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A COMPUTERIZED WHEELCHAIR ERGOMETER

Results of a Comparison Study

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ABSTRACT. To determine the validity of propulsion simulation on a stationary wheelchair ergometer, nine male able-bodied subjects performed submaximal exercise tests on the ergometer and on a motor driven treadmill (MDT). Oxygen uptake, ventilation and stroke parameters were equal for both devices, but heart rate was lower and trunk movement was less for the ergometer test. Analysis of forces and power output on the ergometer indicated that power output was equal for both wheels. The ratio between applied forces and the effectively directed force component was approximately 80%. Also a small torque was applied by the hand onto the handrim surface which contributed to the total propulsion torque around the axle. It is concluded that the ergometer is capable of simulation of wheelchair propulsion, although the different trunk motion may necessitate sufficient wheelchair propulsion experience. Force analysis results are discussed.

Key words: wheelchair ergometer, motor driven treadmill, submaximal exercise, power output, torque, propulsion technique.

In studies on the optimization of manually propelled wheelchairs, the integrated use of biomechanical models and work physiological parameters is highly useful. Based on simplified inverse-dynamical models and on information about muscular activity, the contributions of muscles and muscle groups to the external power output which is needed for propulsion can be estimated. In relation to specific wheelchair design variations, such contributions may elucidate the reasons why in particular wheelchair designs higher or lower mechanical efficiencies are found.

Generally, in the analysis of complex (cyclic) movements a combined physiological and biomechanical approach has proven to be clarifying. In manual wheelchair propulsion this approach has not often been adopted (2, 5, 16, 17), although interest is increasing (7). Possible explanations for the limited number of studies in this realm are easily found: Firstly, the analysis of kinematics in manual wheelchair propulsion is complicated because of its non-planar character. This makes three-dimensional (3-D) movement analysis necessary. Secondly, instead of more current two-dimensional inverse-dynamics models considerably more complicated 3-D models are desirable. Thirdly, the latter necessitates 3-D measurement of forces. For this purpose a wheelchair ergometer was built which would enable an integrated biomechanical and physiological approach in the study of wheelchair design.

At present most ergometers in wheelchair propulsion research were not constructed for the approach described but for the purpose of studying physiological responses. Studies measuring work of wheelchair-dependent subjects during manual wheelchair propulsion frequently used a testing track or a motor driven treadmill (10, 16, 21). Also, static ergometer designs have been reported on which allowed for the measurement of either average torque or individual torque curves. These ergometers were developed as devices on which subjects can be tested in their own chair, either with the use of rollers (8, 13), or with the wheelchair connected to a bicycle ergometer (1, 9). Alternative devices made use of modular designs consisting out of separate wheels and seat (4, 11, 15, 17). None of the above devices were however instrumented for measuring more than torques on the handrims and resultant speed and thus do not permit the integrated approach advocated here. To measure propelling forces in a 3-D coordinate system the development of instrumented wheelchair wheels as well as static ergometers have been chosen as an option (7, 14). For such a measuring device the following requirements were defined:

- The wheelchair ergometer should allow for the measurement of forces and torques. In the current design this included torques and forces on the handrims and forces on seat and backrest.
- The wheelchair ergometer should allow simulation

Table I. *Relevant anthropometric data of subjects*

′ = film data available, * = force data missing, $^{+}$ = physiology data missing

Subject	Age (yrs)	Size (cm)	Weight (kg)	Rolling resistance (N)
1′	34	187.4	95.8	14
2′	22	179.5	70.3	9
3′*	30	184.0	68.4	12
4′$^{+}$	25	178.9	71.5	12
5′$^{+}$	28	187.2	88.1	14
6	22	177.2	69.5	9
7	30	175.5	68.0	9
8	25	185.5	82.0	12
9	23	185.1	80.9	12
Mean	26.6	182.3	77.2	11
SD	3.9	4.2	9.4	2

of wheelchair propulsion. This meant that it should allow for simulation of frictional losses due to air- and rolling resistance, velocity and slope. Also left and right wheel should be controlled separately. The ergometer should realistically simulate linear inertia of the wheelchair-user system.

- Adaptation of wheelchair dimensions to the subject should be possible. This meant that ergometer dimensions had to be adjustable in terms of wheel camber, width, rim diameter and seat height or seat angle.

On basis of above requirements a wheelchair ergometer was constructed. The final design is described extensively by Niesing et al. (14). It comprised the following elements:

- a mechanical construction which allows for the study of individual characteristics of the wheelchair-user interface. This construction is highly adjustable in a wide range of different positions of handrims, seat and backrest positions and angles without interfering with the instrumentation.
- an electronic control system which allows for the simulation of frictional losses and inertia of the wheelchair user system on basis of feedback. The system also allows for isokinetic measurements. Moreover both rear wheels can be controlled separately, which enables simulation of driving on a side slope.
- a force measuring system which allows for the measurement of applied torques and forces on the handrims of both wheels. Forces can be measured in three dimensions. The system also enables measurement of forces on seat and backrest.

To test the wheelchair ergometer submaximal dynamic tests were performed on both the ergometer and a motor driven treadmill under theoretically equal conditions of speed and workload. Purpose of the test was to determine validity of the simulation of wheelchair propulsion on the wheelchair ergometer. Moreover force and torque measurements were studied on consistency.

METHODS

Subjects

Nine male able-bodied subjects participated. All subjects gave informed consent. Relevant subject data are listed in Table I. Due to measurement errors physiological data of two subjects (Table I, $^{+}$) and force data of one subject (Table I, *) were unavailable.

Protocol

Subjects performed twelve-minute exercise tests on a motor driven treadmill (MDT) and on the wheelchair ergometer. Both tests consisted out of four three-minute bouts at target velocities of 0.56, 0.83, 1.11 and 1.39 m s^{-1} and a theoretically equal workload.

On the MDT tests were performed in a modified basketball wheelchair (Morrien Tornado). The weight of the chair was 20.4 kg. Seat height was standardized at 120° elbow-angle and for-aft position was standardized such that when sitting upright, the subjects' trochanter major was situated approximately 5 cm in front of the wheel axle. Handrims were standard chromium 20.5″ (diameter 52 cm) rims with an internal diameter of 2 cm. Load was imposed by setting a 2° slope. Camber was 2°. Rolling resistance of the wheelchair-user system was determined with the use of a drag test (20). Wheelchair rolling resistance was on average 11 ± 2 N (Table I). From the measured drag force and imposed treadmill velocity external power output P_{out} on the MDT could be determined:

$$P_{out} = (F_r + m * 9.81 * \sin(2°)) * v \quad \text{(W)} \qquad (1)$$

where F_r = wheelchair rolling resistance measured in a drag test (20), m = total weight of wheelchair user + chair, and v = treadmill velocity during the test.

Ergometer dimensions were kept as much equal to the wheelchair dimensions in the treadmill test as possible. However, since it was impossible to obtain a width as small as in the wheelchair, a larger camber angle or 9° had to be used. On basis of previous research (19) it was assumed that this would not lead to significant differences in physiological parameters. To obtain equal levels of P_{out} as imposed in the MDT test, equivalent rolling resistance and slope values were used as input in the appropriate wheelchair ergometer control equations.

Physiology

During tests expired gasses were collected continuously with the use of an Oxycon (Mijnhardt, OX-4). The analyzers for oxygen and carbon dioxide were calibrated before and after

Table II. *Mean values for physiological and kinematic parameters of the treadmill and ergometer tests*

* $p<0.05$, ** $p<0.01$

	Treadmill				Ergometer				ANOVA		
	v1	v2	v3	v4	v1	v2	v3	v4	Device	Speed	Inter-action
P_{out} ($N=9$) (W)	24.0	35.0	47.7	59.7	26.6	36.4	46.7	58.9	NS	**	NS
(SD)	2.7	4.2	5.6	7.0	4.9	5.7	7.1	7.7			
$\dot{V}O_2$ ($N=7$) (l min^{-1})	0.87	1.14	1.48	1.91	0.83	1.04	1.39	1.84	NS	**	NS
(SD)	0.09	0.14	0.15	0.30	0.06	0.12	0.15	0.32			
$\dot{V}E$ ($N=7$) (l min^{-1})	25	32	43	68	23	30	42	63	NS	**	NS
(SD)	3	4	5	18	2	4	8	18			
HR ($N=7$) (bpm)	11:	110	126	147	85	94	113	135	*	**	NS
(SD)	21	21	22	22	12	13	23	28			
SF ($N=9$) (min)	36	41.1	45.3	50.1	40.6	43.9	50.2	54.6	NS	**	NS
(SD)	4.8	5.7	8.2	8.2	5.2	6.9	7.1	8			
PT ($N=5$) (s)		0.69	0.54			0.63	0.52		NS	**	NS
		0.17	0.09			0.11	0.11				
RT ($N=5$) (s)		0.86	0.84			0.77	0.7		NS	NS	NS
		0.3	0.29			0.15	0.23				
CT ($N=5$) (s)		1.55	1.38			1.4	1.22		NS	**	NS
		0.34	0.33			0.24	0.25				
EA ($N=5$) (rad)		2.84	2.97			2.88	2.91		NS	*	*
		0.20	0.15			0.05	0.07				
TR ($N=5$) (rad)		0.16	0.20			0.07	0.10		**	*	NS
		0.09	0.11			0.07	0.08				

each session with a known reference gas mixture. Oxygen uptake $\dot{V}O_2$ (STPD, 1 min^{-1}), ventilation $\dot{V}E$ (BTPS, 1 min^{-1}) and respiratory exchange ratio RER were determined for every third minute of the experiment. Heart rate HR was monitored according to Woude et al. (20).

Kinematics

Halfway in the third minute of the second and third speed condition, five subjects were filmed for at least three cycles (Fs = 50 Hz, DBM-55, Teledyne Camera System). Markers were placed on the anatomical landmarks C7, acromion, epicondylus lateralis, articulatio manus, caput ossis metacarpalia III and on the wheel axle of either wheelchair or ergometer. After digitization (Summagraphics Supergrid), the following parameters were determined: push time PT, recovery time RT, cycle time CT as well as the angle of the hands on the rims with the horizontal at the end of the push phase EA; all according to Veeger et al. (18). From segment markers an estimation of trunk angle range of motion TR was calculated as the angle of a line through C7 and the wheel axle with the vertical.

Ergometer force data

Ergometer data were collected synchronously with kinematic data. Sampling took place over a period of 10 sec with a sample frequency of 50 Hz. From these data torque M and rear wheel velocity v_w for both wheels and force components F_x, F_y and F_z for the right wheel, were selected for further analysis. Prior to determination of maximal values time series were filtered (2nd order recursive Butterworth filter, Fc = 10 Hz).

Power output P_{out} was defined as the product of torque M, wheel radius r_w and rear wheel velocity r_w:

$$P_{out} = M * v_w * r_w^{-1} \quad \text{(W)} \tag{2}$$

where r_w = wheel radius (0.31 m).

From equations (3) and (4) the fraction effective force can be determined:

$$F_{tot} = sqrt(F_x^2 + F_y^2 + F_z^2) \quad \text{(N)} \tag{3}$$

Also determined was effective force on the handrims F_m:

$$F_m = M * r_r^{-1} \quad \text{(N)} \tag{4}$$

where r_r = rim radius, which was 0.26 m.

From equations (3) and (4) the fraction of effective force can be determined:

$$\text{FEF} = F_m * F_{tot}^{-1} * 100\% \quad (\%) \tag{5}$$

FEF was expressed in two ways: FEFpeak as the ratio between peak values of F_m and F_{tot} within each push and FEFmean as the ratio between mean values of F_m and F_{tot} of each push phase.

When information on hand position was available, the

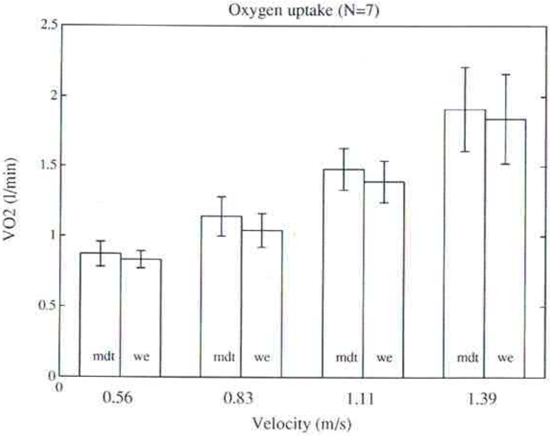

Fig. 1. Mean oxygen uptake ($\dot{V}O_2$) for seven subjects during tests on a motor driven treadmill (mdt) and a stationary wheelchair ergometer (we). Error bars indicate standard deviation.

effective force component F_{eff} of F_{tot} could be calculated from F_{tot} and hand position:

$$F_{eff} = F_x \cos(\alpha) + F_y \sin(\alpha) \sin(\beta) + F_z \sin(\alpha) \cos(\beta) \quad \text{(N)} \quad (6)$$

where α = angle of the third metacarpal on the handrim relative to the vertical through the wheel axle; β = camber angle relative to the above vertical.

The difference between the torque produced by F_{eff} and the torque registered by the torque transducer (M) could be defined as M_h; the torque of the hand on the handrim surface. M_h was calculated as the mean difference between torque M and the torque stemming from F_{eff}.

Statistics

Differences between results stemming from the MDT and ergometer were tested with a two-factor analysis of variance with repeated measurements. Factors in the analysis were "device" (two levels) and "speed" (four levels). Film analysis data comprised two speed levels. Significance level was chosen as $p<0.05$.

RESULTS

Since test conditions were equal for both the ergometer and MDT, P_{out} was comparable for both tests (Table II). Physiological responses indicated no significant differences for either $\dot{V}O_2$ or $\dot{V}E$ and RER. Results for $\dot{V}O_2$ are illustrated in Fig. 1. However, in the treadmill test HR was significantly higher (Table II). Stroke frequency was comparable for both tests. As expected all parameters increased strongly with speed.

Stroke analysis over the five subjects for which film data were available (Table I) indicated no differences in stroke results strong enough to be significant between ergometer and treadmill. Push time PT and cycle time CT decreased in relation to propulsion velocity. As indicated by the significant velocity and interaction effects, end angle EA increased stronger with speed on the treadmill than on the ergometer. The range of motion of the trunk TR was generally small: mean TR was at the most 0.2 ± 0.1 radians ($11.6 \pm 6.5°$) for the treadmill test at 1.11 m s^{-1}. However, despite the small magnitude, TR was signifi-

Table III. *Peak force parameter values calculated for the right handrim*

Two factor ANOVA with repeated measurements was executed over the factors "devices" and "speed". * $p<0.05$, ** $p<0.01$

Speed	(m s^{-1})		0.56	0.83	1.11	1.39	ANOVA
F_m	(N)	(N=8)	109.0	132.3	142.7	138.7	**
(SD)			14.6	19.0	12.8	24.4	
F_{tot}	(N)	(N=8)	133.4	162.6	182.6	173.9	**
(SD)			23.4	26.2	12.5	32.0	
FEF_{peak}	(%)	(N=8)	83	82	78	80	NS
(SD)			9	6	6	6	
FEF_{mean}	(%)	(N=8)	81	78	73	75	*
(SD)			7	7	9	10	
F_x	(N)	(N=8)	55.9	59.1	58.6	52.8	NS
(SD)			7.4	12.7	11.1	10.9	
F_y	(N)	(N=8)	19.0	20.0	27.9	30.3	NS
(SD)			13.2	12.7	15.2	15.4	
F_z	(N)	(N=8)	120.1	148.6	171.5	168.8	**
(SD)			24.3	20.4	9.7	29.8	
M_h	(N m)	(N=4)		0.56	1.95		NS
(SD)				2.16	2.64		

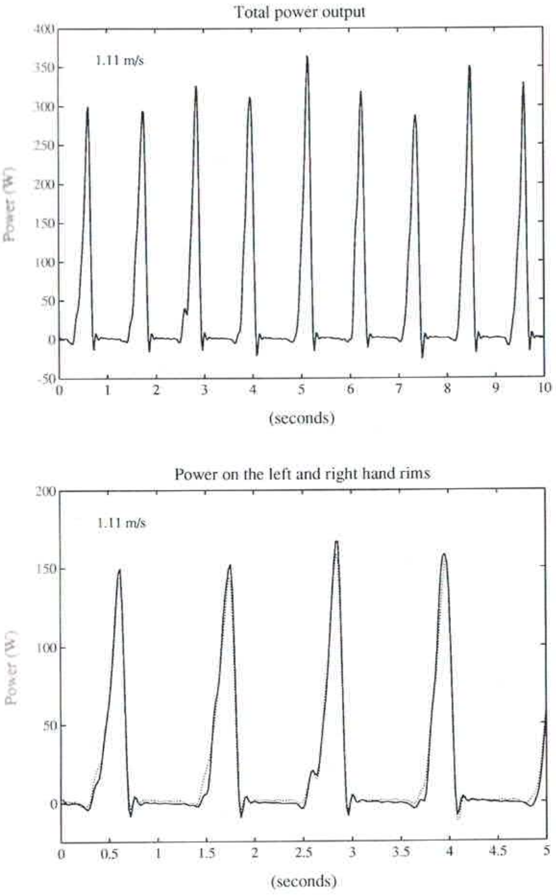

Fig. 2. Typical example over the full 10-sec sampling period of total power output at a target speed of 1.11 m s^{-1} (top). Bottom: separate calculations of power output on the left rim (solid line) and on the right rim (dotted line) plotted over a period of 5 sec.

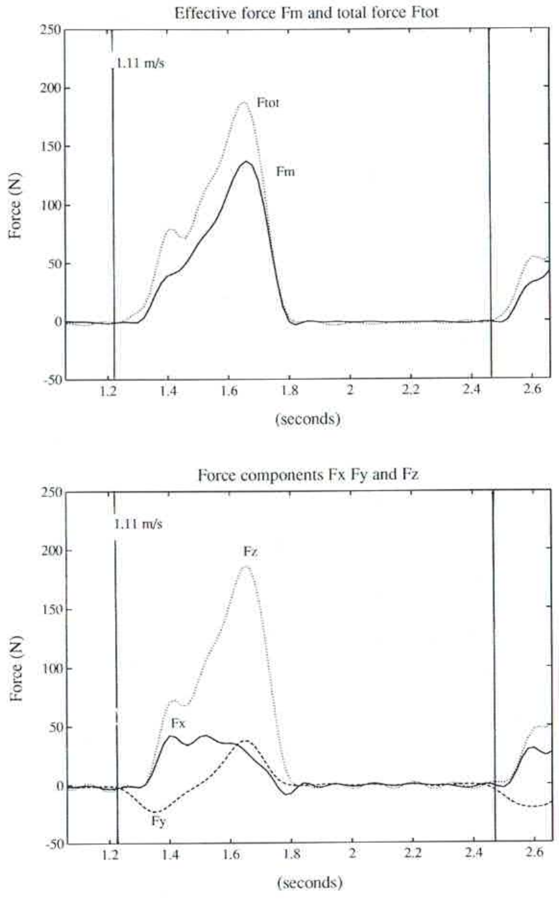

Fig. 3. Effective force F_m and total force F_{tot} over one stroke at a target speed of 1.11 m s^{-1} (top). The lower graph depicts the three force components F_x (solid line), F_y (dashed) and F_z (dotted). Vertical lines indicate the beginning of a push phase.

cantly larger for the MDT than for the ergometer. Moreover TR increased significantly with target speed (Table II).

Ergometer results indicated that differences in mean power output between left and right wheel were not significant. Fig. 2 illustrates total power output and results for left and right handrims separately. Analysis of force values took place for the right wheel only. Peak effective forces F_m and total force F_{tot} increased with speed. Fig. 3 *a* depicts examples of F_m and F_{tot} over a full stroke at a target speed of 1.11 m s^{-1}. Of the force components F_x, F_y and F_z (Fig. 3 *b*) only F_z was found to increase strongly with speed. The low interindividual consistency in FEF values relative to speed led to a small and non-significant decrease in FEFpeak and a significant decrease in FEFmean (Table III). FEFpeak and FEFmean were not higher than 83% and 81%, respectively and decreased with propulsion velocity (0.56 m s^{-1}, Table III).

For the calculation of F_{eff} and M_h the combined film data and force data of four subjects were available (Table I). For three out of the four subjects the difference between peak F_{eff} and F_m was small. However, for one subject F_{eff} was approximately 30% higher than F_m. Due to the small differences between F_{eff} and F_{tot}, M_h was small. At a speed of 1.11 m s^{-1}, M_h was 1.95 ± 2.64 N m, which was again strongly influenced by the results for this one subject (Table III).

DISCUSSION

The physiological comparison between treadmill and ergometer indicated that the wheelchair ergometer properly simulates wheelchair propulsion. Submaximal results between ergometer and treadmill revealed no differences in P_{out}, $\dot{V}O_2$ or $\dot{V}E$ (Table II). However, as is illustrated in $\dot{V}O_2$ results, values for the ergometer seemed to be generally lower than for the treadmill. This trend might be explained by the possibility that the imposed P_{out} on the ergometer had been too low when compared to the actual P_{out} on the treadmill. This would have been the result of an inaccuracy in the drag test procedure with which P_{out} on a treadmill is determined. During the drag test subjects sit in an upright position, while during actual testing they tend to lean more forward. This difference in trunk position increases the pressure on the front casters of the wheelchair and thus increases rolling resistance. In pilot tests we found that this procedure led to an average underestimation of rolling resistance of approximately 10%. It would thus probably have been more correct if simulated rolling resistance in the ergometer had also been increased 10%. As a consequence, correct $\dot{V}O_2$, $\dot{V}E$, SF and HR values would then have been slightly higher.

The limited results concerning the comparison of stroke and timing parameters indicated even as the physiological parameters that simulation of wheelchair propulsion was possible. Trunk movement range TR was however larger for the treadmill than for the ergometer, despite the fact that these movements were relatively small: at the most 0.2 ± 0.1 radians ($11.6 \pm 6.5°$). It is likely that this difference is caused by differences in rotational stability in driving a wheelchair on a treadmill and simulated propulsion on an ergometer. In every day wheelchair propulsion, the application of a large torque on the handrims may lead to involuntary backward rotation of the user and chair. The occurrence of this rotation around the pitch axis is (among others) strongly influenced by the angle at which the center of gravity of user and wheelchair less rear wheels is positioned relative to the vertical through the wheel axle (12). As a consequence a way to prevent the chair of toppling backward when a high resistive force must be overcome is to shift the center of gravity forward by leaning more forward. Since on the ergometer the rearward toppling effect is not apparent, the smaller trunk forward lean was not really needed. Brown et al. (3) reported a more upright position during a wheelchair ergometer test for unexperienced able-bodied subjects relative to experienced wheelchair-dependent subjects. It is not unlikely that this difference was due to a difference in experience in wheelchair propulsion. Some further research seems to be useful.

A practical consequence of the rearward rotational effect will be that in situations in which high propulsion forces are wanted (as in wheelchair sprinting), the angle which the center of gravity makes with the vertical through the wheel axis should be larger to prevent the amount of force being applied to the wheels to be limited because of this rotation. The possible trade-off with increasing rolling resistance due to an increase in weight on the front casters should then be taken into account.

Manual wheelchair propulsion generally seems to be a symmetrical movement. This apparent symmetry was underlined in this study by power output values as measured for the left and right wheel. Not only were mean values identical, time series of both power curves were also well comparable (Fig. 2). Data reduction by one-sided analysis of ergometer data thus seemed to be acceptable.

The most effective direction of forces applied on the handrims is tangential to those rims. FEF values indicated that forces are not optimally directed and that effectiveness decreased with increasing speed (Table III). The latter seems to indicate a decrease in co-ordination of movements as a result of higher handrim velocities. However, within this test power output also increased with speed. Changes in FEF could thus not exclusively be attributed to velocity effects.

The effect of the mean 'internal' torque M_h of the hands on the handrims on the total propulsion torque around the wheel axle was found to be fairly small (Table III). However, for one subject F_{eff} was approximately 30% higher than the effective force F_m measured on the torque transducer. The cause for such a large torque is likely to be related to a different propulsion technique. How this torque is applied and whether the occurrence of such high torques is related to a lower propulsion efficiency will be subject of future research. The existence of a small internal torque in three out of four subjects during propulsion against a relatively high workload of 2° suggests that the contribution of M_h might especially under lower resistance conditions be negligible relative to the forces concerned. This would imply a considerable simplification of measurement and calculation procedures.

CONCLUSIONS

Since apart from differences between HR responses, (submaximal) physiological responses of subjects on a treadmill and on the wheelchair ergometer were comparable, it can be concluded that the ergometer is capable of simulating manual wheelchair propulsion. This device is thus useful for the optimization research approach as intended to follow.

However, rotational stability of a wheelchair around its pitch axis is not simulated. Since results indicate that this difference might affect stroke technique, one should be aware that subjects should adopt a pushing technique they also use during real-life propulsion. As a consequence care must be taken when using subjects without real-life wheelchair propulsion experience on a static wheelchair ergometer. In submaximal wheelchair propulsion effectiveness of directed forces was at the most 83% and decreased with increasing propulsion velocity. The contribution of M_h seemed to be limited. It is possible that under dynamic circumstances and low workloads the contribution of M_h to wheelchair propulsion can be neglected. This would imply a considerable simplification of measurement and analysis procedures.

A detailed technical description of the wheelchair ergometer can be obtained from the authors.

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