Mechanical advantage in wheelchair lever propulsion: Effect on physical strain and efficiency

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Abstract—In this experimental study on a prototype lever-propelled wheelchair, the effect of a range of mechanical advantages (MA) on physical strain, oxygen uptake, energy cost, mechanical efficiency, stroke frequency and perceived exertion was examined. Nine out of 10 male nonwheelchair users successfully performed five submaximal tests on a motor-driven treadmill in a prototype bi-manual asynchronous lever-propelled tricycle.

Each test contained the same protocol, but made use of one of the five different MAs. In every test the inclination level increased by 1% every third minute, starting on 0% up to 3%. The velocity was kept constant at 0.97 m·s⁻¹. Variables measured included oxygen uptake, minute ventilation, respiratory exchange ratio, heart rate, and stroke frequency. Analysis for repeated measures was conducted on the main factors slope and MA and their interaction. Additional analysis include a multiple regression analysis. All statistics were conducted with a $p < 0.05$ level of significance.

MA had a significant effect ($p < 0.05$) on oxygen uptake, energy cost, mechanical efficiency, and stroke frequency. These results suggest that the implementation of a range of MAs on a lever-propelled wheelchair may accommodate different external conditions (slope, climatic, surface conditions,

sports, and recreational conditions) and different user groups more readily. This may improve the social radius of action and freedom of mobility of individuals confined to wheelchairs.

Key words: *lever-propelled wheelchair, mechanical advantage, mechanical efficiency, oxygen uptake, physical strain.*

INTRODUCTION

In general, manual wheelchair propulsion is an inefficient form of human locomotion (1,2). Handrim wheelchair propulsion will lead to a relatively high strain on the cardiorespiratory and musculoskeletal systems, resulting in a high energy consumption (En), high heart rate (HR), low mechanical efficiency (ME), and, on the long term, to complaints related to anatomical structures of the upper limb (2–4). These factors, in combination with the general characteristics of the wheelchair-confined population (a physical disability, relatively high age, a sedentary life style, untrained and small muscle mass in arms and shoulder girdle), will lead to high physical strain (short-duration peak loads) and quick exhaustion, which may lead to a more pronounced sedentary lifestyle, which, in turn, may lead to health risks with respect to the cardiorespiratory system (2,5,6). However, different cardiorespiratory responses are seen when different wheelchair propulsion mechanisms are used. Traditionally, crank- and lever-propelled wheelchairs appear to be less

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straining forms of locomotion (3,6–10). The continous motion, use of flexor and extensor muscles, and a less complex coupling of the hands seem the major contributing positive factors (6,11). More recently, the beneficial physiological characteristics of a so-called hubcrank propulsion mechanism (a set of more or less traditional cranks fixed to the hubs of the rear wheels of a racing wheelchair) for track wheelchairs have been stressed (11).

Indeed, lever-propelled wheelchairs lead to lower physical strain compared to handrim-propelled wheelchairs, as has been substantiated in previous studies (6,9,12). Moreover, the lever propulsion mechanism in general also is fairly practical: it is used in Europe and Northern America both in recreation and sports-oriented wheelchairs, as well as in wheelchairs for daily life. Indeed, significantly higher ME has been found using lever-propelled wheelchairs in comparison with handrim wheelchairs (3,6–8,12). This implies that levers allow individuals to propel the wheelchair for a longer duration or at a higher mean velocity, thus increasing the social radius of action or freedom of mobility. Clearly, the use of lever-propelled wheelchairs is increasing with the availability of lightweight materials and contemporary designs. Especially in outdoor and recreational use and under sports conditions, these wheelchairs are suggested to have major advantages in terms of speed and endurance (10). Lever propulsion mechanisms further allow ergonomic optimization to individual physical characteristics on different mobility-related design characteristics: not only with respect to the seat configuration, but also the lever design. One may think of lever length, grip form and orientation, and spatial orientation of the levers as well as the number of gears (thus varying the mechanical advantage). This allows fine tuning of the lever-propelled wheelchair to physical characteristics and aspects of different disabilities as well as to personal requirements and different task conditions. However, little experimental work has been conducted in this realm to date.

Limited information is present in literature on the effects of different gears and/or mechanical advantage on physiology in wheelchair arm work. One study on handrim wheelchair propulsion showed that a range of mechanical advantages (MA) can influence the physical strain and the ME (13). MA is defined as the ratio of the load, or output force, to the effort, or input force (13). A lower MA has been shown to lead to a higher ME and lower physical strain (1,13–15).

The purpose of the current study was, therefore, to compare five MAs on a lever-propelled prototype wheel-chair with respect to the physical strain, different physiological parameters, and subjective strain while propelling the chair at different levels of submaximal power output (different slopes but constant speed) on a motor-driven treadmill. Nondisabled subjects were chosen to study the initial effect of MA on wheelchair arm work per se. The following question was addressed: Is there an optimum MA under the given experimental conditions of wheelchair propulsion?

Since the protoype wheelchair was not adjustable to individual anthropometric characteristics, the additional question was raised whether individual anthropometric parameters of trunk and upper limbs had a significant effect upon the physiological parameters?

METHODS

Subjects

Ten male nonwheelchair users participated in this study on a voluntary basis. The subjects were not specifically trained in the shoulder and arm region. Written informed consent was obtained.

The following personal and anthropometric data were measured: age, hours of sports activity per week, body height (bh), body weight (bw), arm length (al), shoulder-to-seat height (sh) and shoulder width (sw). Sh was the shoulder-to-seat distance: the vertical distance between the acromion and the seat, measured while the subjects were sitting in the wheelchair in a standardized posture (the back against the backrest). The personal and anthropometric data (mean±SD) of the subjects are summarized in **Table 1.**
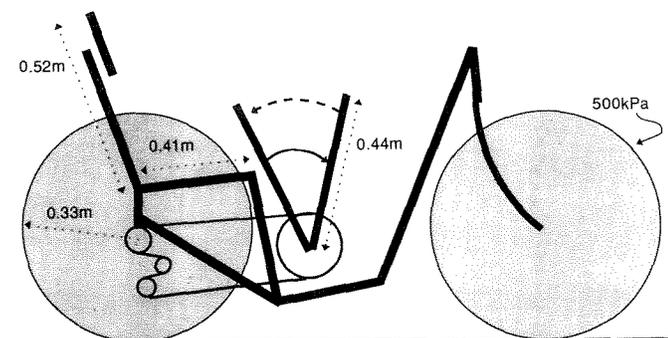
The Wheelchair

The wheelchair used in this study was a prototype three-wheeled, lever-propelled wheelchair designed by Tilley at Technical University, Eindhoven, in 1983 (**Figure 1**). The levers were coupled to the rear wheels by a chain and racing cycle gear box (5 gears; Simplex S001™) mounted to the hub of the rear wheels for the left and right lever separately. The levers were coupled with a Bowden cable, and their orientation only allowed asynchronous arm use, at a fixed position 180° out of phase. Steering of the front wheel was controlled by adjusting the position of the left and right hand grips in the frontal plane. Force application was possible during both the pull and the push phase. The subjects had a free choice in the range of push/pull angle.

Table 1.
Individual characteristics (mean ± SD; n=9).

Age yr	bh cm	Sport hrs/wk	Po (W) 0%	Po (W) 3%	bw kg	al m	sh m	sw m
24.5	1.87	3.4	7.65	39.25	78.4	0.79	0.57	0.412
±2.9	±0.07	±2.1	±0.7	±2.78	±8.8	±0.04	±0.02	±0.02

bh = body height; bw = body weight; hrs/wk = hours sports activity per week; Po = mean power output at 0% and 3%; al = armlength; sh = shoulder height; sw = shoulder width.

Figure 1.
Schematic picture and characteristics of the prototype bi-manual asynchronic lever-propelled tricycle (Tilley, Technical University Eindhoven, 1983) used in this study; weight=33.4 kg, width=0.83 m.

With a lever-propelled wheelchair, the MA can be influenced by varying the length of the lever and/or by varying the gear ratio[1]. In this study, the gear ratio was varied by using the five gear box settings, resulting in five different MAs (**Table 2**) that varied from 0.56 (MA1) to 0.28 (MA5). The length of the lever was kept identical and constant for subjects and MAs. The characteristics of the wheelchair (**Figure 1**) were not adjustable to the anthropometry of the subjects during the tests.

Procedure

Each subject performed five submaximal exercise tests on a motor-driven treadmill (Enraf Nonius, model 3446; 3 m×1.25 m). Two tests were performed on one day, separated by a 30-min rest period. The five tests contained the same protocol, but made use of five different MAs, in random order. Every subject performed the five tests at the same time of the day. Before testing, the subjects were made familiar with the test procedure and instrumentation. They completed test runs with the wheelchair overground and on the treadmill.

[1]W.J.L. Tilley, *Het ontwerp van een recreatie- en racerolstoel* (Design of a recreation and racing wheelchair). Department of Mechanical Engineering, Technical University Eindhoven, 1983.

To enable comparison of the results of the current study with previous experiments (6,9), the subjects propelled the wheelchair at a constant speed of V=0.97 m·s^{-1}. The inclination of the treadmill was increased by 1 percent every third minute, from 0 to 3 percent. The total duration of one exercise test was 12 min.

Physiology

During the test period, the expired air was continuously analyzed with a gas analyser (Ox-4, Mijnhardt). Every third minute, oxygen uptake (VO$_2$, l·min^{-1}, STPD), pulmonary ventilation (Ve, l·min^{-1}, BTPS), and respiratory exchange ratio (RER) were determined. Since the tests were submaximal, a RER<1 was a requisite. The gas analyzer was calibrated before every test with room air, nitrogen, and a reference gas mixture (4 percent CO$_2$ and 16 percent O$_2$).

Heart rate (HR, b·min^{-1}) was monitored electrocardiographically (Lectromed). The three leads were placed on the forehead, on the left, and on the right tibial tuberosity, to avoid electromyographic noise from trunk and shoulder girdle muscles used while propelling the wheelchair (9). Additionally, to determine the overall relative strain in this subject group for the different experimental conditions, the percentage HR reserve (%HRR) was determined according to:

$$\%HRR = (HR-HRrest)/(HRmax-HRrest)*100\% \quad [1]$$

The HR in rest (HRrest) was determined after a 15-min rest period, while the subject was sitting quietly. Peak HR for arm work (HRmax) was determined according to the equation formulated by Sawka et al. (16): HRmax=220−10−age. After each test a rating of perceived exertion (BORG), using a Borg scale with a range of 6–20 (17), was used to determine the individual appreciation of the different MAs. Additionally, a very simple measure of technique, stroke frequency (SF, stroke·min^{-1})

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Table 2.

Mean ± SD for MA1, MA2, MA3, MA4, and MA5 for a slope of 3% and results of the multivariate analysis of variance (MANOVA): P_{Fratio} for main factors Slope, MA and Interaction.

variable	MA1,3%	MA2,3%	MA3,3%	MA4,3%	MA5,3%	P_{Fratio} Slope	P_{Fratio} MA	P_{Fratio} Interaction
VO_2, ($l.min^{-1}$)	1.14±0.16	1.07±0.13	1.14±0.14	1.01±0.16	1.05±0.16	0.000	0.014	0.71
Ve ($l.min^{-1}$)	28.6±5.9	27.8±3.7	27.8±5.0	25±5.1	26.8±5.1	0.000	0.094	0.45
HR ($b.min^{-1}$)	101.3±9.5	98.8±12.7	102.9±9.5	94.9±10.1	97.9±9.5	0.000	0.061	0.44
En ($kJ.min^{-1}$)	23.4±3.5	22.0±2.9	23.3.±2.9	20.7±3.4	21.7±3.43	0.000	0.016	0.64
ME (%)	10.2±1.1	10.8±1.05	10.2±1.0	11.6±1.61	11.1±1.78	0.000	0.031	0.56
RER	0.89±0.06	0.90±0.03	0.88±0.07	0.88±0.06	0.89±0.06	0.000	0.935	0.92
BORG#	11.8±1.9	11.9±2.7	12.3±1.1	11.1±2.8	12±2.2	0.000	0.81	0.11
SF ($1-.min^{-1}$)	41.8±10.3	36.6±7.2	33.8±8.6	30.1±7.7	24.1±4.4	0.28	0.000	0.31

MA = mechanical advantage; En = energy expenditure; ME = mechanical efficiency; VO_2 = oxygen uptake, Ve = ventilation, RER = respiratory exchange ratio, HR = heart rate, SF = stroke frequency; BORG#: Borg scale was evaluated orally and actually is just an ordinal scale, but evaluated here as an interval scale.

was observed in the third minute of every inclination level for every MA.

Power Balance

The mean drag force was determined with a separate drag test at every inclination level for each wheelchair-subject combination. During the drag test, the subject sat passively in the wheelchair while holding the levers in a fixed position. The drag force (F_d) was determined with the use of a strain-gauged force transducer at nine different levels of inclination (from 0 to 8 percent). The wheelchair was connected by a cable to the force transducer. This cable was parallel to the belt in every position of the treadmill. The force signal was amplified and low-pass filtered at 18 dB·octave^{-1}, 0.4 Hz (9). The relationship between F_d and the inclination level of the treadmill was determined for every wheelchair-subject combination with linear regression analysis (r=0.995). With the help of this linear relationship, the mean power output (P_o) at every level of inclination could be determined with Equation 2:

$$P_o = F_d * V \text{ (W)} \qquad [2]$$

where V is the speed of the treadmill (0.97 m·s^{-1}) and F_d is the drag force (N) measured with the force transducer.

Subsequently, the ME was determined with Equation 3:

$$ME = P_o * En^{-1} * 100 \text{ (%)} \qquad [3]$$

where P_o is the mean power output (W) and En is the energy expenditure (kJ·s^{-1}), determined according to Fox

et al. (18) with RER and VO_2, measured every third minute during the tests.

Statistics

The differences between the five MAs for the cardiorespiratory parameters (HR, VO_2, Ve, En, ME), BORG, and the SF were statistically analyzed with an analysis of variance for repeated measures on the main factors MA and inclination level (MANOVA, SPSS·PC+). With this approach a possible interaction could be analyzed as well. Differences were considered significant at p<0.05 when appropriate Bonferonni post hoc tests were conducted, using SPSS·PC+. Clearly, BORG, although actually an ordinal-scaled parameter, is treated here as an interval-scaled parameter.

To examine the possible influence of individual anthropometric dimensions and characteristics on performance parameters, Pearson correlations (Rp) were determined for VO_2, HR, En, ME, and SF with anthropometric parameters on all observations in the experiments (p<0.05). Additionally, stepwise (forward) multiple regression analyses (dependent parameters: ME and VO_2; independent personal, anthropometric, and experimental parameters) were conducted.

RESULTS

Subjects and Protocol

After a brief learning period in negotiating the tricycle on the treadmill, all subjects performed the five tests successfully. One subject, however, was excluded from further data processing since RER values for this individ-

ual exceeded 1.0 and could not be considered submaximal.

External power output varied on average between 7.6 W (± 0.7) at 0 percent slope and 39.2 W (± 2.8) at 3 percent for the nine subjects, determined with Equation 2.

Physiological Data

Manova showed no significant effect for MA on the parameters HR, RER, and Ve. However, the level of significance was almost met for HR (p=0.061; see **Figure 2**). MA significantly affected VO_2, Ve, En, and thus ME (**Figure 3; Table 2**). MA4 showed a significantly higher ME (average value for 4 slopes: 9.1 ± 1.3) and a lower VO_2 and En (average value for 4 slopes: 15.0 ± 2.7) when compared with the other MAs (**Figure 3; Table 2**). BORG showed no significant effect with MA, whereas SF was highly affected by MA (**Table 2**).

For all the parameters but SF, a significant effect was found for inclination level or P_o. As expected, the HR, VO_2, RER, Ve, and En increased with increasing inclination level. The mean value for VO_2 increased from 0.56 l·min⁻¹ at 0 percent to 1.08 l·min⁻¹ at 3 percent, and the mean value for HR increased from 78 b·min⁻¹ at 0 percent to 99 b·min⁻¹ at 3 percent, less than 30 percent of their HRR. Physical strain expressed in %HRR was on average 30.1 percent (± 7.9) at MA1 and 3 percent slope. A somewhat lower value was seen for MA5 (27.3 ± 7.9), which was almost significantly different (p=0.052). ME showed a more or less curvilinear increase with increasing inclination level, up to 10.8 ± 0.9 (average value over 5 MAs) at 3 percent (**Figure 3**).

BORG scores varied between 6 to 12, showing significant increments with inclination level but not with MA.

HEART RATE (beats/min)
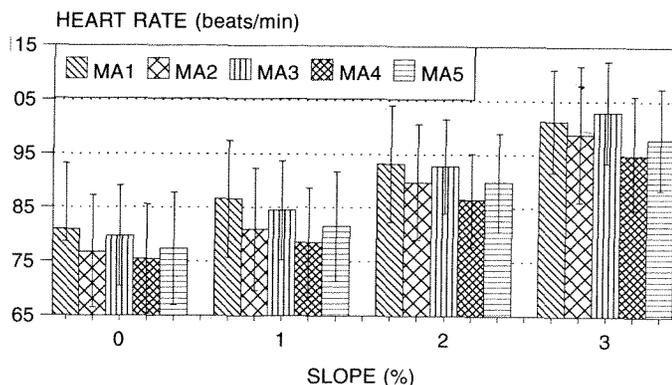
Figure 2.
Effect of the five different mechanical advantages on heart rate (mean and SD) versus the inclination level (percent); V=0.97 m·s⁻¹; motor-driven treadmill; P(Fma)=0.06; P(Fslope)=0.000.

GROSS MECHANICAL EFFICIENCY (%)
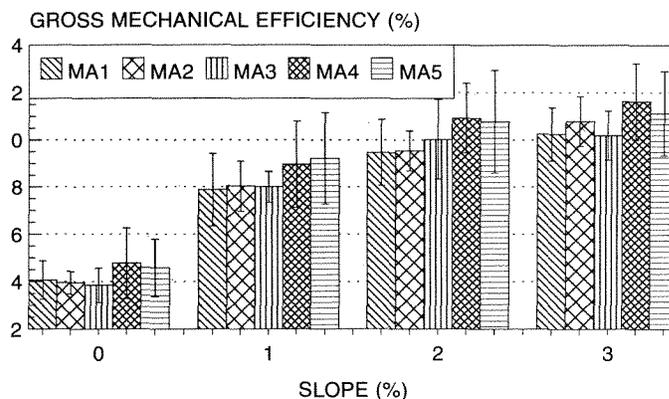
Figure 3.
Effect of the five different mechanical advantages on mechanical efficiency (mean and SD) versus the inclination level (percent); V=0.97 m·s⁻¹; motor-driven treadmill; P(Fma)=0.03; P(Fslope)=0.000.

SF showed an effect for MA (**Table 2**); MA 5 showed the lowest SF (25 ± 1). No effect was found for inclination level, as was to be expected.

No significant interaction effect between the factors Slope and MA has been found for any of the variables (**Table 2**).

Physiology and Anthropometric Data

To have a preliminary indication of a possible relationship between personal and anthropometric parameters, and VO_2, HRR, En, ME, and SF, Pearson correlations were determined for N (= 9 subjects $\times$ 4 slopes 5 = MAs =) 180 observations, followed by multiple regression analysis. The results are presented in **Table 3**. Significant correlations were found for En, ME, and VO_2 with sh. SF was significantly correlated with al and bh. %HRR was related to bh, al, and sw. Clearly, all correlation coefficients were low, due to the repeated use of a low number of subjects.

Multiple regression analysis (stepwise; dependent variable ME; independent variables: age, bh, bw, sh, MA, slope, hours sports·week⁻¹) showed the following result: ME = 0.3 MA + 2.1 Slope − 0.29 SH + 6.85 (R^2=0.74; n=180 observations), indicating again a significant role of MA and slope but also of sh on ME, the latter contributing an additional 5 percent in the explanation of the variance. A similar analysis for VO_2 significantly included slope, MA, and sh, explaining 80 percent of the variance ($R^2 = 0.80$).

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DISCUSSION

Subjects

The subjects in this study were nonwheelchair users. The choice for young male adults was purely methodologic and directed to the objective of the study: analysis of effects of MA upon wheelchair arm work. Since little is known about effects of MA on arm work in general and wheelchair arm work specifically, the initial step was to analyze effects in a small but highly homogeneous subject group with as few disturbing methodological influences as possible. Basically, this comes down to homogeneity of the subject group, not only in terms of age or gender, but especially with respect to the following points: equally (in)experienced in wheelchair propulsion, thus not having a preference for a certain movement regime, MA, or movement frequency; equally (un)trained in arm work and overall work capacity; and finally, of course, a group that has equal ability to work out in terms of cardiovascular and neuromuscular responses. Thus, the intersubject variance will be more reduced and the effects of MA on mere arm work can be more accurately determined, still, however, including effects of interindividual differences of personal and anthropometric characteristics. At this stage, effects of disability are—willingly and knowingly—excluded from the experiment. At a later stage "disablity" will be incorporated in new experiments that study effects of disability upon the relationship between physiological parameters and MA in wheelchair arm work.

Earlier studies have shown that indeed a lower physical strain and higher mechanical efficiencies are seen for proficient wheelchair users compared to nonwheelchair users (9,19,20). This may be explained by experience in wheelchair-propulsion of wheelchair users, functionality, and "use of the lower body" of nonwheelchair users (2). In spite of these differences, a more or less identical trend in physiological data for wheelchair users and nonwheelchair users is generally seen for different work loads (speed, inclination angle), propulsion mechanisms, and cycle frequencies (9,20). It is, therefore, expected that the choice for nonwheelchair users in the current study will lead to similar trends, but at a different absolute level than would be seen in trained wheelchair users: within this framework, nonwheelchair users can usefully participate in tests for trend analysis of cardiorespiratory parameters in relation to different characteristics of the wheelchair-user interface, as is also indicated in other publications where a similar approach is taken (8,12). For comparison with the disabled population the absolute values in the current study must, of course, be treated with caution.

Protocol

The power output, as determined with the drag test, might be slightly underestimated. During this test, the subjects were passively seated and were asked to stay immobile; therefore, no effects of inertia due to arm or trunk motion were seen. Internal losses of energy, due to internal friction of the chain, gear box, and lever mechanism, were not taken into account because the propulsion mechanism was not used (free-wheel condition) during the drag test. Figures of Whitt and Wilson (21) suggest an energy loss of 1.5 percent in well-maintained bicycle systems. Since a double chain mechanism is used, the value may be double in the current study. A small underestimation of power output will imply a small underestimation of ME, according to Equation 3 (9).

The experimental wheelchair is a three-wheeled prototype. The lever mechanism is based on bicycle technology. Left- and right-hand levers are linked 180° out of phase through the coupling of a Bowden cable. Arms move asynchronously, which slightly hampers the steering action. Steering is performed through the orientation of the hand grips, fixed to levers, in a left or right direction. None of the subjects experienced problems with the experimental procedure after the initial familiarization.

The use of bicycle technology in the current prototype will have led to some energy loss in the transfer of the hand force to the rear wheel: setting of the levers, ratchet mechanisms, Bowden cable, and chain. Also, some variation between the different MAs in the magnitude of internal energy loss may have occurred; for instance, through the increased movement frequency at higher MA. The magnitude of this energy loss is unknown, but may be somewhat reduced by using more professional material and through careful fine tuning.

In contrast with traditional lever mechanisms, such as a gripping-roller mechanism (6), the current lever mechanism does not have an idle phase nor does it have a "dead point" at the end of the push or pull phase, as is frequently seen in the more common crank-to-rod lever mechanisms (as a consequence of the lever design, the external force varies sinusoidally with the lever orientation). In this respect, force transfer to the wheels with the prototype is more effective.

Apart from mechanical improvement of the tested prototype, further ergonomic optimization with respect to handgrip, lever orientation, lever length, and seat config-

uration requires future attention for different potential user groups and conditions of the wheelchair use (6,9).

Physiology

The trends in the data for VO_2, HR, Ve, and ME are comparable with data of earlier studies on a lever-propelled wheelchair. The trends seen for physiological data plotted against the power output were quite similar (6,7,9,15,22). The subjects performed the tests at a relatively low %HRR, slightly exceeding 30 percent at a slope of 3 percent. This fairly low submaximal performance level could have resulted in a relatively strong influence of other external factors (temperature variations, emotion), as well as power output and MA on the physiological data. This may account for the almost but finally not significant effect on absolute HR (and %HRR) in relation to MA.

MA was shown to have an effect on VO_2 and ME and, of course, on En. At the same power output level, the ME of lever wheelchair propulsion in this study tended to increase and En tended to decrease as MA decreased. Studies on handrim-propulsion and cycling confirm these results (1,13,14,23,24). Several suggestions have been given in the literature for these effects. A low MA may be more advantageous, because of a reduction in wasted energy from extra limb movements due to a lower SF (1,13). Higher SF goes with more and higher acceleration and decelerations of the arms (20). This leads to a higher energy cost and lower ME. Another suggestion is the force-velocity relationship of the muscles used in wheelchair-propulsion. This relationship shows an optimum, in which the En of the muscle contraction is minimal (25). MA will influence the force-velocity relationship, because both force and velocity at which the levers are propelled change with various MAs. Based on cycling experiments, Gaesser and Brooks (24) stated that fast-twitch fibers, which are energetically inefficient compared to slow-twitch fibers, are recruited as speed of movement increases. An expected increase in SF was found with higher MA (**Table 2**). A study on the distribution of fiber types in human muscles has shown that arm muscles consist of fast-twitch fibers and slow-twitch fibers, as in the leg muscles (26). So, if during arm exercise more energetically inefficient, fast-twitch fibers are selectively recruited as the speed of movement (SF) increases, this may explain an increase in VO_2 and can be an explanation for the differences found in En and ME between the five MAs.

No effect has been shown for MA on RER, HR, and Ve. This may be due to the relatively low submaximal

performance level used in this study (30 %HRR). On the other hand, Veeger et al. (13) found a significant increase for VO_2 and HR with a higher MA in a handrim wheelchair, but this wheelchair had lower gears at a higher propulsion speed and higher power outputs than the one in this study. Also the relative strain (%HRR) in Veeger et al (13) exceeded the values in the current study.

Power Output

Results have shown that ME increased with slope, thus with power output, and the increase becomes smaller with a higher power output. This is in agreement with results of earlier studies on wheelchair-propulsion (2,7,13,14,22,23). This smaller increase is at least partially the result of a decrease in the percentage that the metabolic rate at rest contributes to the total En during that particular load (23).

Anthropometry

Based on the statistical results and the graphical representation, the current physiological results indicate an optimum for MA4, followed by MA5. The variation seen between MA1 to MA5 may partially be explained by differences in personal and anthropometric characteristics of the subjects. Although correlations are low and may be hampered by the low number of subjects (n=9!) and the repeated use of the same subject group over the 180 observations used in the correlations and the regression analysis, the significant results seem to suggest an association between anthropometry and the physiological reactions (**Table 3**). Sh showed a significant correlation with VO_2, ME, and En. Multiple regression analysis indicate the role of sh, where MA, slope, and sh explain 74 percent of the variance. Different shs imply different positions of the shoulder with regard to the lever, which seems to indicate a varying accommodation in propulsion technique, kinematics and muscle coordination between

Table 3.
Pearson correlations (two-tailed) for anthropometric parameters

	bh	al	sh	sw
VO_2	0.11	0.11	0.30***	0.15
HR	0.17*	0.41***	0.24***	−0.07
En	0.11	0.11	0.31***	0.15
ME	−0.06	−0.06	−0.21**	−0.04
SF	0.22***	0.43***	−0.034	−0.16

* = p<0.05; ** = p<0.01; *** = p<0.001; sh = seat-to-shoulder-height; SW = shoulder width; al = arm length; bh = body height.

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different subjects, which affects physiological measures. Differences in sh between subjects could influence the trajectory of contraction within the muscle force-length relation and the interplay between contracting groups of muscles, which consequently affects energy cost (27). Its consequence seems to be summarized in the multiple regression equation presented above. Different studies on handrim-propelled wheelchairs also showed a significant effect of sh on physical strain and ME (27,28). For lever-propelled wheelchairs, Brubaker (28) found no significant effect of sh on ME. However, effect of seat position was studied in a different manner as in the current study. The current results seem to stress the relevance of optimization of the wheelchair-user interface in lever propulsion for future studies.

CONCLUSIONS

Implementation of MAs on a lever-propelled wheelchair will lead to significant differences in ME and En. Moreover, implementation of different gears in a lever wheelchair is practically feasible. Higher MAs showed to be less efficient and showed a higher oxygen cost and En compared to lower MAs at every level of power output. MA4 seemed to be optimal, followed by MA5, but care should be taken since anthropometric variation between subjects may have influenced these results. It seemed that implementation of MA also showed differences in physical strain, but the differences were not significant under the given experimental conditions. This might be due to the relatively low power outputs used in this study. This may also explain the lack of differences in scores on the BORG scale.

The implementation of a range of MAs on a lever-propelled wheelchair might be desirable, because a large variation in external power output exists among the wheelchair-user population. A high gear (low MA) results in high efficiency and is useful at high speeds and long distances, but this requires high propulsion forces that may not be possible for all individuals confined to wheelchairs, due to differences in performance capacities. For fast accelerations, steep inclinations, and for wheelchair users with a low physical condition and/or reduced muscle strength, a low gear may be more appropriate. The prototype lever-propelled wheelchair, subject to experimentation in the current study, is mainly designed for outdoor and recreational use, so further research with a higher propulsion velocity and power output is desirable. At

higher levels of power, output effects of MA will probably be more prominent. Further research of the influence of the anthropometric factors is also desirable.

In this study, the MA has been studied through varying the gear ratio. The MA also must be studied with varying lever length, especially in relation to anthropometry. The prototype lever mechanism also should be further optimized with respect to vehicle mechanics (weight reduction, steering characteristics) and overall ergonomic features. Orientation of the seat, the hand-grip, and the lever in space are important features in this respect. Finally, current results must be verified for different populations of wheelchair users.

REFERENCES

1. Glaser RM, Sawka MN, Young RE, Suryaprasad AG. Applied physiology for wheelchair design. J Appl Physiol 1980: 48(1):41–4.
2. Woude LHV van der, Veeger HEJ, Rozendal RH. Propulsion technique in handrim wheelchair propulsion. J Med Eng Technol 1989:13:136–41.
3. Gangelhoff J, Cordain L, Tucker A, Sockler J. Metabolic and heart rate responses to submaximal arm lever and arm crank ergometry. Arch Phys Med Rehabil 1988:69(2):101–5.
4. Pentland WE, Twomey LT. Upper limb function in persons with long term paraplegia and implications for independence. Paraplegia 1994:32:211–8.
5. Bhambhani YN, Eriksson P, Steadward RD. Reliability of peak physiological responses during wheelchair ergometry in persons with spinal cord injury. Arch Phys Med Rehabil 1991:72(8):559–62.
6. Woude LHV van der, Veeger HEJ, Boer YA de, Rozendal RH. Physiological evaluation of a newly designed lever mechanism for wheelchairs. J Med Eng Technol 1993:17:232–40.
7. Boer YA de, Cambach W, Veeger HEJ, Woude LHV van der. Wheelchair prototype testing: a new lever-drive system in comparison with existing propulsion mechanisms. J Rehabil Sci 1992:5:39–44.
8. Engel P, Seeliger K. Technological and physiological characteristics of a newly developed hand-lever drive system for wheelchairs. J Rehabil Res Dev 1986:23(4):37–40.
9. Woude LHV van der, Groot G de, Hollander AP, Ingen-Schenau GJ van, Rozendal RH. Wheelchair ergonomics and physiological testing of prototypes. Ergonomics 1986:29:1561–73.
10. Maki KC, Langbein WE, Reid-Lokos C. Energy cost and locomotive economy of handbike and row-cycle propulsion by persons with spinal cord injury. J Rehabil Res Dev 1995: 32:170–8.
11. Woude LHV van der, Kranen E van, Ariens G, Rozendal RH, Veeger HEJ. Physical strain and mechanical efficiency in hubcrank and handrim wheelchair propulsion. J Med Eng Technol 1995:19:123–31.
12. Engel P, Knoch M, Moog R, Seeliger K, Spiroergometrische

Untersuchungen mit einem neuentwickelten Handhebelantrieb fuer Rollstuehle. Oesterr Z Phys Med Rehabil 1996:6:2,51–7.

13. Veeger HEJ, Woude LHV van der, Rozendal RH. Effect of handrim velocity on mechanical efficiency in wheelchair propulsion. Med Sci Sports Exerc 1992:24(1):100–7.

14. Brubaker C. Determination of the effect of mechanical advantage on propulsion efficiency with handrims. In: Wheelchair mobility 1983-1984. Charlottesville, VA: University of Virginia, 1985:1–3.

15. Veeger HEJ, Woude LHV van der, Rozendal RH. Loads on the upper extremity in manual wheelchair propulsion. J Electrom Kinesiol 1992:1:270–80.

16. Sawka MN, Glaser RN, Laubach LL, Al-Samkari O, Suryaprasad AG. Wheelchair exercise performance of the young, middle-aged and elderly. J Appl Physiol 1981:50:824–8.

17. Borg G. Psychophysical bases of perceived exertion. Med Sci Sports Exerc 1982:5:377–81.

18. Fox EL, Bowers RW, Foss ML. The physiological basis of physical education and athletics. Dubuque, IA: Wm. C. Brown Publishers, 1989.

19. Brown DD, Knowlton RG, Hamill J, Schneider TL, Hetzler RK. Physiological and biomechanical differences between wheelchair-dependent and able-bodied subjects during wheelchair ergometry. Eur J Appl Physiol 1990:60:179–82.

20. Woude LHV van der, Veeger HEJ, Rozendal RH, Sargeant AJ. Optimum cycle frequencies in handrim wheelchair propulsion. Eur J Appl Physiol 1989:58:625–32.

21. Whitt FR, Wilson GR. Bicyle science, ergonomics and mechanics. London: MIT Press, 1982.

22. Woude LHV van der, Hendrich KMM, Veeger HEJ, et al. Manual wheelchair propulsion: effects of power output on physiology and technique. Med Sci Sports Exerc 1988:20(1):70–8.

23. Powers KE, Beadle RE, Magnum M. Exercise efficiency during arm work: effects of speed and work rate. J Appl Physiol 1984:56:445–9.

24. Gaesser GA, Brooks GA. Muscular efficiency during steady state exercise: effects of speed and work rate. J Appl Physiol 1975:38:1132–9.

25. Åstrand PO, Rodahl K. Textbook of work physiology. Singapore: McGraw-Hill Book Co., 1986.

26. Johnson MA, Polgar J, Weightman D, Appleton D. Data on the distribution of fibre types in thirty-six human muscles; an autopsy study. J Neurol Sci 1973:18:111–29.

27. Woude LHV van der. The wheelchair-user interface: the core of ergonomics? In: Woude LHV van der, Grinten BA van der, Boer YA de, eds. Ergonomics of manual wheelchair propulsion, state of the art. Amsterdam: IOS-Press, 1993:271–92.

28. Brubaker C. Determination of the effects of seat position on propulsion performance. In: Wheelchair mobility 1983-1984. Charlottesville, VA: University of Virginia, 1985:4–6.