position at peak-torque is portrayed from a lateral view (the shadowed area equals the push angle). Below, the correspondent torque-curves are drawn (HC = Hand Contact, HR = Hand Release, TI = Trunk Inclination).

Table III. *Physiological parameters (VO_2: oxygen consumption; PO: Power Output; VE: Pulmonary ventilation; $RQ = VCO_2/VO_2$) for the three functional ability groups*

	VO_2 (l/min) mean & sd	VO_2 (ml/kg/min) mean & sd	PO (Watt) mean & sd	VE (l/min) mean & sd	RQ mean & sd
Group 1	1.86 (0.49)	29.70 (8.62)	67.84 (14.59)	67.29 (22.49)	1.02 (0.07)
Group 2	2.44 (0.44)	36.28 (9.31)	91.25 (26.56)	80.64 (20.01)	1.04 (0.07)
Group 3	2.59 (0.32)	37.93 (5.18)	99.71 (18.79)	96.88 (13.94)	1.05 (0.07)

changing the push angle significantly (Fig. 7). Only subjects with a good sitting balance (group 3, 80% level) were able to increase their trunk inclination.

Comparing the three functional ability groups, no statistical difference could be found in time parameters. However, subjects with less trunk control (group 1) applied forces more at the top of the handrim in comparison with amputees and subjects with lumbar spinal cord lesions (group 3).

In Fig. 8, a typical movement pattern of a subject is portrayed from a lateral view for each test condition. No important differences were found during the push phase as the arm is guided by the handrim and moving in a closed chain. Nor do any differences in the

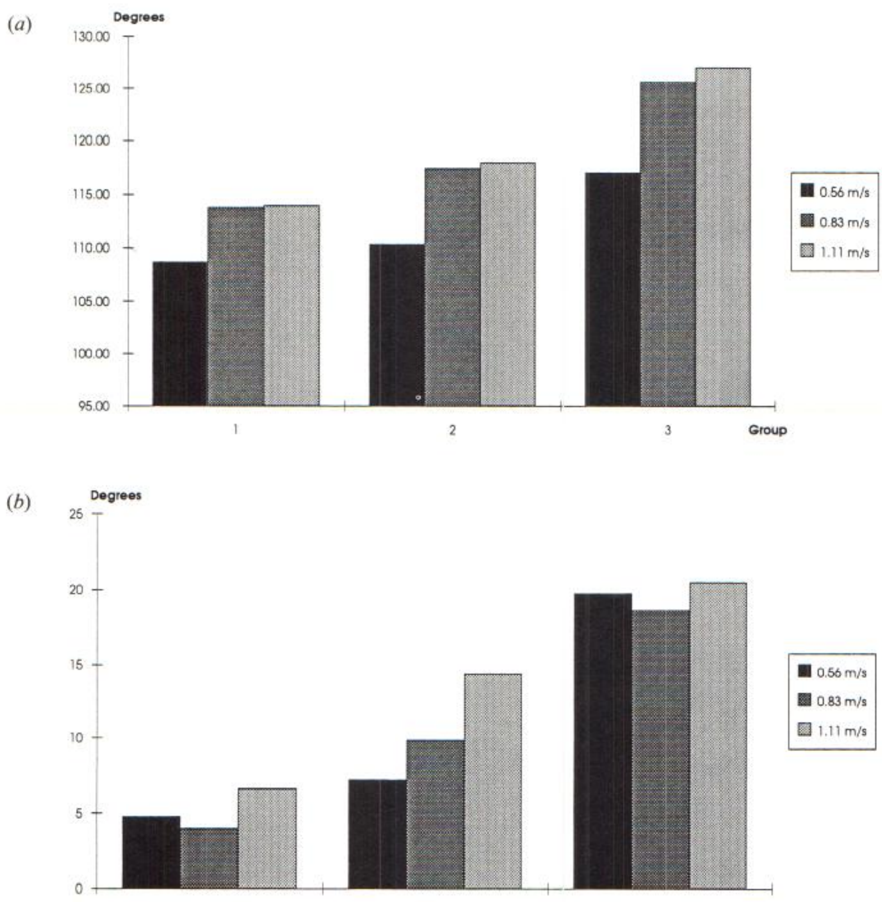

Fig. 5.(a) Push angles for the three functional ability groups. Mean angles are presented for the maximal dynamic force application test at three handrim velocities. *(b)* Trunk inclination for the three functional ability groups. Mean angles are presented for the maximal dynamic force application test at three handrim velocities.

Table IV. *Gross mechanical efficiency at different velocities, 60 and 80% exercise level and constant power output for each level*

Means and standard deviations are presented for the three functional ability groups.

Level	60%			80%		
Speed	1.11 m/s	1.67 m/s	2.22 m/s	1.11 m/s	1.67 m/s	2.22 m/s
Group 1	9.67 (1.58)	9.53 (2.05)	6.81 (0.79)	10.39 (1.50)	9.81 (1.63)	7.79 (1.07)
Group 2	10.33 (2.13)	9.57 (2.41)	7.35 (2.09)	11.04 (1.36)	10.28 (2.32)	8.12 (1.98)
Group 3	10.5 (1.42)	9.76 (1.34)	7.62 (1.47)	10.96 (1.55)	10.54 (1.48)	8.33 (1.21)

movement patterns become immediately apparent when velocity is increased. Remarkable interindividual differences in propulsion technique appeared when focussing on the total movement pattern of the upper limbs (i.e. push and recovery phase) during wheelchair propulsion. Propulsion technique, however, could not be proven to be group-dependent. Subjects with remarkable differences in functional ability, often demonstrate a comparable wheelchair propulsion style (Fig. 9).

Further analysis of the movement pattern revealed maximal abduction of the shoulder at hand contact (at the beginning of the push phase) in all subjects. Subdivision of the push angle into two consecutive phases, each representing 50% of the total push angle, revealed only slight movements at the different joints in the first phase (mean values for all subjects are: elbow flexion-extension = 14°, shoulder adduction = 5°, shoulder flexion = 13°, trunk inclination = 4°). Intergroup differences were only significant for trunk inclination (Table VI). From a top view, external rotation at the shoulder (the ulna rotating underneath the humerus) dominated the movement pattern of the subjects in the first 50% of the push angle (Fig. 10). In the second 50% of the push angle (Table VI and Fig. 10), the movement pattern was characterised by a combined shoulder flexion (mean = 36°), shoulder adduction (mean = 18°) and elbow extension (mean = 34°). Amplitude of movement was group-dependent for each joint, ability group 3 using the largest movement pattern. Trunk inclination (mean = 2°) decreased in comparison with the first phase but no differences could be shown between groups.

DISCUSSION

Upper-body muscle strength of wheelchair users has often been measured analytically (8, 9, 18). Techniques previously used to assess muscular force included hand dynamometry (27, 28), static upper-arm cable tensiometry (4, 12) and isokinetic dynamometry (8, 12,

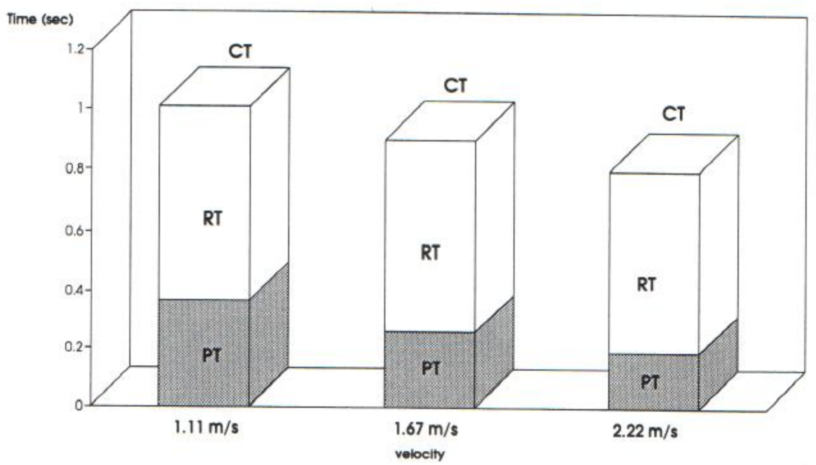

Fig. 6. Cycle Time (CT) expressed as the sum of Push Time (PT) and Recovery Time (RT). Mean values are presented for group 3 at 80% exercise level.

Table V. *Mean values (seconds) and standard deviations for timing parameters*

A three-way Analysis of Variance (ANOVA) was executed with 'functional ability', 'exercise level' and 'handrim velocity' as the main factors (ns = not significant, * = $p < 0.05$, ** = $p < 0.01$, CT = Cycle Time, CF = Cycle Frequency, PT = Push Time, RT = Recovery Time).

Speed m/s	60% 1.11		60% 1.67		60% 2.22		80% 1.11		80% 1.67		80% 2.22		Speed	Level	Group
	mean	sd	mean	sd	mean	sd	mean	sd	mean	sd	mean	sd			
CT															
Group 1	1.02	*0.21*	0.92	*0.11*	0.79	*0.15*	1.01	*0.29*	0.87	*0.13*	0.74	*0.13*			
Group 2	1.02	*0.14*	0.87	*0.08*	0.81	*0.12*	0.97	*0.14*	0.83	*0.08*	0.75	*0.08*	**	*	ns
Group 3	1.02	*0.31*	0.98	*0.27*	0.88	*0.17*	0.98	*0.24*	0.94	*0.24*	0.85	*0.15*			
CF															
Group 1	60.88	*12.2*	65.76	*10.4*	76.74	*14.7*	66.92	*18.4*	71.61	*10.3*	82.89	*14.8*			
Group 2	60.25	*10.4*	68.12	*6.12*	70.93	*13.2*	62.71	*9.24*	70.29	*6.73*	76.81	*9.88*	**	*	ns
Group 3	60.33	*13.9*	63.98	*12.8*	67.55	*11.9*	64.15	*13.3*	66.59	*12.69*	73.57	*11.1*			
PT															
Group 1	0.37	*0.06*	0.26	*0.03*	0.19	*0.02*	0.38	*0.07*	0.28	*0.03*	0.19	*0.03*			
Group 2	0.39	*0.06*	0.29	*0.03*	0.23	*0.03*	0.39	*0.06*	0.28	*0.02*	0.24	*0.02*	**	ns	ns
Group 3	0.36	*0.09*	0.27	*0.03*	0.21	*0.03*	0.37	*0.09*	0.27	*0.04*	0.22	*0.04*			
RT															
Group 1	0.64	*0.18*	0.64	*0.09*	0.61	*0.13*	0.61	*0.26*	0.58	*0.12*	0.54	*0.09*			
Group 2	0.61	*0.13*	0.57	*0.08*	0.58	*0.12*	0.58	*0.12*	0.54	*0.07*	0.51	*0.07*	ns	*	ns
Group 3	0.66	*0.23*	0.69	*0.23*	0.69	*0.15*	0.61	*0.17*	0.65	*0.19*	0.63	*0.13*			

28). Generally, the authors' conclusions point in the same direction: regardless of the technique used, few differences in upper limb strength can be traced to differing severity of disability. Minimal differences in strength were particularly observed over the ISMWSF disability Classes II through V. More recently, as a result of wheelchair-simulator development (13), it has been possible to measure forces applied on the handrims in a functional way, specific to wheelchair propulsion. Our results from test 1 revealed little impact of the level of functional ability on isometric and dynamic force application on the handrims. Differences in force application strategy between functional ability groups need to be interpreted with caution, when transfered to natural wheelchair propulsion. Wheelchair simulators are stationary devices, protecting the wheelchair user from turning over backwards when exerting maximal force, especially at

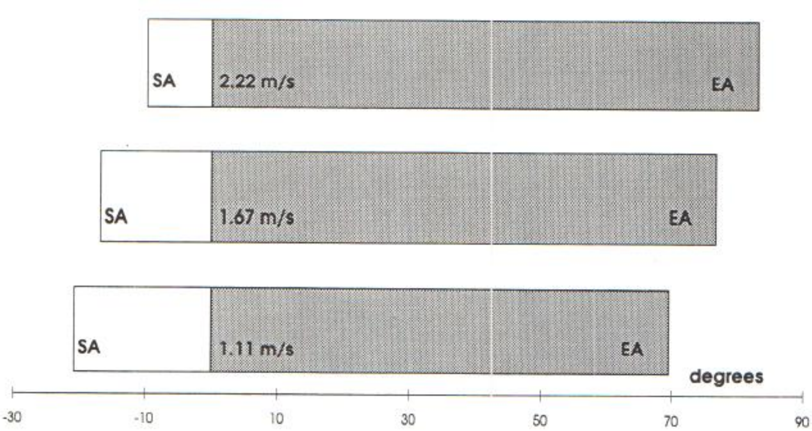

Fig. 7. Anterior shift of the Push Angle at three different velocities (SA = Start Angle, EA = End Angle). Results are presented for group 3 at 80% exercise level.

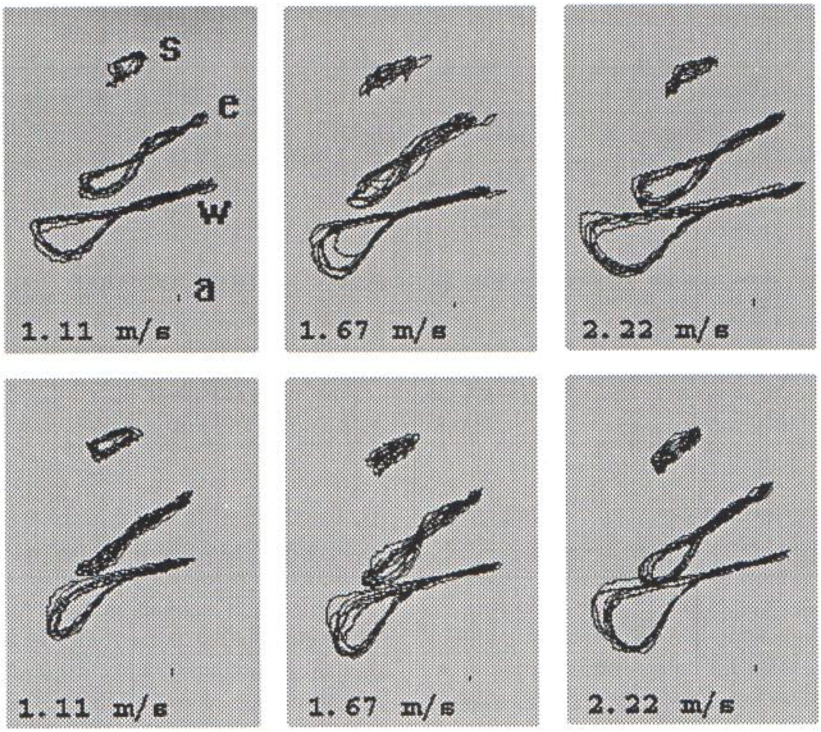

Fig. 8. The movement pattern of the left arm of one subject (group 3) is portrayed from a lateral view. (Above: 60% exercise level; below: 80% exercise level, s = shoulder, e = elbow, w = wrist, a = rear wheel axle).

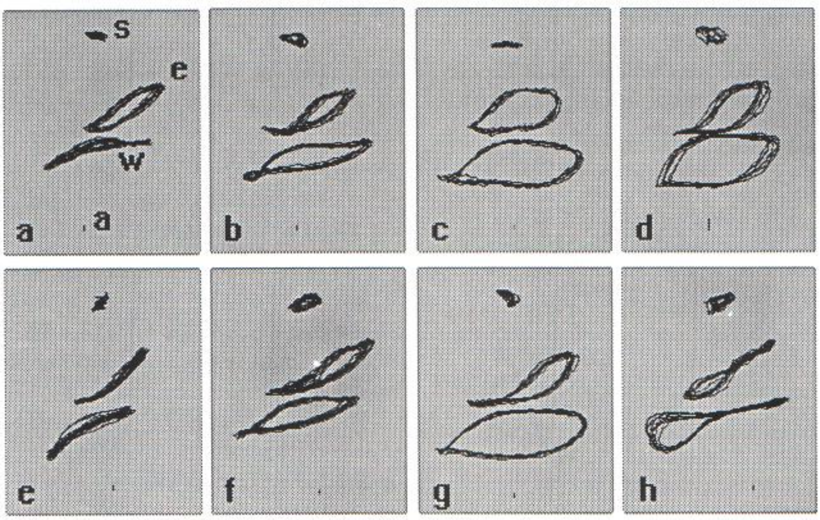

Fig. 9. The movement pattern of the left arm is portrayed from a lateral view of 8 subjects, driving the wheelchair at 2.22 $m.s^{-1}$ on the treadmill at 60% exercise level (a, b, c, and d = subjects from group 1; e, f, g, and h = subjects from group 3; s = shoulder, e = elbow, w = wrist, a = rear wheel axle).

Table VI. *Range of motion (degrees) of the elbow, shoulder and trunk at 1.11 m.s^{-1} and 60% exercise level*
Means and standard deviations are presented for the three functional ability groups.

	Elbow real angle		Shoulder frontal plane	
	First 50% push	Second 50% push	First 50% push	Second 50% push
Group 1	15.08 (6.00)	27.15 (7.66)	5.27 (2.02)	15.17 (4.62)
Group 2	13.27 (5.06)	32.40 (12.44)	3.81 (0.76)	19.10 (5.65)
Group 3	16.78 (5.59)	40.57 (12.84)	5.17 (1.97)	21.25 (6.14)
	Shoulder sagittal plane		Trunk sagittal plane	
	First 50% push	Second 50% push	First 50% push	Second 50% push
Group 1	13.08 (1.49)	31.69 (5.66)	2.65 (0.77)	1.82 (0.42)
Group 2	12.84 (3.76)	34.66 (9.08)	5.64 (3.36)	2.71 (1.49)
Group 3	15.67 (4.25)	43.45 (11.47)	5.02 (3.21)	1.95 (1.23)

the top of the handrims. Moreover, in a comparison between treadmill and wheelchair simulator tests, Veeger et al. (26) stated a significantly larger range of motion of the trunk when propelling the wheelchair on the treadmill, indicating differences in propulsion strategy. To generalise the different force application patterns observed in test 1, force generation needs to be examined over several propulsion cycles under different external conditions. As the device used in test 1 only allowed 180° rotation of the handrims, only one propulsion cycle could be imposed on the subject. Moreover, in test 1, subjects were asked to apply 'maximal' force on the handrims. Comparison with submaximal efforts such as wheelchair propulsion in daily living, is therefore inappropriate.

Differences in muscle strength and cardiorespiratory fitness between individuals differing in activity level have indicated the benefits from sports participation and other forms of fitness conditioning (10). It has been suggested that the functional ability level of the individual is less important than other factors in influencing post-morbid physical fitness except in those individuals with profound upper limb neuromuscular restriction (10). Coutts et al. (7) noted the existence of an empirical basis for grouping the ISMWSF classifications into three groups: tetraplegics, high-lesion paraplegics and low-lesion paraplegics. Examinations of results with respect to the ISMWSF classification as described by Huellemann et al. (11), Wicks et al. (27), and Veeger et al. (23) support this view. In the current study on elite wheelchair basketball players, no differences in maximal power output and maximal aerobic power between classes II to IV (ISMWSF wheelchair basketball classification) were observed. Whereas these results agree with the above findings, the question remains whether power output, aerobic capacity and muscular strength are determinant factors in respect to performances in wheelchair basketball. Based on a skill proficiency test on 79 male wheelchair basketball players, Brasile (1) also suggested a re-evaluation of the current NWBA (National Wheelchair Basketball Association) classification system, keeping Class I athletes as they currently are and combining the other participants into one class II.

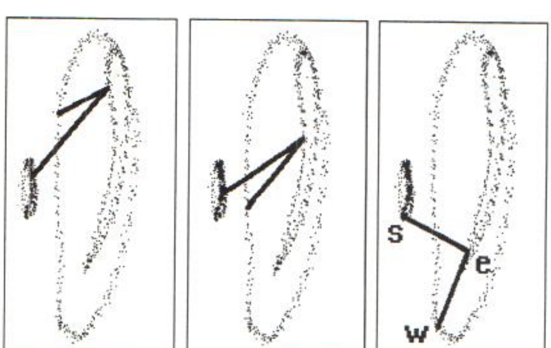

Fig. 10. Movement pattern for one subject, portrayed from a top view (*left*: at hand contact; *middle*: the hand is in the middle of the push angle; *right*: at hand release). The full movement pattern is shown as background (s = shoulder, e = elbow, w = wrist).

To our knowledge, no data have been published to date on the differences in wheelchair propulsion technique related to the subject's ability level. The power output and velocity dependency of mechanical efficiency in handrim wheelchair propulsion, as observed in the current study, had already been discussed in previous studies (21, 25). Unexpectedly, propulsion technique could not be proven to be functional ability dependent. Subjects with remarkable differences in functional ability often demonstrate comparable wheelchair propulsion styles (Fig. 9). The functional capacity of group 3 to incline the trunk as observed in the maximal force application tests (test 1) was only partly used when driving the wheelchair on the treadmill at the same velocity (1.11 $m.s^{-1}$). No differences were observed in mechanical efficiency between the three functional ability groups, irrespective of wheelchair velocity and external load. It seems that an intensive use of the wheelchair in daily living and sports leads to user optimisation of wheelchair propulsion technique, regardless of the individually chosen propulsion style.

The movement pattern during the push is described by Veeger et al. (24), as starting with flexion of the upper arm from a retroflected position combined with abduction during the first part of the push, which changes into adduction during the last part of the push phase. Part of this movement in the frontal plane is addressed to endorotation (internal rotation) of the arms, which, together with the restricted hand trajectory along the rims, leads to an outward movement of the elbows visible as abduction of the upper arm. These results were obtained from a study (22) on 5 wheelchair athletes, performing exercise tests on a motor driven treadmill with a two or three degree slope and velocity changes from 0.56 to 1.39 $m.s^{-1}$. The current results, however, revealed a dominant exorotation (external rotation) at the shoulder in the first 50% of the push angle in all subjects, and maximal shoulder abduction at hand contact, i.e. the start of the push phase. One possible explanation for this inconsistency might be the position of the subjects in the wheelchair, which was standardised in the current study with respect to the antropometric measures of each individual. Unfortunately, seat adaptations to the subjects' antropometry were not mentioned in Veeger's study (22). Another explanation might be found in the different slopes used in both investigations. Adaptations to slope changes are slightly evident as they shift the centre of gravity of the wheelchair-user system to the back, which implies less stability of the system at the sagittal plane. To compensate for instability, the user may bend the trunk forward and shift the push angle to the front of the handrims, hereby orienting the tangential force more vertically, and preventing the wheelchair from tipping over backwards. Obviously, alterations in trunk inclination may have influenced the movement pattern of the wheelchair-users in Veeger's study (22). As the external rotation of 40 athletes with significant differences in functional ability level always dominated the movement pattern in the first 50% of the push phase, the external rotators of the upper arm, i.e. *M. teres* minor and *M. infraspinatus*, should be included as prime movers in wheelchair propulsion.

CONCLUSIONS

User related parameters such as propulsion technique, force application on the handrims and mechanical efficiency of wheelchair propulsion, are not functional ability dependent in highly trained wheelchair basketball players. Maximal aerobic power is functional ability dependent, however, at least when comparing class I wheelchair basketball athletes to less disabled players. Suggesting a reduction of classes in the wheelchair basketball classification system on the basis of our results would be premature, since other important wheelchair basketball specific parameters such as the volume of action of the players, manoeuvrability with the wheelchair, etc., were not examined.

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