

The effect of rear wheel camber in manual wheelchair propulsion

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Abstract—Eight nonimpaired subjects participated in a wheelchair exercise test using a motor-driven treadmill in order to study the effect of rear wheel camber on wheelchair ambulation. The test consisted of four runs with rear wheels in 0, 3, 6, and 9 degrees camber at four speed steps of 2, 3, 4, and 5 km/hr. There were no significant effects upon oxygen cost, heart rate, and mechanical efficiency. The kinematic parameters of push time, push angle, and abduction showed differences between 3 and 6 degrees camber. The relationship between the findings, using surface EMG results for six shoulder muscles, is discussed. For one subject, data were extended to study the angular velocities of shoulder and elbow.

Key words: *biomechanics, camber, propulsion, wheelchair.*

INTRODUCTION

Tilted rear wheels or camber, a popular feature on racing wheelchairs, is being seen more and more on wheelchairs that are used for activities of daily living. One of the advantages of camber is that it provides better lateral static stability for the wheelchair, as a result of the greater distance between the

contact points of the two wheels. This is especially advantageous when frequent sideways movements are needed (e.g., as in wheelchair basketball). Also, as a result of the larger wheelbase, the downward turning moment of the wheelchair decreases on a side slope (2).

Besides the greater stability of cambered wheelchairs, cambered rear wheels provide an easier reach to the handrims and less hampered arm movements during push and recovery movement (4), leading to a less strenuous propulsion technique. It is also suggested that camber would be more efficient due to the effective application of force and lower losses as the result of less arm abduction and stabilization (1). A study on wheelchair characteristics during the 1980 Paralympics showed a trend of increased success of the athlete with increasing camber (7). The most successful athletes had wheelchairs with a camber angle of 7.16 degrees (sd 1.92). Whether this relationship is a causal one is not known.

The objective of this study was to establish whether increased camber leads to a more efficient level of wheelchair propulsion in terms of physiological and movement pattern parameters.

METHODS

Eight nonwheelchair users participated in this study. All subjects gave written informed consent. Relevant data is reported in **Table 1**.

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Table 1.

Personal data of the subjects participating in this study. All subjects were able-bodied.

Subject	Age (yrs)	Weight (kg)	Stature (cm)	Seat Height (cm)	Shoulder Breadth (cm)
JB	21	68.5	181.3	95.0	38.6
PN	27	84.0	179.7	97.2	39.6
BC	22	79.1	185.1	99.3	38.3
LB	24	78.4	178.6	96.0	41.9
GE	25	80.7	185.1	96.7	42.3
MB	29	69.3	184.9	94.6	32.8
FC	23	61.8	177.7	92.3	35.3
ME	27	71.8	180.7	97.7	42.6
Mean	24.8	74.20	181.63	96.10	38.92

Procedure

The experiment consisted of four 12-minute wheelchair exercise tests on a motor-driven treadmill (Enraf Nonius, model 3446, length 3.0 m, width 1.25 m), in which camber varied from 0 to 3, 6, and 9 degrees. Wheel alignment was kept constant as much as possible. Within each test, the belt speed increased every three minutes (0.56, 0.83, 1.11, 1.39 m/s). Prior to testing, the subjects performed a five-minute warm-up (speed 1.11 m/s) followed by a five-minute rest. Tests were performed on a Morrien-Tornado basketball wheelchair (weight 14.5 kg, rim diameter 0.52 m, tires Vredestein DOETO). Seat height was standardized so that the elbow angle was at 120 degrees when sitting in an upright position, where the elbows were in the sagittal plane as much as possible, and the hands were at top dead center of the rims (12 o'clock). Rolling resistance was determined in a drag test (11). Differences in rolling resistance between the camber angles were compensated for during the tests; the external power outputs corresponded with the four different testing speeds set at 0.17, 0.26, 0.35, and 0.44 Watts/(kg_{total weight}) respectively (total weight was the sum of body weight and wheelchair weight). Normalization took place via weights which were connected to the rear of the wheelchair through a system of pulleys (**Figure 1**). The weights added up to the rolling resistance.

Physiology

During each test, physiological analysis took place (Oxycon Ox-4; Mijnhardt). Oxygen uptake (VO, STPD) and respiratory exchange ratio (RER) were determined on-line and printed out every 30 seconds. The last two values of every step (= the last minute of each step) were averaged to calculate results. Heart rate (HR) was monitored with a Lectromed cardiograph (11). Gross mechanical efficiency (ME) was derived from oxygen consumption, corrected for the RER values measured (5) and external power output. The physiological data of subject ME were discarded due to the fact that he used antidiuretics.

Kinematic data

During the third minute of every run, a number of strokes were filmed (DBM-55, Teledyne camera systems, 60 f/s) perpendicular to the sagittal plane of the subject. A mirror was used to enable motion analysis in the frontal plane (see **Figure 1**). The number of strokes was also counted during the third minute.

In order to facilitate analysis, the subjects bore landmarks on the shoulder (acromioclavicular joint), elbow (lateral epicondyle), wrist (midpoint between ulnar and radial styloid), and hand (third metacarpal). The following timing parameters were derived from film: time over which the hands were in contact with the rims and the rims are apparently pushed (push time, PT); time over which the arms return to the starting position (recovery time, RT); the sum of PT and RT (cycle time, CT); the angle over which the rim is pushed (push angle, PA); and, the difference between the start and end angles of the push (SA and EA, defined with respect to the horizontal). All parameters were calculated and averaged over the total number of strokes per session. The abduction (or the angle of the upper arm with the vertical) of each last stroke, was determined as projected in the frontal plane. To obtain arm- and elbow-flexion for one subject, film data of the four camber conditions, and the highest velocity (including sagittal information), were analyzed. Arm flexion was determined as projected in the sagittal plane; elbow flexion was reconstructed from the double two-dimensionally-determined coordinates of the shoulder, elbow, and wrist. The calculated angles were filtered at 6Hz and numerically differentiated to obtain angular velocities.

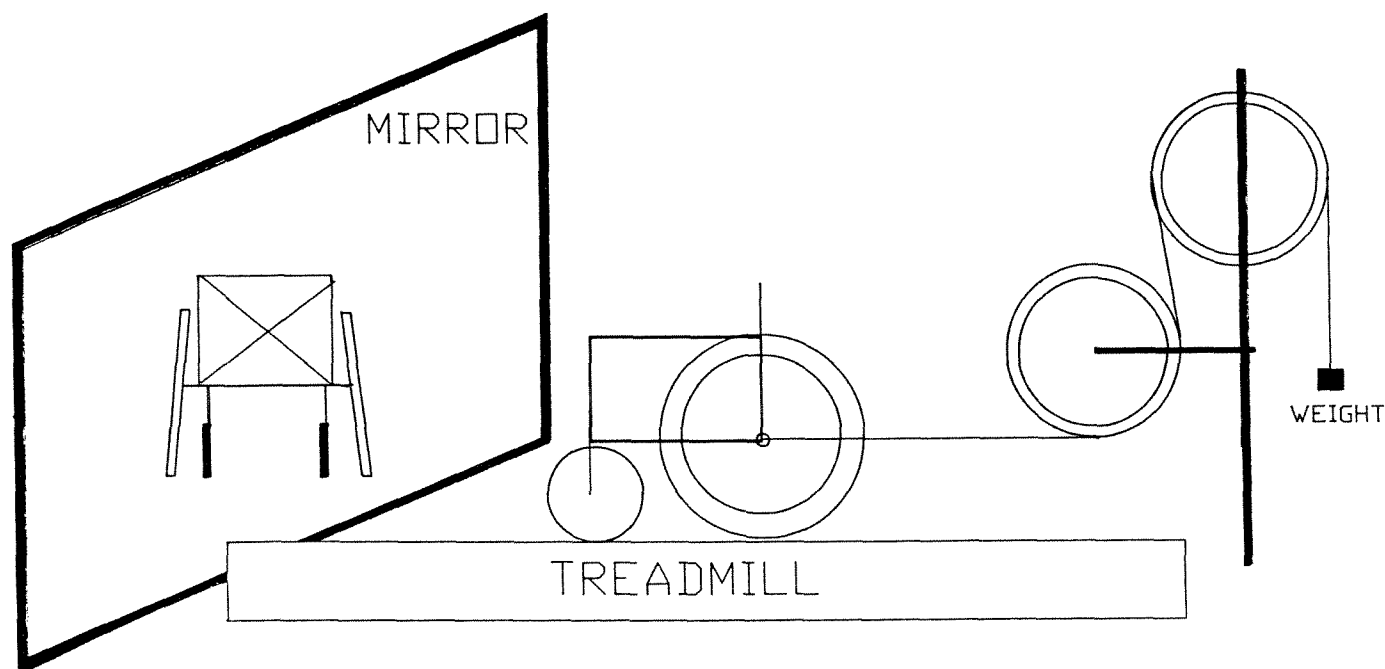


Figure 1.

Schematic representation of the experimental setup as seen through the camera. The pulleys are used for normalization of rolling resistance. The mirror enables analysis in the frontal plane.

EMGs (DISA 15C01, filter 10-500 Hz) of the long head of biceps brachii (BB), lateral head of triceps brachii (TB), descending part of trapezius (TR), anterior and medial parts of deltoid (DA and DM), costal part of pectoralis major (PM), and forearm flexors and extensors for all subjects were registered by surface electrodes (Sentry Medical ECG surface electrodes), and stored on tape for future qualitative analysis (TEAC SR-70). Synchronization of the EMGs with film took place through a pulse which lit a LED visible on film and which was stored on tape simultaneously.

Statistics

Statistical analysis comprised a two-way analysis of variance (ANOVA) with repeated measurements ($N=7$ or $N=8$) or appropriate alternatives. The relationship between the number of strokes counted and the cycle time was established using a Pearson correlation.

RESULTS

The change in camber did result in a minor, but significant, difference in rolling resistance between

camber angles (8.74, 8.3, 8.22 and 7.55 N for 0 to 9 degrees camber). (See **Table 2**.) To exclude the possible effects on the results, the variations in rolling resistance with camber were compensated for in every condition by correction of the imposed power output, using the pulley system described above.

Since the subjects were nonwheelchair users, an ANOVA was performed to check for possible learning effects. No significant influences of experimental sequence on VO and PA for camber angles were found (**Table 2**).

Physiology

The physiological parameters HR, VO, and ME were not dependent on camber, but were highly dependent on belt speed (see **Table 2**, **Figure 2A**, and **Figure 2B**). The average ME ranged from 6 percent at 2 km/hr and 6 degrees camber, to 8.2 percent at 4 km/hr and 9 degrees camber.

Kinematic data

Camber significantly affected SA, PA, and PT. SA was the smallest, i.e., the first touch of the rims takes place farther to the rear, at 6 degrees camber. Three degrees and 9 degrees showed an almost equal

Table 2.

Analysis of Variance (repeated measurements) results for physiological parameters (N=7) and kinematic parameters (N=8).

Parameter	Camber	Belt Speed	Interaction
Rolling resistance	3.21*	(one way)	
Exp. sequence VO	0.88	366**	1.91
Exp. sequence Push angle PA	0.17	5.62**	.63
Heart rate HR	1.48	87.99**	2.91**
Oxygen consumption VO	0.13	364.50**	2.00
Mechanical efficiency ME	0.63	38.68**	1.47
Cycle time CT	0.12	42.40**	0.40
Push time PT	5.73**	137.40**	1.50
Recovery time RT	0.93	5.44**	0.95
Start angle SA	5.42**	1.17	1.29
End angle EA	2.24	12.71**	3.31**
Push angle PA	10.50**	5.62**	2.19*
Abduction recovery	1.34	4.03*	1.34
Adduction recovery	1.76	0.92	1.16
Abduction push	3.80*	3.68*	1.48
Adduction push	0.64	0.82	1.36

*p < .05

**p < .01

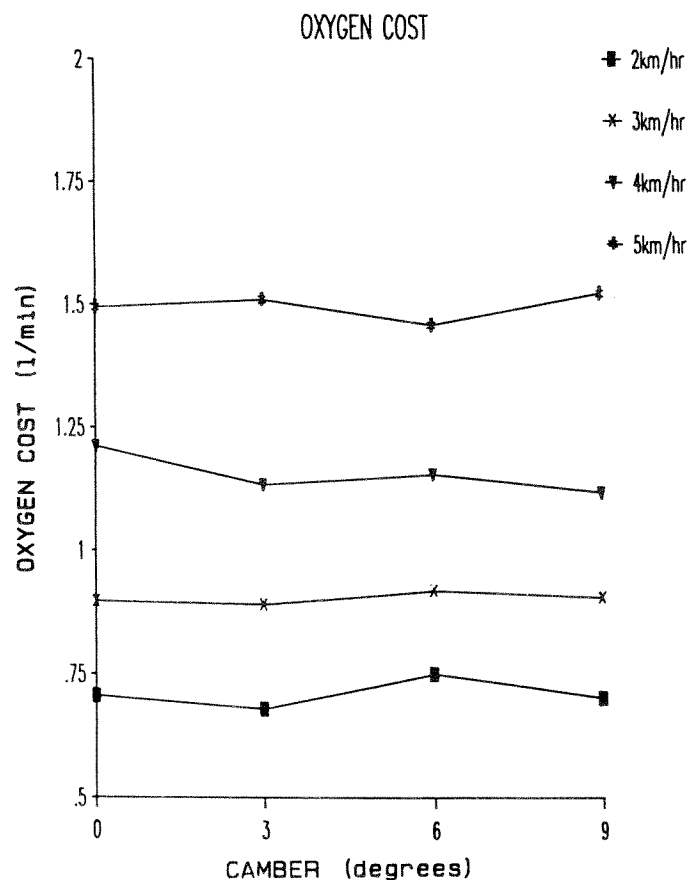
start angle. This relationship was also reflected in the PA, where the longest push was found at 6 degrees camber, reaching 1.87 radians at 5 km/hr. The differences in PA within one speed did not exceed 0.22 radians or 12.6 degrees (at 2 km/hr). SA, EA, and PA are given in **Figure 2**.

PT showed the same relationship as PA; the PT at 6 degrees camber was the longest, and at 3 degrees and 9 degrees camber, the smallest. Recovery time (RT) and cycle time (CT) did not change with camber (see **Figure 3**). CT correlated highly ($r=0.944$; $p<0.001$) with the number of strokes counted over the last minute of each session.

During the push phase, there were no differences in adduction or smallest arm angle. Maximum abduction, or largest arm angle, changed signifi-

cantly with increasing camber. The general pattern was comparable to that for PT and SA: a high value was found for 6 degrees. However, the largest difference was found between 0 degrees and 9 degrees, but was only 0.09 radians (or 4 degrees), at 5 km/hr; 0.74 radians at 0 degrees camber versus 0.65 radians at 9 degrees camber (see **Figure 4**). During the recovery phase, the adduction and abduction of the arms were not dependent on camber.

Figure 5 shows a typical example of the EMG of the shoulder muscles involved, combined with segment angles at a belt speed of 1.39 m/s and 0 degrees and 9 degrees camber. Arm flexion and elbow flexion were calculated for one subject. The EMG results for all the subjects indicated an absence of medial deltoid activity during most of the push phase—especially during that part in which abduction occurs. During most of the recovery phase, DM activity was visible and related to

**Figure 2a.**

Average (N=7) oxygen cost plotted against rear wheel camber angle.

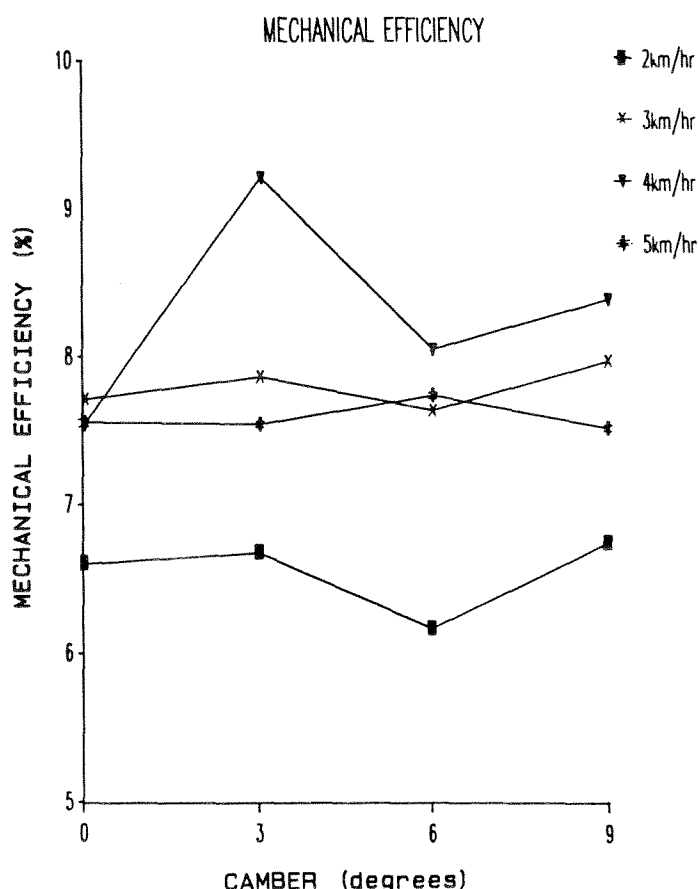


Figure 2b.
Average ($N=7$) mechanical efficiency plotted against rear wheel camber angle.

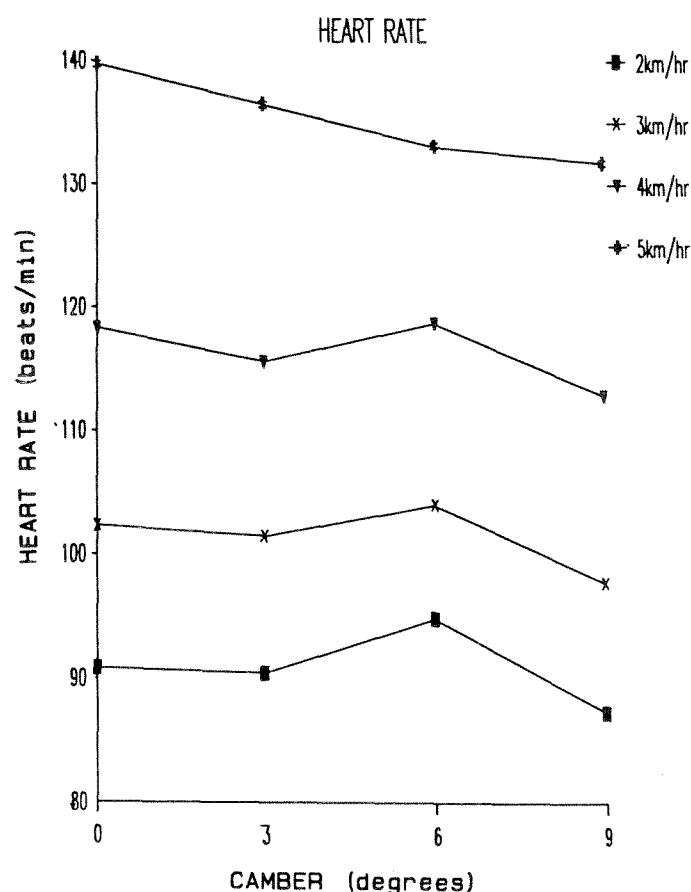


Figure 2c.
Average ($N=7$) heart rate plotted against rear wheel camber.

abduction. No substantial difference in DM activity during the recovery phase could be established between the different camber angles. During the push phase, a clear BB-TB activity sequence could be seen in all but one subject. Both PM and DA had their major activity during the push phase; DA usually somewhat sooner than PM. In all subjects, PM showed no activity during recovery. During recovery, TR and DA were inter-individually less consistent. In some subjects, TR had a silent period during the push; in others, during recovery.

Figure 6 shows the angular velocities of arm and elbow when propelling the wheelchair at 5 km/hr and at 0 and 9 degrees camber. The arm anteflexion peak, and the elbow extension peak, can be found during the last part of the push phase and are sequential: maximum arm flexion is reached before maximum elbow extension. Although the maximum velocities reached are higher at 9 degrees than at 0 degrees (6.21 versus 7.39 rad/s for elbow

extension), there were no consistent differences in maximum velocities related to camber.

DISCUSSION

The finding that rolling resistance decreased with increasing camber, might have been the result of minor changes in wheel alignment that could not be compensated for. However, these differences in rolling resistance were corrected during further tests, and thus did not affect the results.

Physiology

On the basis of this study, it could not be concluded that, for the group and model wheelchair studied, rear wheel camber was physiologically advantageous to vertically-placed rear wheels.

Several alternative explanations can be given for the fact that, contrary to expectations, no effect of camber was found on physiological parameters.

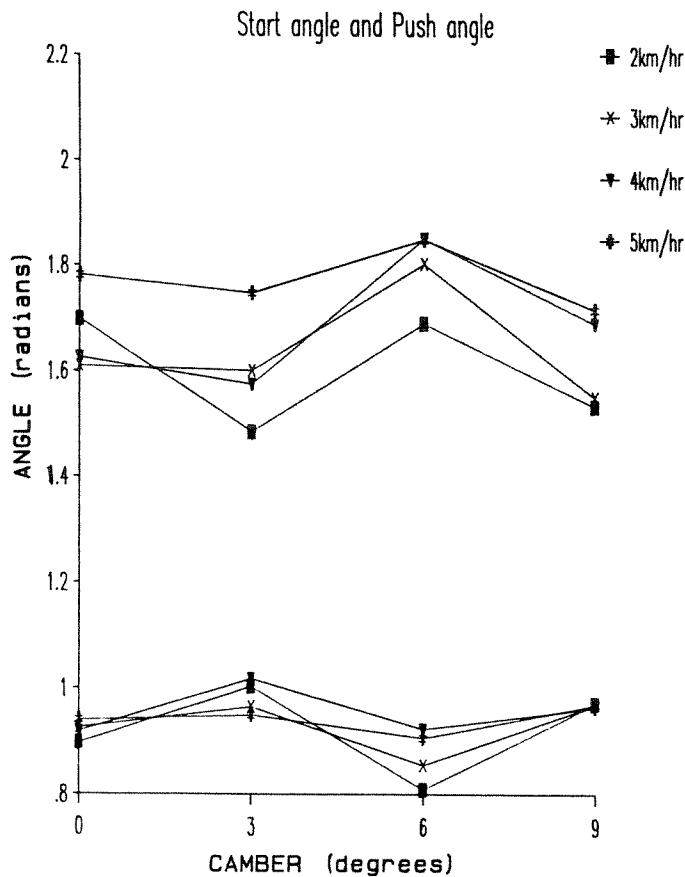


Figure 3a.

Average ($N=8$) start angle and push angle plotted against rear wheel camber. The end angle is the sum of start and push angle, recovery time is the difference between cycle time and push time.

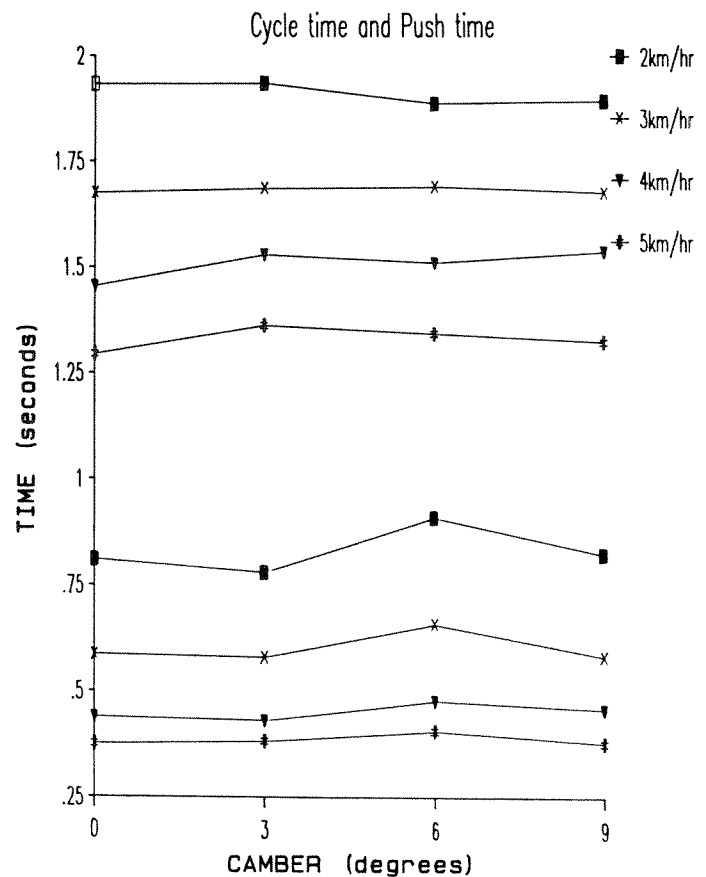


Figure 3b.

Average ($N=8$) cycle time and push time plotted against rear wheel camber. The end angle is the sum of start and push angle, recovery time is the difference between cycle time and push time.

First, the variation might have been contaminated by the differences between subjects in anthropometric dimensions, level of training, or propulsion techniques. More subjects, longer training sessions, or a stricter selection of anthropometric dimensions might have led to significant results. Second, the speed at which the experiments took place was low. However, a 1-Factor analysis for VO and ME at the highest speed, also did not lead to significant results (F -values: $VO=0.57$, $ME=0.31$, $df[3,18]$). Third, based on practical considerations, a basketball wheelchair was used. It is possible that the effect of camber might have been more apparent in a racing wheelchair, with its low seat and thus smaller armpit-to-wheel-top distance.

Kinematic data

The kinematic results did not clearly confirm the expectation that increasing camber facilitates

arm movements. The significantly dependent parameters SA, PA, and PT did not change curvilinearly with camber, but showed a strong difference between 3 degrees and 6 degrees; while between 0 degrees and 9 degrees, no difference was found. This effect is difficult to explain, but it did not show it to be caused by a possible order-effect. It might be possible that the optimum camber angle should be defined as the condition in which the largest push angle is reached. The 6-degree condition can then be seen as optimal, since the trajectory over which force can be applied will be the largest. Since CT does not change significantly, more time will then be available for the production of the same amount of power as in the other camber conditions: thus, possibly, a lower peak torque can be applied. This might lead to a lower local fatigue level which is not necessarily reflected in the general physiological parameters.

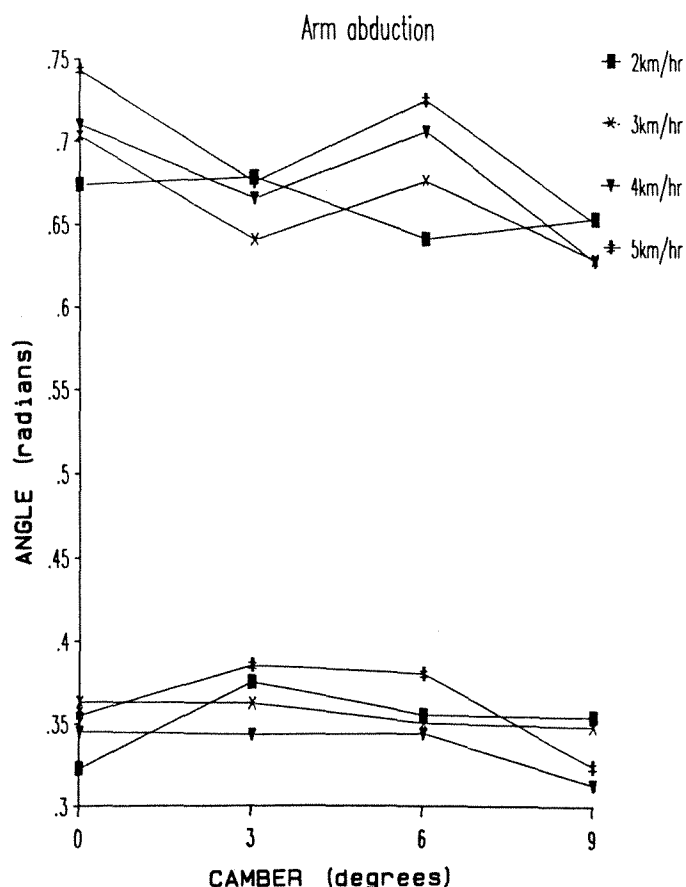


Figure 4.

Average ($N=8$) upper arm maximal abduction and minimal abduction or adduction plotted against camber.

During recovery, neither the return movement of the arms, nor adduction, was affected by the camber angle during the push phase. In this phase, abduction differed only 0.09 radians (approximately 5 degrees) maximally. Since only the camber angle was manipulated, and the wheel top-to-top distance was not kept constant, the top-to-top distance varied from 555 to 475 mm. This change alone may have been responsible for most of the differences in abduction angle. The decrease in abduction thus found may well be negligible.

The suggestion that abduction might influence driving performance could not be supported, because no clear-cut difference was found. However, in this experiment, the seat was relatively high above the wheel axle, and the shoulder high above the rim, in comparison to a racing wheelchair. It might be possible that at a lower seat height, with a smaller armpit-to-wheel-top distance, differences in physiolo-

gy and abduction or adduction due to camber will be found. A follow-up on this study, using a racing wheelchair with a shorter armpit-to-wheel-top distance, and with a constant top-to-top distance independent of the camber angle, again focusing on the influence of camber, is recommended.

The EMG results (Figure 5) showed a generally consistent intra- and inter-individual pattern. This pattern was consistent with that published by Cerquiglini *et al.* (3) on an ergometer, and Tanaka *et al.* (10); but contrary to the findings of Harburn and Spaulding (6).

The fact that DM activity is absent during most of the push suggests that the abduction that occurred during the push was actually forced. Due to the fact that the hands followed the rims from behind top-dead center where the arms were retroflected so as to be approximately horizontal, and that the PM and DA were highly active during that part of the push phase, the elbows were then forced outward as a result of the anteflexing and endorotating torque applied by those muscles. With increasing anteflexion, the adducting torque of PM and DA will increase, and adduction will occur.

As a third advantage of camber, a more efficiently-applied torque on the rims was suggested (1). The torque on the rims could not be measured during this study. However, the possible effect of such differences in torque did not visibly influence the physiological parameters measured. In the near future, the direction of the torque in relation to the rear wheel angle will be studied, using a newly-developed wheelchair ergometer (9).

The angular velocity data for subject PN indicated a sequence in maximal arm anteflexion and elbow extension velocity. If energy transport by biarticular muscles is to take place for the lower extremities, as described by Ingen Schenau (8), such a sequence is necessary. Whether this process takes place can, before long, be studied with the wheelchair ergometer mentioned.

CONCLUSIONS

In this project for cambered rear wheels, no kinematic or physiological advantages were found. The suggested advantages of a less-strenuous propulsion technique, and less abduction of the arms, could not be supported. Cambered wheels do have some advantages, though. They have better lateral

Segment angles and EMG

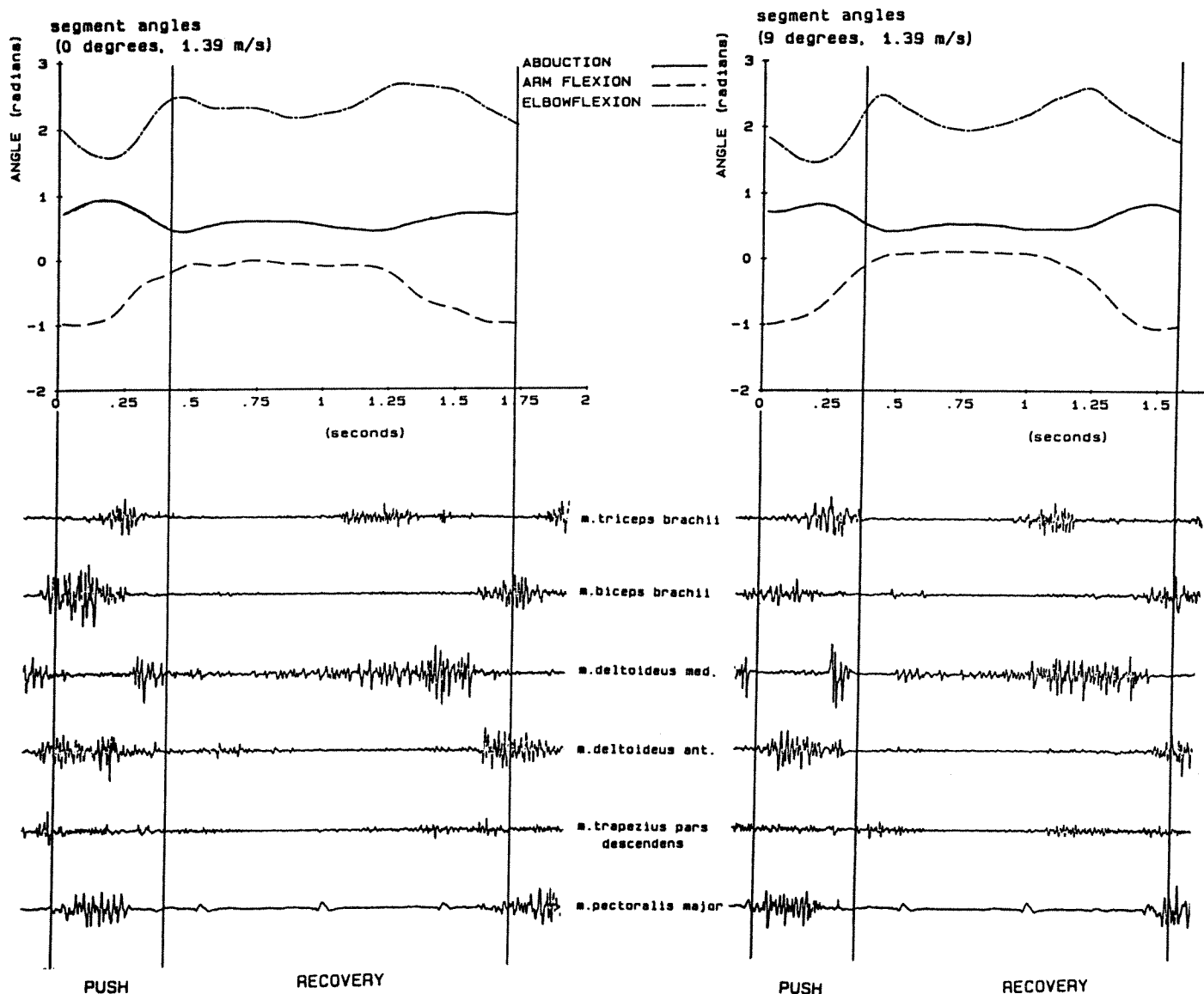


Figure 5.

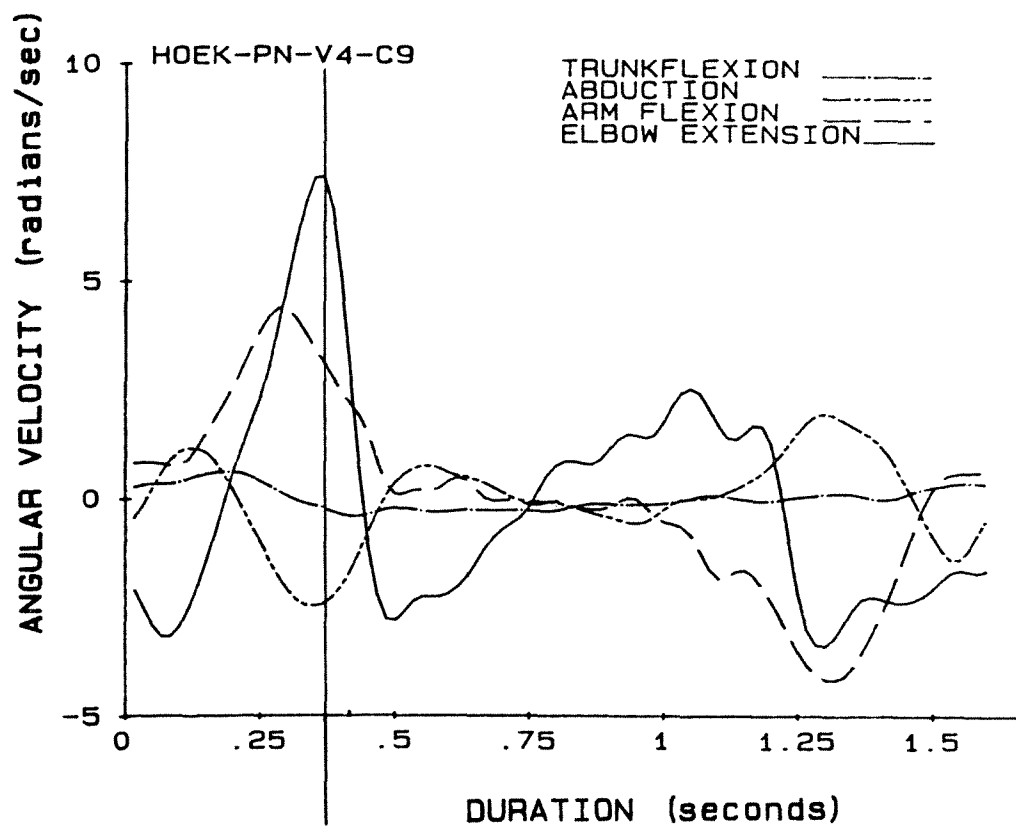
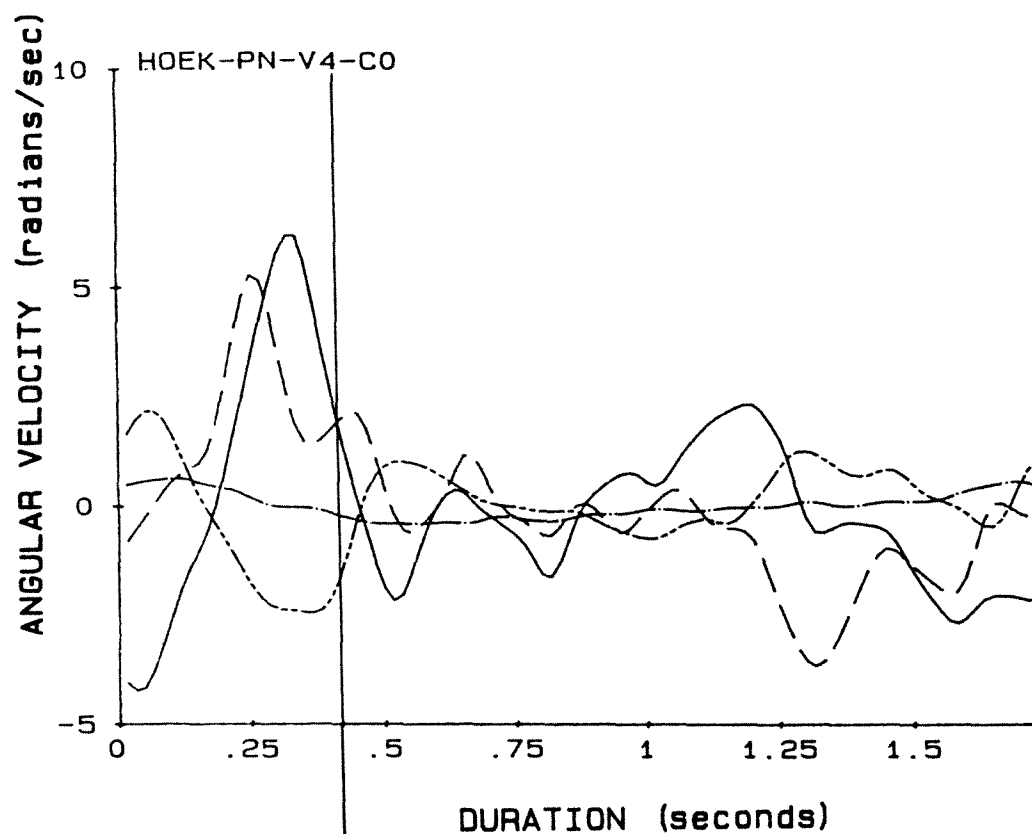
Typical example of the upper arm abduction and the EMG of six shoulder muscles plotted against time. Arm flexion and elbow flexion are pilot results for one subject.

stability, lower rolling resistance, a lower downward turning moment on lateral slopes, and, in turns at higher speeds, there is less stress on the bearings. Also, when used indoors, the hands are protected while moving the chair through doors and along walls. The disadvantages of cambered wheels include: a larger wheelbase that may cause greater difficulty when negotiating narrow passages; greater strain on the rear wheel ball-bearings (4); and a

sharp increase in rolling resistance as a result of the misalignment of the rear wheels when the wheelchair is tilted.

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Angular velocities

Figure 6.
 The angular velocities of the trunk, shoulder (abduction and flexion) and elbows for one subject at 5 km/hr and 0 and 9 degrees camber plotted against time.

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