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# Handcycling: different modes and gear ratios

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*Handrim wheelchair propulsion is a straining form of ambulation. In contrast, arm crank exercise in laboratory settings has shown a higher degree of gross mechanical efficiency and increased levels of peak power output. Moreover, arm crank exercise can be conducted at different gear ratios and in asynchronic or synchronic mode. Although tricycle crank exercise or handcycling has become increasingly popular for recreational use, sports and outdoor wheeling over the last decade, today little is known about the cardiopulmonary strain in handcycling. The physiological and subjective responses during handcycling were evaluated in a group of 12 male non-wheelchair users (age 24.6 ± 2.7 yr; body weight 73.7 ± 9.7 kg). During an incremental submaximal exercise test on a motor driven treadmill (velocity: 1.8 m s$^{-1}$; an incremental slope of 1% per 3 min; 0–3%; mean power output of the subject group varied between 7.6 ± 1.6 W and 47.5 ± 6.2 W), effects of asynchronic and synchronic crank settings and three different gear ratios (1: 0.42, 1:0.59, 1:0.74 (or 24, 36 and 44 rpm)) were evaluated in a random testing sequence. Significantly lower levels of mean oxygen uptake, ventilation, relative heart rate and oxygen uptake were seen during synchronic arm use and for the lighter gear ratios (i.e. higher movement frequency; 44 rpm). Subjective local perceived discomfort showed similar trends. Conversely, gross mechanical efficiency appeared higher for these conditions. The need for strong medio-lateral stabilizing muscle effort during asynchronic arm use (to ensure a proper wheeling direction as well as simultaneous power transfer to the cranks) and the effective use of the trunk in this subject group may explain the advantage of synchronic arm use. Whether this advantage is consolidated among wheelchair confined individuals needs further study. Apart from the important effects of a shift in force–velocity characteristics of the contracting muscles with varying gear ratios, increased static finger flexor and arm muscle activity may explain the increased strain in the somewhat unnatural heavy gear condition (24 rpm) at the studied velocity. Results need to be re-evaluated for wheelchair user populations and different higher velocities and power conditions. Moreover, other aspects of the wheelchair–user interface must be studied in order to generate optimum fitting and design guidelines for different user groups and conditions of use.*

## Introduction

Handrim wheelchair propulsion is relatively inefficient and stressful to the musculoskeletal and cardiopulmonary systems [1–4]. Different alternative propulsion mechanisms have been available over the past for outdoor ambulation in subjects with lower body disability [5–9]. Especially, lever and crank propelled wheelchairs were frequently used in the first half of the twentieth century as a mode of outdoor ambulation for lower body disabled. Probably due to the ill-designed tricycles of that time on the one hand and mechanization and improving economic conditions on the other, electric and mechanical mobility devices have replaced these manually propelled wheelchairs for outdoor use in the developed countries for many years. Stimulated from within the sports world, a revival of the manually propelled tricycle has been seen during the last decade. Light weight and well-designed, these tricycles appeal to a varied group of active wheelchair users. Currently, even attach-unit arm-crank devices are available to be used in combination with the daily handrim wheelchair [5, 10, 11]. For obvious reasons, tricycles have remained popular over the years in the developing countries [12].

Today, stationary arm ergometry is a frequently used form of upperbody ergometer exercise in clinical evaluation and the study of lower body disabled subjects [1, 2, 6, 13–18].

Submaximal arm-crank ergometer exercise leads to a significant reduction in physical strain, compared to handrim wheelchair propulsion [6, 15–17]. Martel *et al.* [6] studied a group of 20 athletes with paraplegia during wheelchair and arm-crank ergometry and found a significantly higher gross mechanical efficiency for arm-crank exercise compared to wheelchair exercise (16.3% versus 11.6%), which is in accordance with Tropp *et al.* [19]. Also Sedlock *et al.* [20] found significantly higher oxygen uptake, ventilation and heart rate levels in 9 female non-wheelchair users during wheelchair ergometry compared to arm-crank ergometry. According to Glaser *et al.* [1], Martel *et al.* [6], Sedlock *et al.* [20] and McConnell *et al.* [21] peak power output in arm-crank ergometry is substantially higher compared to wheelchair ergometry, but results for peak cardio-pulmonary parameters are contradictory. According to Martel *et al.* [6] peak power output in 20 subjects with paraplegia were on average 97 ± 25 W in arm-crank exercise versus 74 ± 19 W in wheelchair ergometry. Glaser *et al.* [1] presented similar figures for a mixed group of wheelchair users and able bodied individuals: respectively 93 W and 59.5 W.

Until today only few experimental studies were conducted on handcycling or tricycle arm cranking [5, 7–9]. In a field study, Oertel *et al.* [8] showed a reduced physical strain as well as an increased endurance time and average speed during handbike use compared to handrim and lever propulsion in a group of able-bodied subjects. Maki *et al.* [5] compared a row and crank propelled tricycle in seven wheelchair



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users and found no significant differences between the two propelling modes.

Conventional handcycles have used an asynchronous mode of limb movement. Today however, a preference for synchronic arm use is seen in handcycling [10, 11, 22], which is in contrast to the subjectively expressed preference for the asynchronic mode in an arm ergometry study of Marincek and Valencic [23]. Whether a synchronic or asynchronic work mode should be preferred from a physiological point of view is yet unclear. Physiological literature shows contradictory results. In contrast to the expressed preference, Marincek and Valencic found no significant physiological difference between synchronic or asynchronic arm ergometry at 25, 50 or 75 W in a mixed group of able-bodied and spinal cord injured subjects [23]. Glaser *et al.* [2] showed significantly ($p<0.05$) lower cardiopulmonary responses for asynchronous arm-crank ergometry at given submaximal power output levels in comparison to synchronic handrim propulsion. Hopman [24] found a significantly higher gross mechanical efficiency at 30 W submaximal exercise in an asynchronic cranking mode for 10 male able-bodied subjects. At 60 and 90 W submaximal exercise and at peak performance no significant differences between the two modes were seen however. Mossberg *et al.* [14] found no differences at submaximal exertion levels, but showed significantly higher levels of peak power output and endurance in asynchronic arm-crank exercise in a mixed group of able-bodied subjects ($n=6$) and 11 subjects with paraplegia. Engel and Hildebrandt [4] demonstrated that asynchronous lever propulsion reduces cardiopulmonary demands relative to synchronous lever propulsion. Woude *et al.* [25], however, found no significant differences during lever propelled wheelchair exercise in using the arms in a synchronous or asynchronous propulsion mode. None of these studies involved handcycling however. It may be expected that essential differences emerge between (stationary) arm-crank ergometry and handcycling.

An important advantage of handcycles is the application of variable gearing. Different gear ratios give the advantage of better matching a wheelchair task to the physical capability of the user. Powers *et al.* [26] demonstrated a strong influence of gear ratio at equal power output during arm-crank ergometry in 10 able-bodied male subjects. A higher number of revolutions led to a lower mechanical efficiency and higher cardio-pulmonary strain. Fink [9] found crank propulsion with a heavier gear ratio (1:0.73), which resulted in a decrease in the speed of hand movement, compared with a lighter gear ratio (1:1), to be more efficient and less strenuous.

In the current study the effect of synchronic and asynchronic modes of handcycling and the effect of different gear ratios upon cardio-pulmonary parameters and mechanical efficiency during standardized exercise tests on a motor driven treadmill are examined. Based on existing experimental evidence it is hypothesized that a physiological preference will emerge for the asynchronic arm mode and for the heavier (lower rpm) gear setting. The research question is: what is the effect of asynchronic and synchronic arm motion and three different gear ratios (1:0.74; 1:0.59; 1:0.42 (or 24, 36, 44 rpm)) on cardio-pulmonary parameters, gross mechanical efficiency and local perceived discomfort during steady state submaximal handcycling on a motor driven treadmill in a group of non-wheelchair dependent male subjects?

## Methods

### *Subjects*

Twelve healthy male non-wheelchair users (age: $24.6 \pm 2.7$ yr; body weight: $73.7 \pm 9.7$ kg; body length: $1.85 \pm 0.07$ m) participated in this study on a voluntary basis. Previous to all testing each subject expressed his understanding and agreement to participate in the experiment by signing a statement of informed consent.

### *Protocol*

On two different days subjects performed two sets of 3 incremental 12 min handbike exercise tests on a motor driven treadmill (Enraf Nonius model 3446, Delft, the Netherlands; belt width 1.25 m, length 3.0 m, continuous variation of speed $0-5$ m s$^{-1}$ and inclination angle $0-14\%$). They used either the synchronic or asynchronic arm-crank mode. Each of the three exercise tests was conducted at one of three different gear ratios (together with the mode in a systematic counter-balanced order): heavy (1:0.42; 44 rpm), middle (1:0.59; 36 rpm) and light (1:0.74; 24 rpm). Speed of the treadmill was constant ($v=1.8$ m s$^{-1}$) during each test, and the slope of the treadmill increased from 0 to 3%, one percent each three minutes. Subsequent tests were separated by at least 30 min rest, until heart rate was up or close to resting heart rate (HRrest). In the third minute of each workload the cardio-pulmonary parameters were collected, assuming that the subjects were in a steady state condition. Prior to each test, subjects completed a familarization session on the motor driven treadmill with the experimental arm-crank wheelchair and settings (modes and gear ratios; $v=1.8$ m s$^{-1}$; slope $=0\%$; $t=180$ s).

### *Handcycle*

The handcycle (figure 1.; Double Performance, Gouda, The Netherlands) was equipped with six gears and could be used in both synchronic and asynchronic crank positions. The handcycle (aluminium; 16.5 kg) was designed for outdoor use (figure 1.)

### *Power output and gross mechanical efficiency*

Power output ($P_{out}$) was calculated as the product of the drag force and the treadmill belt velocity. The drag force is determined in a separate drag test for each wheelchair–user combination on the treadmill accord-

Specifications and image of the tricycle arm crank wheelchair.

| specifications | measure (m) |
|---|---|
| Seat height | 0.35 |
| Seat width | 0.33 |
| Seat length | 0.42 |
| Seat to footrest | 0.32 |
| Handle length | 0.12 |
| Crank length | 0.18 |
| Circumference wheel | 1.79 |
| Circumference front wheel | 1.88 |
| Distance between backsupport and midpoint crank axis | 0.56 |

*Figure 1. The experimental handcycle.*

ing to Woude *et al.* (7). From $P_{out}$ and oxygen uptake ($VO_2$), the mechanical efficiency (ME) was deduced.

$$ME = P_{out}/\text{En} \times 100\%$$

The internally energy (En) was calculated from $VO_2$ and respiratory exchange ratio (RER) according to Fox *et al.* [27].

*Physiology*

During the exercise test the expired gasses were collected each 30 s with the use of an Oxycon (Mijnhardt OX4, the Netherlands) after calibration with a known reference gas mixture. Oxygen uptake ($VO_2$, $l\,min^{-1}$, STPD), carbon dioxide output ($VCO_2$, $l\,min^{-1}$, STPD), expiratory ventilation (VE, $l\,min^{-1}$, BTPS), breathing frequency (BF, $br\,min^{-1}$), tidal volume (VT, l) and respiratory exchange ratio (RER) were determined for every third minute of each workload.

The heart rate (HR) was recorded continuously during the experimental test at a 5 s storage interval with a sport-tester PE4000 (Polar Electro, Finland). Heart rate responses were expressed relative to the individual heart rate reserve (%HRR), thus reducing interindividual variation in heart rate. The HRR is defined as the difference between resting (HR rest) and peak heart rate (HR peak) [28]:

$$\%\text{HRR} = (\text{HR recorded} - \text{HR rest}) / (\text{HR peak} - \text{HR rest}) \times 100\%.$$

The HR rest was collected during a period of 15 min sitting quietly in a chair. The HR peak during arm work is defined as 210 minus the age of the subject [29].

*Local perceived discomfort*

A subjective measure of local discomfort in the arm-shoulder region (local perceived discomfort [LPD]) was included in the measurements, based upon the Borg-scale (0: nothing at all; 10: extremely strong (almost maximum); [30]). Subjects were asked to give an indication of their local perceived discomfort (LPD) in the shoulder-arm region during the last minute of each 3 min exercise bout.

*Statistics*

Using SPSS 7.5, analysis of variance for repeated measures was used to analyse the effects of the independent variables (mode of propulsion, gear ratio and slope) on the dependent variables (cardio-pulmonary parameters, ME), as well as LPD. Note: LPD was treated as a ratio scale parameter although it is measured on the level of an ordinal scale. Statistical significance was set to $p < 0.05$.

## Results

Eleven healthy male subjects out of twelve were able to perform all tests and workloads. One subject was not able to complete one of the asynchronic tests at the slope of 3%. The missing value at the inclination of 3% during this asynchronic propulsion test was substituted with the value scored on the 3% inclination during the synchronic propulsion test.

$P_{out}$ for the subject group increased from 7.6 ± 1.6 W ($0.10\,Wkg^{-1}$) at 0% slope up to 47.5 ± 6.1 W ($0.62\,Wkg^{-1}$) at 3% slope. % HRR showed values of 31% (synchronic mode, light gear) up to 54% (asynchronic, heavy gear). Table 1 shows means and standard deviations for $n = 12$ subjects of all measured variables for the two modes of propulsion ((A-) Synchronic), gear ratios (Light, Medium, Heavy), and different slopes (0–3%).

Both mode and gear ratio showed significant effects upon cardiopulmonary parameters, ME and LPD. Significant interactions between mode and gear ratio were seen for ME, $VO_2$ and VE. Slope showed significant interactions with mode and gear ratio for the majority of parameters. LPD scores were generally low, but clearly showed parallel trends with the physiological parameters. Statistical results are given in table 2, whereas trends in the group mean results are presented in figures 2–4 for ME, % HRR and LPD.

*Table 1. Mean ± standard deviation of 12 subjects propelling a wheelchair at 2 different modes of propulsion. Legend: A = asynchronic, S = synchronic mode; L = light, M = medium, H = heavy gear ratio.*

| | Breathing frequency (br min$^{-1}$) | Tidal volume (l) | Minute ventilation (l min$^{-1}$) | Oxygen uptake (l min$^{-1}$) | Gross mechanical efficiency (%) | % Heart rate reserve | Local perceived discomfort |
|---|---|---|---|---|---|---|---|
| **AL 0** | 17.0 ± 7.5 | 1.0 ± 0.3 | 15.3 ± 2.2 | 0.57 ± 0.08 | 3.98 ± 0.50 | 16.2 ± 4.7 | 0.04 ± 0.14 |
| **1** | 16.0 ± 3.4 | 1.1 ± 0.2 | 17.2 ± 2.6 | 0.72 ± 0.10 | 6.12 ± 0.68 | 21.2 ± 5.5 | 0.21 ± 0.25 |
| **2** | 18.1 ± 3.8 | 1.3 ± 0.2 | 24.0 ± 5.1 | 0.95 ± 0.17 | 9.96 ± 2.38 | 31.4 ± 7.3 | 1.17 ± 0.55 |
| **3** | 21.5 ± 4.2 | 1.6 ± 0.3 | 33.0 ± 4.4 | 1.35 ± 0.17 | 10.30 ± 1.01 | 44.1 ± 10.8 | 2.58 ± 0.86 |
| **AM 0** | 16.4 ± 5.0 | 1.0 ± 0.3 | 15.6 ± 4.1 | 0.57 ± 0.09 | 4.02 ± 0.33 | 16.2 ± 5.2 | 0.04 ± 0.14 |
| **1** | 17.8 ± 4.3 | 1.1 ± 0.2 | 17.0 ± 2.0 | 0.72 ± 0.09 | 6.17 ± 0.72 | 19.9 ± 5.1 | 0.25 ± 0.25 |
| **2** | 17.9 ± 2.4 | 1.3 ± 0.2 | 23.8 ± 3.3 | 1.04 ± 0.15 | 8.89 ± 0.90 | 31.9 ± 7.6 | 1.46 ± 0.78 |
| **3** | 22.5 ± 5.8 | 1.6 ± 0.3 | 35.6 ± 5.4 | 1.41 ± 0.15 | 9.78 ± 0.77 | 47.6 ± 12.4 | 3.25 ± 1.16 |
| **AH 0** | 17.0 ± 3.9 | 1.0 ± 0.4 | 15.7 ± 3.5 | 0.61 ± 0.10 | 3.82 ± 0.44 | 15.1 ± 4.0 | 0.04 ± 0.14 |
| **1** | 17.6 ± 3.7 | 1.1 ± 0.2 | 18.2 ± 3.3 | 0.76 ± 0.10 | 5.83 ± 0.67 | 21.1 ± 5.4 | 0.50 ± 0.41 |
| **2** | 20.4 ± 4.1 | 1.4 ± 0.3 | 27.5 ± 4.9 | 1.16 ± 0.19 | 7.99 ± 0.96 | 35.8 ± 8.9 | 2.08 ± 1.02 |
| **3** | 23.7 ± 4.9 | 1.8 ± 0.3 | 41.7 ± 5.8 | 1.61 ± 0.18 | 8.54 ± 0.79 | 53.9 ± 11.9 | 4.67 ± 2.05 |
| **SL 0** | 16.4 ± 5.3 | 1.0 ± 0.2 | 14.7 ± 2.3 | 0.55 ± 0.07 | 4.19 ± 0.70 | 12.3 ± 3.3 | 0.04 ± 0.14 |
| **1** | 17.4 ± 6.6 | 1.0 ± 0.3 | 16.5 ± 2.5 | 0.65 ± 0.10 | 6.81 ± 0.77 | 15.1 ± 3.8 | 0.17 ± 0.24 |
| **2** | 18.2 ± 6.0 | 1.2 ± 0.2 | 21.0 ± 3.3 | 0.87 ± 0.14 | 10.72 ± 1.20 | 22.3 ± 5.7 | 0.63 ± 0.51 |
| **3** | 20.1 ± 5.1 | 1.4 ± 0.3 | 30.3 ± 10.8 | 1.15 ± 0.18 | 12.23 ± 1.17 | 31.9 ± 9.2 | 1.75 ± 1.16 |
| **SM 0** | 16.8 ± 5.1 | 0.9 ± 0.3 | 14.7 ± 2.9 | 0.52 ± 0.10 | 4.39 ± 0.65 | 11.9 ± 3.6 | 0.04 ± 0.14 |
| **1** | 17.6 ± 5.2 | 1.0 ± 0.3 | 16.0 ± 2.6 | 0.61 ± 0.09 | 7.27 ± 0.83 | 14.5 ± 3.9 | 0.25 ± 0.32 |
| **2** | 17.8 ± 4.6 | 1.3 ± 0.4 | 20.6 ± 3.4 | 0.87 ± 0.14 | 10.75 ± 1.24 | 22.3 ± 5.0 | 0.75 ± 0.66 |
| **3** | 20.3 ± 4.5 | 1.4 ± 0.3 | 28.2 ± 4.4 | 1.19 ± 0.17 | 11.88 ± 1.13 | 32.9 ± 9.1 | 2.00 ± 1.35 |
| **SH 0** | 16.0 ± 4.0 | 0.9 ± 0.3 | 14.2 ± 2.6 | 0.52 ± 0.09 | 4.38 ± 0.64 | 13.0 ± 4.2 | 0.04 ± 0.14 |
| **1** | 16.3 ± 4.3 | 1.0 ± 0.3 | 15.7 ± 2.9 | 0.62 ± 0.12 | 7.20 ± 1.14 | 16.8 ± 5.7 | 0.17 ± 0.24 |
| **2** | 17.7 ± 4.0 | 1.2 ± 0.2 | 21.0 ± 4.3 | 0.89 ± 0.13 | 10.52 ± 1.22 | 25.2 ± 6.7 | 1.17 ± 0.85 |
| **3** | 21.5 ± 4.7 | 1.5 ± 0.3 | 31.3 ± 8.4 | 1.29 ± 0.20 | 10.95 ± 1.42 | 38.1 ± 12.1 | 2.75 ± 1.48 |

*Table 2. Results of analysis of variance for the main factors (a) synchronic propulsion mode (Mode), gear ratio (gear) and workload (Slope), and their interactions for the physiological variables (ME, $VO_2$, % HRR, VE, BF, VT,). \$: $p < 0.05$; #: $P < 0.01$; *$p < 0.001$; NS: not significant.*

| | Mode | Gear | Slope | Mode × Gear | Mode × Slope | Gear × Slope | Gear × Mode × Slope |
|---|---|---|---|---|---|---|---|
| BF (br min$^{-1}$) | 3.1 NS | 0.9 NS | 30.2 * | 2.3 NS | 2.2 NS | 1.8 NS | 1.0 NS |
| VT (l) | 8.8 # | 1.1 NS | 80.8 * | 0.5 NS | 3.1 $ | 1.7 NS | 0.7 NS |
| VE (l min$^{-1}$) | 87.0 * | 4.5 $ | 158.8 * | 6.0 # | 18.8 * | 4.1 # | 1.7 NS |
| $VO_2$ (l min$^{-1}$) | 47.1 * | 18.0 * | 575.0 * | 11.3 * | 19.8 * | 18.3 * | 1.7 NS |
| ME (%) | 50.2 * | 11.1 * | 473.6 * | 6.9 # | 16.7 * | 9.7 * | 1.5 NS |
| % HRR | 41.6 * | 16.2 * | 117.8 * | 0.1 NS | 21.7 * | 18.5 * | 2.5 $ |
| LPD | 29.7 * | 17.9 * | 84.5 * | 5.0 $ | 23.6 * | 13.5 * | 1.6 NS |

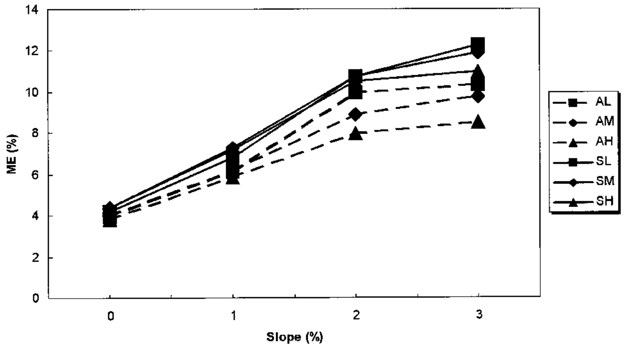


*Figure 2.* *Mean gross mechanical efficiency for 12 able bodied subjects during steady state handcycling on a motor driven treadmill, using the synchronic or asynchronic (A,S) arm mode and the light (L; 44 rpm), medium (M; 36 rpm) or heavy gear setting (H; 24 rpm).*

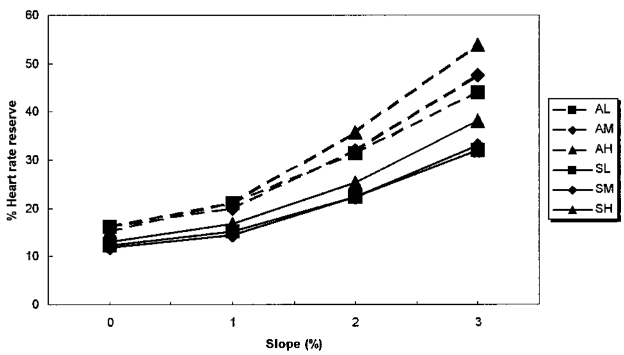


*Figure 3.* *Mean % heart rate reserve for 12 able bodied subjects during steady state handcycling on a motor driven treadmill, using the synchronic or asynchronic (A,S) arm mode and the light (L; 44 rpm), medium (M; 36 rpm) or heavy gear setting (H; 24 rpm).*

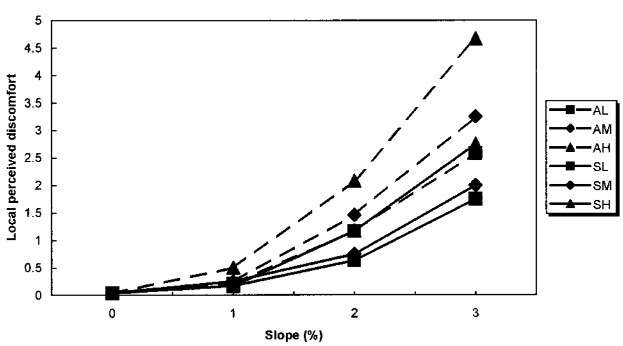


*Figure 4.* *Mean local perceived discomfort for 12 able bodied subjects during steady state handcycling on a motor driven treadmill, using the synchronic or asynchronic (A,S) arm mode and the light (L; 44 rpm), medium (M; 36 rpm) or heavy gear setting (H; 24 rpm).*

*Mode of propulsion*

Synchronic handcycling appeared less straining (lower %HRR, $VO_2$, VE) and more efficient than asynchronic arm cranking ($p<0.01$; tables 1 and 2; figures 2–4.). Similar trends were seen for LPD. The difference between the modes are more marked at the higher slopes. Efficiency varied from on average $4 \pm 0.5\%$ (0%; asynchronic, light gear) to $12.2 \pm 1.2\%$ (3%; synchronic, light).

*Gear ratios*

Overall the light gear ratio showed a higher efficiency in comparison with the medium and heavy gear (24 rpm) use ($p<0.01$; tables 1 and 2; figure 2.). The lightest gear ratio (44 rpm) appears to be the most efficient and subjectively least straining (LPD) under the given testing conditions in both the synchronic and asynchronic mode. No effects of gear on BF and VT were seen.

*Interaction effects*

There are significant interaction effects between gear ratio and propulsion mode for ME, $VO_2$, VE and LPD. This implies that the effect of mode is not constant for the different gear ratios. Interactions between slope and propulsion mode as well as gear ratio are significant ($p<0.01$; table 2) for VE, $VO_2$, ME, %HRR and LPD. Except for %HRR, no significant interaction between mode, gear and slope was seen (table 2).

## Discussion

*Subjects*

Able-bodied subjects participated in the current study instead of wheelchair-dependent persons. This choice was made on the principle that little is known until now of upper body work in handcycling. Thus initial studies should focus on effects in the intact organ system before effects of impairment or training/expertise may be studied and interpreted. Moreover, the able-bodied subject group was not trained in upper body (wheelchair) exercise which makes these subjects equally (un)trained on each of the experimental conditions (mode or gear ratio). Clearly, able-bodied subjects have been frequently studied in arm-crank exercise [14, 24, 26], but also in association with wheelchair use [1, 7, 15, 20]. Indeed, subsequent studies must be conducted to study the possible effects of an impaired upper body, a limited arm–hand function or effects of training on physiological responses in handcycling.

*Protocol*

In the current study a handcycle was used to evaluate the exercise capacity of non-wheelchair users during an incremental submaximal treadmill test with increasing slope at a relatively low speed of $1.8\ m\ s^{-1}$. The handcycle obviously does allow much higher velocities in daily life [8], but speed must be limited during treadmill exercise testing, simply for safety reasons. The protocol with this relatively low velocity ($6.5\ km\ h^{-1}$) will have its impact on the results. The combination of treadmill belt speed and the gear setting determined the number of revolutions per minute (rpm), and thus the eventual linear velocity of the hands. This consequently impacts contraction velocity of the muscles involved.

In the current study the increase in external power output is reached through increments in slope, thus generating sufficiently strong metabolic responses as can be seen in table 1. In some instances, at the 3% gradient RER tended to exceed 1.0, indicating an anaerobic energy contribution. Overall this will have led to some overestimation of ME at 3%.

Up to the mean power output of 47 W at 3%, values for ME do seem to follow the curvilinear trend for stationary crank exercise as described by Martel *et al.* [6] and Powers *et al.* [26], although the current values tend to be somewhat lower. ME compares well however with results on asynchronic handcycling of Woude *et al.* [7] in a group of non-wheelchair users on a treadmill, indicating a substantially higher ME in handcycling compared to handrim wheelchair propulsion under similar conditions of slope and speed. Results on ME of the current non-wheelchair user group are higher than values presented by Janssen *et al.* [22] in a group of spinal cord injured handcyclists at 35 W ($10.6 \pm 0.7\%$; $n = 8$). In these comparisons one must consider however that ME is not only influenced by functional ability and training status but also by gear ratio and speed of propulsion.

As was expected, the incremental exercise test led to a systematic workload effect: an increasing %HRR, BF, VT, VE and $VO_2$ with slope. The sensitivity of LPD in the current experiment for the imposed interventions is in line with the results of Hardison *et al.* [13], but in contrast to the conclusion of Martel *et al.* [6]. Further study on the use of subjective measures in handcycling is required.

Handcycling, as well as arm-crank ergometry exercise, in general is less straining and more efficient than hand rim wheelchair propulsion [6, 7, 15, 21]. The overall more effective use of a larger number of arm and trunk muscles during the push and pull phases seems basic to this notion. Due to the orientation of the arms in space — with the external force vector possibly closer orientated to the centre of the shoulder mechanism — shoulder muscles seem to be less strained in arm-crank exercise. Apart from total active muscle mass (flexors and extensors!), the external force vector orientation may lead to a reduction of the mean strain per unit muscle. Moreover, the gripping action of the hands is more natural and possibly less straining, compared to handrim propulsion. This can explain the increased mechanical efficiency and reduced cardiovascular strain in arm-crank exercise [6, 7]. Within arm-crank exercise, however, different aspects may still be subject to ergonomic optimization [31, 32], such as mode of propulsion and crank rate.

*Mode of propulsion*

In contrast to the initial hypothesis, in the current study synchronic arm use appeared more efficient and less straining. This seems to coincide with the general preference for a synchronic crank setting as is seen in sports and general daily practice of handcycling [10, 11, 22].

One might argue that the beneficial effects of the synchronic arm mode are associated with the larger effective muscle mass of the trunk and the more effective use of the trunk muscles and trunk weight during the push and pull phase. Synchronic arm use seems to allow the weight of the trunk to be used in propulsion in conjunction with a larger muscle mass (trunk flexors and extensors). However, these phenomena seem to be present also in arm ergometry, where even a preference for the asynchronic mode is seen [2, 14, 24].

A different explaining factor seems the strong need for stabilizing muscle effort in asynchronic handcycling. In the asynchronic arm mode there is a need for stabilizing muscle activity of the trunk in response to the rotating effects upon the trunk of the arms propelling the cranks in an alternating manner in combination with the non-stable condition of the crank set and front wheel. The current results seem to be in contrast with (sub)maximal results of Glaser *et al.* [2], Mossberg *et al.* [14] and Hopman [24] on stationary arm-crank exercise, which does seem to make a difference with handcycling. The inherent instability of the crank set as part of the front steering wheel in handcycling may explain the preference for the synchronic mode.

A final explaining factor may be the difference in coasting and steering characteristics of the synchronic and asynchronic modes. In the current study a handcycle was used on a treadmill, which inherently requires steering as well as simultaneous power production through the crank set as part of the front wheel. Asynchronic crank power production seems to influence the steering direction much more than synchronic arm use, in contrast to arm-crank ergometry, where power production does not have a destabilizing effect on the crank set in the asynchronic mode, since it is fixed to the wall or the floor. This may imply (a) a less straight coasting line in asynchronic handcycling, which leads to a longer distance of travel as well as increased rolling friction between tyre and belt. During steering manoeuvers this will lead to an increased power requirement and thus a greater amount of external work as well as energy requirement. There seems to be a need in asynchronic handcycling for extra muscle activity (b) — possibly even in the form of the energetically inefficient process of co-contraction of agonists and antagonists. This should ensure stabilization of the arms with respect to the shoulders in a medio-lateral direction, as well as of the trunk along the longitudinal axis so that it can serve as a stable base for the arms to generate power against. Not only may this lead to co-contractions of muscles at the joint level, but also to 'co-contractions' at the level of the wheelchair – user interface (i.e. handles, crank set, arms, trunk and seat), indeed to keep the handcycle in the proper direction while producing effective work. As a consequence, subjects will perform the test with extra (isometric) muscle contractions of the arm and trunk in the asynchronic mode, in order to solve the steering inaccuracy and instability. This will increase energy cost. Isometric and co-contractions will also increase the overall tension in the specific (power generating) muscles. This may lead to decreased local blood flow in and to the working muscles. This condition is more likely to lead to a reduced oxygen supply to the muscle and an accumulation of metabolic by-products in the blood and may reduce the duration of the exercise [6]. The trends in LPD in the current study may support this notion.

### *Gear*

Based on literature, it was hypothesized that a higher gear (increased rpm) would lead to an increased physical strain and lower ME. The addition of gear ratio variability to wheelchair design allows a better match of the propulsion mechanism to the physical capabilities of the user within the context of different locomotive tasks.

The detrimental effects with a heavier gear (table 2) in the current study are however in contrast with different previous studies in arm-crank exercise [9, 13, 26, 31], as well as lever [33] and handrim use [2]. These studies generally showed a decrease in ME with an increasing number of revolutions per minute. In other words a higher efficiency with a higher mechanical advantage, this is a lower number of rpm or lower velocity of muscle contraction, is generally described. The difference with the current results may be explained with the difference in average arm–hand speed or movement frequency among the presented experiments and thus with differences in the evaluated range of mechanical advantage in relation to the average coasting speed. Coasting speed was indeed low in the current study. Low to very low rpm values were the consequence (lowest 24 rpm; highest 44 rpm) compared to, e.g. Romkes *et al.* ([31] 50 and 70 rpm), Hardison *et al.* ([13]; 50–80 rpm) and Powers *et al.* ([26] 50–90 rpm). With a constant power output the HR and $VO_2$ plotted against the pedal rate is expected to form a U-curve, as is more or less described by Hardison *et al.* [13] and Powers *et al.* [26]. Simultaneously, for ME a hill-shaped curve will be found, indicating that there is a most economical pedal rate with a given power output. This can be exemplified if one combines findings of Powers *et al.* [26] for stationary arm-crank exercise with the results of the current study. In Powers *et al.* [26] at a mean power output of 45 W, efficiency dropped from 15 to 13% when crank rate went up from 50 to 90 rpm. In the current study at a mean power output of 47.5 W and an estimated rpm of respectively 24, 36, 44 rpm, efficiency was respectively 10.9, 11.9 and 12.2%. A physiological optimum may therefore be close to 50 rpm if one combines both studies. As a consequence this will have an impact on the 'position' in the force–velocity relationship of the various muscles involved in this task [34]. With a heavier gear the force to be exerted will increase whereas the velocity of muscle contraction drops. At the given speed in the current experiment this may have had an additional negative effect on the force–velocity characteristics of the muscles involved, since the treadmill belt velocity was rather low. At low contraction velocity a high peak force must be generated to produce an equal level of power output. This may impact local blood flow and again may lead to a reduced oxygen supply and production of an increased level of metabolic by-products. Finally, an increase of cardio-pulmonary strain with a heavier gear may also be explained with the need for increased finger and hand flexor activity to secure the grip on the handle through the 360° circular action at the low speed high resistance condition [35].

### *Optimization continued*

Various studies investigated the effects of crank axle height, crank length and cranking rate in stationary arm-crank ergometry [31, 32], but apart from crank rate no univocal indications in terms of geometry could be stated. Romkes *et al.* [31] did not find any effect of crank orientation, either horizontal to the shoulder or positioned above shoulder height. The role of a chest restraint appeared to be negligible as well in the thirteen healthy males they studied. Up to now no studies have been conducted into crank optimization of handbikes, apart from the current study. Clearly there is a need for detailed studies into crank specifications and orientation in space.

### *Conclusions*

Synchronic arm use is more efficient than the asynchronic mode. The heavier gear however is less efficient under the current conditions. The effective stabilization of both the trunk and arms as well as the crank set and front wheel in handcycling may explain the disadvantage of asynchronic handcycling, together with the steering action in terms of extra distance travelled and increased rolling friction. The increased strain and reduced mechanical efficiency at a lower rpm seems in line with literature if one considers the very low range of rpms studied in the current experiment. The general effect of contraction velocity within the framework of the force–velocity characteristic of contracting muscle seems associated with this gear effect. Future studies must verify the curvilinear association between ME and rpm at a given speed and power output. Moreover, static grip forces of the hand and lower arm may be partially responsible for the increased strain at the lower rpms.

To what extent individual functionality has an impact on (a-)synchronic arm use (and any other design characteristic) requires further study. The increasing and widespread practical use of attach-unit crank systems ('5th wheel' coupling mechanisms)—in addition to the rigid frame handbikes in sports—does require study into subject-related questions of stress, strain and work capacity. Both the cardio-respiratory and musculo-skeletal consequences must be studied for further ergonomic optimization of the wheelchair–user system.

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