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Anaerobic power output and propulsion technique in spinal cord injured subjects during wheelchair ergometry

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Abstract—In order to investigate the influence of the level of the spinal cord injury (SCI) on anaerobic or short-term power production and propulsion technique, 23 male SCI subjects performed a 30-second sprint test on a stationary wheelchair ergometer. Kinematic parameters were studied both inter- and intra-individually. Subjects with a cervical lesion showed a lower mean power output (21.5 Watt, one-sided) than the other subjects; whereas, no differences were found between subjects with a thoracic or lumbar injury (46.9, 63.7, and 49.1 Watt, one-sided). Unexpectedly, no differences were found for the effectiveness of the force applied on the rim between subjects with a cervical injury and the other subjects. It is suggested that the high hand rim velocity reached by subjects with a lower injury cause coordination problems. Reduced arm functionality of subjects with a cervical lesion appeared to cause a higher inward directed force. Arm functionality and rim velocity may have a compensating effect with respect to the effectiveness of force. The kinematics of subjects with a cervical lesion differed strongly from subjects with a lower lesion. Propulsion technique appeared to be intra-individually consistent, which is reflected in the consistency of the force curves, the power output curves, and the movement patterns. Large inter-individual differences in propulsion technique were found. It is concluded that the large diversity in capacity of the SCI population should be taken into account with respect to guidelines and requirements for the environmental space of the SCI population.

Key words: *anaerobic power output, paraplegics, propulsion technique, quadriplegics, sprint test, wheelchair ergometry.*

INTRODUCTION

To improve the freedom of mobility of wheelchair dependent subjects, one can focus on three interrelated issues: first, one can try to improve the mechanical properties of the wheelchair; second, one can focus on improvement of the 'fit' of the wheelchair user interface; and, in the third place, one can try to enhance the physical performance capacity of the wheelchair user (1). When focusing on the latter, it is important to have information on the capabilities of wheelchair users, preferably in relation to influential factors such as impairment, age, sex, or training status.

In the light of the above, the physical performance capacity of a group of spinal cord injured (SCI) male subjects was evaluated as the anaerobic or short-term power production during a 30-second wheelchair test. Usually, physical work capacity has been investigated at a submaximal level or in a

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maximum aerobic exercise test (2–5). However, as was indicated by Janssen, et al. (6), a large number of daily activities in wheelchair users are of a short but intensive character, and especially seem to stress the anaerobic metabolism.

Anaerobic power production in wheelchair arm work has not been studied often. Most studies focused on wheelchair athletes (7,8), whereas the anaerobic power output of the sedentary wheelchair user with a spinal cord lesion has, to the knowledge of the authors, not been investigated. To improve the knowledge base on anaerobic wheelchair arm work, it is of the utmost importance to study different groups of sedentary wheelchair users. In the current study, the influence of the level of the spinal cord injury on anaerobic performance of subjects with a sedentary lifestyle is investigated.

Wheelchair propulsion is a complex form of arm work. Differences in the neuromuscular system as a consequence of differences in lesion level are expected to influence overall functionality, and as such the anaerobic power production and the characteristics of propulsion technique. To evaluate the effect of lesion level and propulsion technique during anaerobic wheelchair arm work, different kinematic characteristics of wheelchair propulsion technique were studied both inter- and intra-individually. To analyze propulsion technique in terms of force application, the propulsion forces on the rim were measured. It was expected that subjects with a diminished arm functionality would have a lower effectiveness of force application.

METHODS

Subjects

Twenty-three male subjects with SCI voluntarily participated in this experiment after having given written informed consent. On the basis of lesion level, the subject population was divided into four groups (group I: C4–C8 (n = 6); group II: Th1–Th5 (n = 5); group III: Th6–Th10 (n = 5); and group IV: Th11–L4 (n = 7). Six subjects had an incomplete lesion (3, 1, and 2 subjects of group I, III, and IV, respectively).

The characteristics of the subjects, including their peak oxygen consumption (VO_2peak) measured in a separate maximal exercise test (6), are given in **Table 1.**

Wheelchair Ergometer

The 30-second tests were performed on a custom-built wheelchair ergometer. This ergometer is a stationary, computer-controlled wheelchair simulator, that allows for direct measurement of propulsive torque around the wheel axle, propulsive force applied on the handrims, and resultant velocity of the wheels. Its final design and technical specifications are described extensively by Niesing, et al. (9). The wheel and handrim radii were 0.31 and 0.26 meter, respectively. During each test, torques, forces, and velocities were measured over the full 30-second test period, with a sample frequency of 65 Hz.

Protocol

After a 3-minute warming up period the subjects performed two 30-second sprint tests on the ergometer. Each of the sprint tests had a rolling start. Based on lesion level, age, and sport activity, resistance level was individually applied at 0.25, 0.50, or 0.75 $N \cdot Kg^{-1}$, and was chosen such that peak propulsion velocity was expected to stay below 3.0 $m \cdot s^{-1}$. The first sprint test was used as a practice trial. Required corrections in the resistance level were made before the second test, which was used for data analysis. The second test was performed after a rest period of 8 minutes.

Wheelchair ergometer settings were individually adjusted. Seat height was standardized at 110° elbow flexion (180° defined as full extension) with the subjects' hands on the top of the rim, and the shoulders (acromion) directly above the wheel axle (10). Rear wheel camber was set at 4°. Seat angle and back rest angle were set at 5° to the horizontal and 15° to the vertical axis, respectively.

Two-dimensional video recordings were made of the right hand side of the body, with the camera perpendicular to the sagittal plane of the subjects. Video recordings were used for a limited analysis of a selection of kinematic parameters, such as push time and recovery time, or begin and end angle of the push.

Power Output

The output signals from the ergometer were recursive low-pass filtered (Butterworth recursive, 11 Hz). As a consequence of resonance in the system, the medio-lateral directed force (Fy) had to be filtered at a lower frequency (Butterworth recursive, 4 Hz).

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Table 1.
Group means and standard deviations of personal data.

Group	I (n=6) Mean (sd)	II (n=5) Mean (sd)	III (n=5) Mean (sd)	IV (n=7) Mean (sd)	p-value
Age (years)	37.3 (9.5)	37.8 (8.4)	26.0 (3.0)	36.0 (12.5)	ns
Weight (kg)	82.5 (17.7)	82.6 (11.9)	78.4 (11.6)	77.6 (15.9)	ns
VO_2 peak (litre·min^{-1})	1.06 (0.10)	1.56 (0.23)	2.02 (0.16)	2.00 (0.43) (n=6)	***

***$p<0.001$; ns: not significant

From the measured torques and wheel velocities, the power output (P) was calculated for the right side only:

[1] $$P = M \cdot V_w \cdot r_w^{-1} \quad (W)$$

where: M = Torque on the handrim
V_w = velocity of the wheel
r_w = wheel radius

Power, torque, and velocity were averaged over the full 30-second period (P30, M30, V30). From the full time series (minus the first three strokes that formed the start), the three strokes with the highest peak power were selected for analysis. From these strokes the parameters Pmax, Vmax, and Mmax were determined.

Force Application

In addition to the above, forces applied on the rim were determined (Fx-cycle, Fz-cycle, and Fy-cycle) from the complete cycles of the three selected strokes. The positive forces applied with the hand on the rim were defined as follows: Fx: horizontally forward, Fy: horizontally outward, and Fz: vertically downward.

From the measured forces, the following parameters were calculated for the right side only:

—From force components Fx, Fz, and Fy, the momentary total force vector (Ftot) was calculated over 30 seconds.

[2] $$Ftot = \sqrt{(Fx^2 + Fz^2 + Fy^2)} \quad (N)$$

Maximum of Ftot over 30 seconds was defined as Ftot-max.

—Fy-min and Fy-max were calculated as a percentage of the maximal total force:

[3] $$(\text{Fy-min or Fy-max} \cdot \text{Ftot-max}^{-1}) \cdot 100 \quad (\%)$$

—From M and rim radius (r_r), the effective force on the handrims (F_m) was calculated:

[4] $$F_m = M \cdot r_r^{-1} \quad (N)$$

—From Equations [2] and [4], the fraction effective force of the total force was determined over 30 seconds:

[5] $$FEF30 = F_m \cdot Ftot^{-1} \cdot 100 \quad (\%)$$

Kinematics

Movement analyses were performed with video recordings (Camera: Panasonic M5, shutter 1/1000, Fs=25 Hz). To facilitate digitization, land marks were positioned on the hand (third metacarpal), wrist (caput radii), elbow (epicondylus lateralis), shoulder (most ventral part of the acromion), and trunk (processus spinosus C7). All kinematic parameters were calculated over three strokes: the stroke with the highest peak power, the one preceding, and the one following this stroke.

Cycle time (CT) and push time (PT) were determined from video. PT was defined as the amount of time that the hand appeared to be in contact with the handrim. CT was defined as the period of time from the onset of one push phase to the next. PT was also expressed as a percentage of CT. This relation was determined as:

[6] $$PT/CT = (PT \cdot CT^{-1}) \cdot 100 \quad (\%)$$

The following push parameters were determined: begin angle (BA), end angle (EA), stroke angle (SA), and trunk angle (TA) as shown in **Figure 1**. The movement pattern of the hand was analyzed with the aid of the marker on the third metacarpal (MCIII).

Statistics

Differences between groups were analyzed with a one-way analysis of variance (ANOVA). The Tukey *post hoc* test was used to locate significant differences. To investigate the association between velocity and kinematic parameters, Pearson correlation coefficients between mean velocity and stroke parameters were calculated. Significance level was set at $P<0.05$.

RESULTS

Subjects

The subject data are listed in **Table 1**. Lesion groups were found not to differ significantly in age and weight. Peak oxygen consumption differed significantly between group I and the other lesion groups.

Power Output

Table 2 shows the results of power output data for the different lesion groups. As indicated by the standard deviations of the power parameters, large inter-individual differences were found within all lesion groups. Despite this, P30 and Pmax for the subjects of lesion group I were found to be significantly lower than for the other lesion groups. For P30, no differences were found between lesion groups II, III, and IV; whereas, Pmax differed significantly between group II and III (248.7 and 398.1 Watt, one-sided, respectively). Lesion group III achieved the highest P30 (63.7 Watt, one-sided), which was almost three times the power output delivered by group I (21.5 Watt, one-sided). The individual values for P30 ranged from 13.1 Watt, achieved by a subject with a cervical lesion, to 81.8 Watt (one-sided) for a subject with a mid-thoracic lesion.

The mean and maximal torque, and velocity showed the same pattern between groups as found for the power output; the lowest values for M30 and V30 were achieved in lesion group I (4.6 N·m and 1.45 $m \cdot s^{-1}$, one-sided), the highest values in lesion group III (8.5 N·m and 2.35 $m \cdot s^{-1}$). M30 and V30 differed significantly between lesion group I and lesion groups III and IV. A higher torque, as well as a higher velocity, contributes to a higher power output. The P30 and V30 differed significantly between the resistance groups. The highest resistance group showed both a higher P30 and V30.

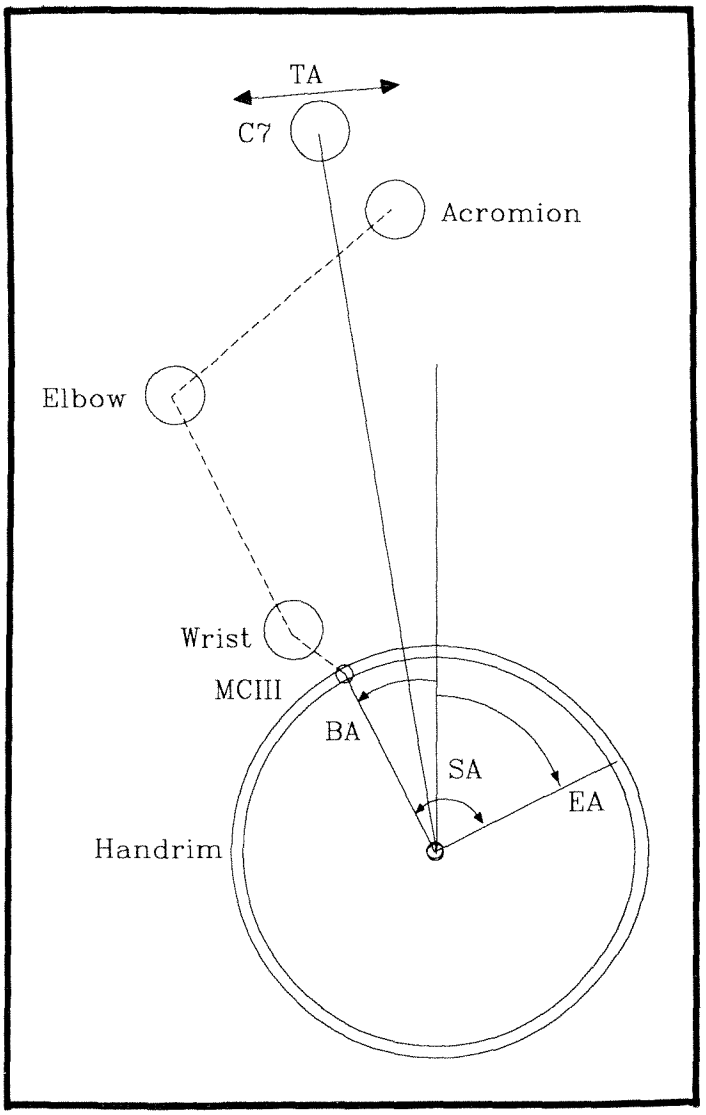

Figure 1.
Definition of the kinematic parameters; BA: begin angle, EA: end angle, MCIII: third metacarpal, SA: stroke angle, TA: trunk angle.

Figure 2 shows the power curves for three cycles with the highest peak power for two subjects with strongly differing power curves. Although large inter-individual differences existed in the shapes of the power curves, intra-individual patterns were found to be highly consistent.

Force Application

The results for the force analysis are shown in **Table 3**. Fx-cycle, Fz-cycle, and Ftot-cycle were found to be significantly lower for lesion group I when compared to the other lesion groups, except

Table 2.
Group means and standard deviations for mean and maximal power output, torque, and velocity for the right side only.

Group	I (n = 6) Mean (sd)	II (n = 5) Mean (sd)	III (n = 5) Mean (sd)	IV (n = 7) Mean (sd)	p-value
P30 (W)	21.5 (12.0)	46.9 (14.1)	63.7 (11.8)	49.1 (12.6)	***
Pmax (W)	121.8 (45.2)	248.7 (82.7)	398.1 (66.0)	310.3 (75.6)	***
M30 (Nm)	4.6 (2.2)	7.4 (1.6)	8.5 (1.4)	7.3 (1.8)	*
Mmax (Nm)	24.8 (6.6)	36.4 (9.3)	47.2 (5.5)	43.8 (11.6)	**
V30 ($m \cdot s^{-1}$)	1.45 (0.28)	1.95 (0.29)	2.35 (0.37)	2.15 (0.45)	**
Vmax ($m \cdot s^{-1}$)	1.68 (0.29)	2.37 (0.38)	2.93 (0.32)	2.70 (0.62)	**

*$p<0.05$;**$p<0.01$; ***$p<0.001$

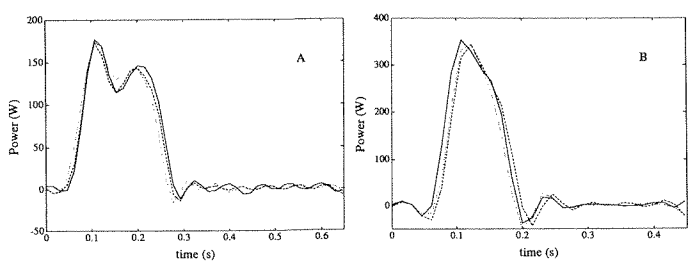

Figure 2.
Example of two typical forms of power curves for the three cycles with the highest peak power for two subjects.

for Fx-cycle of group III, which was relatively low. Apparently, subjects of lesion group III applied their force in a more downward direction onto the rim.

Lesion group I showed a significantly higher Fy-min, the inwardly directed force component. No differences between lesion groups were visible for the Fy-max.

FEF30 did not differ significantly between lesion groups (46.1 percent for group I versus 53.9, 57.2, and 54.6 percent for groups II, III, and IV).

In **Figure 3**, the force curves of two different cycles are shown for a subject with a cervical lesion and a subject with a low thoracic lesion. As was found for the power curves, the force curves also showed large inter-individual differences in the shapes of the curves, whereas intra-individual patterns were found to be highly consistent.

Kinematics

The kinematic parameters are listed in **Table 4**. Video data were available for 21 subjects. Although the stroke angle (SA) found for lesion group I appeared to be larger than for the other three lesion groups (90.4° for group I versus 78.4, 65.5, and 73.0° for groups II, III, and IV), no statistical differences between the lesion groups were found for begin angle (BA), stroke angle, or end angle (EA). Stroke angle ranged from 36.6° for a subject in group IV to 108.5° for a subject in group I. No significant correlations with velocity were found for BA, SA, or EA.

Cycle time (CT) and push time (PT) were significantly larger for lesion group I (0.85 sec), when compared to the other lesion groups (approximately 0.50 sec). Hence, when expressed as a percentage of the CT, PT was equal for all groups and varied between 45 and 55 percent.

A significant correlation was found for CT ($r = 0.62$) and PT ($r = 0.69$) compared with velocity.

The hand trajectory in the recovery phase of the cycle was very consistent within all subjects; in each cycle the same movement pattern recurred (**Figure 4**).

Since trunk angle (TA) data were only available for 16 subjects, the group size was too small to allow proper statistical analyses between groups. Nevertheless, TA was found to be very small for all subjects (mean: 6.7° and sd: 3.3°), ranging from 3.2 to 17.5°.

Table 3.
Group means and standard deviations for forces applied on the rim and fraction of effective force, for the right side only.

Group	I (n = 6) Mean (sd)	II (n = 5) Mean (sd)	III (n = 5) Mean (sd)	IV (n = 7) Mean (sd)	p-value
Fx-cycle (N)	16.5 (3.5)	26.6 (4.6)	20.1 (6.9)	25.5 (6.4)	*
Fz-cycle (N)	24.8 (9.9)	37.4 (8.8)	41.5 (5.9)	36.6 (10.4)	*
Ftot-cycle (N)	38.1 (8.6)	54.4 (11.3)	57.1 (4.7)	56.3 (9.8)	**
Fy-max (%)	5.4 (7.3)	3.5 (1.8)	6.2 (5.7)	7.5 (6.1)	ns
Fy-min (%)	−19.2 (17.2)	−7.3 (2.3)	−6.5 (6.9)	−3.1 (1.6)	*
FEF30 (%)	46.1 (10.9)	53.9 (8.2)	57.2 (3.6)	54.6 (9.4)	ns

*p<0.05; **p<0.01; ns: not significant

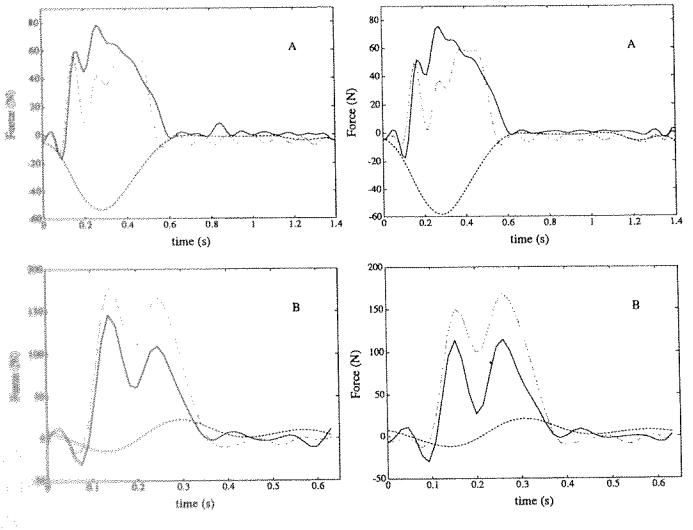

Figure 3.
Force curves of Fx (solid line), Fy (dashed), and Fz (dotted) for two different cycles. A: subject with a cervical lesion; B: subject with a low thoracic lesion.

DISCUSSION

Protocol

The power output produced during a 30-second test can be influenced by the experimental protocol. Two important variables that influence the performance are the dimensions of the wheelchair ergometer (11) and the magnitude of the applied resistance (8,12). In the present study, the dimensions of the wheelchair ergometer were standardized for all subjects. However, the fact that the subjects did not use their own (and presumably better-fitting) wheelchairs, could of course have influenced their maximum power output. Concerning the magnitude of the resistance, previous studies have shown that an increase in resistance up to very high values will result in an increase in mean external power output (8,12). Despite the effect of resistance on power output, it was, however, unavoidable to classify subjects with a spinal cord lesion in different resistance groups. These different resistance groups were necessary to avoid very high propulsion speeds. Too little resistance can result in a less than maximal power output because of coordinative problems that occur at high tangential rim velocities (12). The increase of velocity with increasing resistance in the present study justifies the choice for a variable resistance. However, some subjects who were able to deliver a high power output, still reached propulsion velocities above the set limit of 3 $m \cdot s^{-1}$, which may have limited their anaerobic power output. Therefore, it seems sensible to extend the classification in three different resistance groups of 0.25, 0.50, and 0.75 $N \cdot kg^{-1}$ body mass with an additional resistance group of 1.00 $N \cdot kg^{-1}$. Resistance should be individually applied according to predetermined relationships with lesion level, age, and sport activity.

Power Output

Several situations in the daily lives of wheelchair dependent persons seem to draw upon anaerobic power output (6). Mean power output is

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Table 4.
Group means and standard deviations of begin-angle, end-angle, stroke-angle, cycle-time and push-time as a percentage of the cycle-time.

Group	I (n = 6) Mean (sd)	II (n = 5) Mean (sd)	III (n = 4) Mean (sd)	IV (n = 6) Mean (sd)	p-value
BA (deg)	−28.6 (9.0)	−26.7 (6.5)	−14.7 (12.1)	−21.8 (15.7)	ns
EA (deg)	61.8 (10.4)	52.6 (11.1)	50.8 (11.5)	51.2 (14.3)	ns
SA (deg)	90.4 (13.1)	78.4 (7.8)	65.5 (20.6)	73.0 (23.2)	ns
CT (sec)	0.85 (0.30)	0.56 (0.05)	0.49 (0.08)	0.51 (0.10)	**
PT/CT (%)	46.8 (6.0)	54.2 (9.5)	45.2 (3.4)	47.6 (4.1)	ns

**$p < 0.01$; ns: not significant

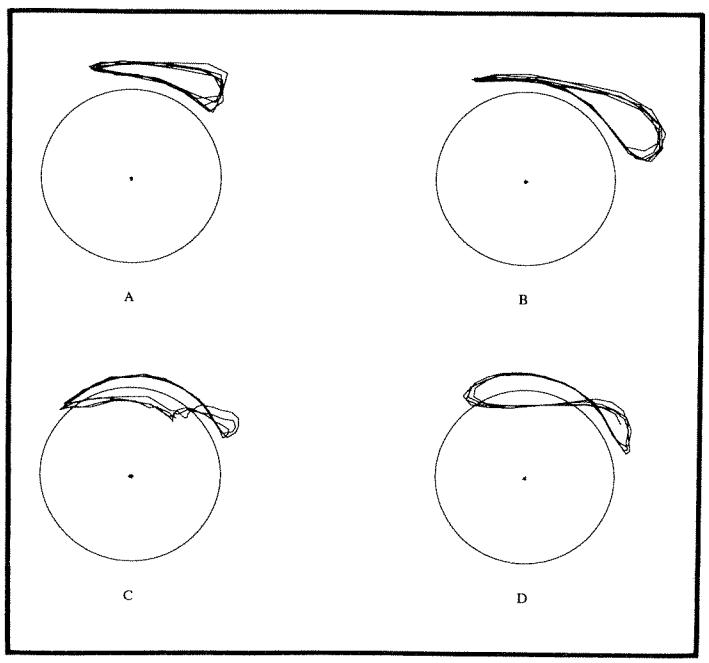

Figure 4.
Movement pattern of the hand. A: subject with a high thoracic lesion; B: subject with a low thoracic lesion; C and D: subjects with a cervical lesion.

assumed to be an indicator for anaerobic performance of the subjects. The anaerobic power delivered by subjects with a cervical spinal cord lesion was found to be very low; on average no more than 21.5 Watt (one-sided), which is comparable to propelling a wheelchair against a 1.5° slope at a propulsion speed of 4 $km \cdot hr^{-1}$. The P30 of subjects with a cervical lesion was, as expected, significantly lower compared to subjects with a thoracic or lumbar lesion; whereas, no differences were found between the latter groups (46.9, 63.7, and 49.1 Watt, respectively). The wide range for mean power reflects the large variety in performance. The relatively low power output found for subjects with a cervical lesion illustrates the low physical performance capacity of this group. This finding underlines the conclusion by Janssen, et al. (6), that persons with a cervical lesion have higher risks for overload situations in daily life. The large diversity in capacity of the SCI population should therefore be taken into account with respect to guidelines and requirements for the environmental space of the SCI population. For wheelchair users with a thoracic or lumbar lesion, anaerobic power output was found to be less strongly related to lesion level. Apparently, other factors besides lesion level seem to have a major effect on the short-term performance of those groups.

Force Application

In theory, the most effective direction of the force applied by the arms will be tangent to the rim. Both elbow flexors and elbow extensors will be needed for an effective force direction (12). In general, the fraction of effective force found in this study agrees with findings of Veeger et al. (13) for able-bodied subjects under similar conditions. However, based on the limited arm functionality of some of the subjects in lesion group I, a significantly lower fraction of effective force was expected for this group. Unexpectedly, no significant differences were found between the fraction of effective force

values for lesion group I and the other lesion groups (46.1 percent for group I versus 53.9, 57.2, and 54.6 percent for groups II, III, and IV). An explanation for this finding might be found in the large differences in rim velocity between the subjects. The higher velocities, reached by well-trained subjects with a low lesion level, could have caused coordination problems for the arms. In previous research, Veeger, et al. (14) found a decrease of the fraction of effective force while increasing speed. Since the velocity reached by subjects in lesion group I was much lower than in the other groups, arm functionality and rim velocity may have compensated each other with respect to the effectiveness of the force. The fraction of effective force in lesion group I was probably lower due to a reduced arm functionality; whereas, the fraction of effective force of the lower lesion groups seems to be limited by the higher velocity. The effect of a reduced arm function on the force direction may well be shown by the force analysis of this study; a significantly higher inward directed force (Fy-min) was found for lesion group I (see **Figure 3**). Since Fy is applied perpendicular to the direction of propulsion, Fy-max and Fy-min are ineffective force components in terms of propulsion. The higher Fy-min could be the consequence of a reduced triceps function in subjects with a cervical lesion. However, a higher Fy-min could also be associated with the need for additional hand-rim friction, caused by the lack of grasping ability in subjects with a cervical lesion. This friction can hardly be provided in a downward direction (due to a limited triceps function), which leaves Fy-min as a viable alternative.

The observation that the fraction of effective force is low, should not be used as an argument for specific training in the direction of a more effective force application. It is highly possible that the mechanically ineffective force direction is in fact the most efficient solution for the application of a propulsion force by the human 'motor,' given the limitations of the system. Enhancing the effectiveness of force direction should therefore rather be sought in an improved adjustment of propulsion system and human 'motor' to each other. It is possible that this might lead to an increase in effectiveness, in conjunction with an increase in physiological performance.

Kinematics

As a result of a reduced triceps function, subjects with a complete cervical lesion are not able to make an (active) extension in the elbow. Therefore, they tend to make a 'pull movement' with the arms on the rims, which is initiated in the shoulders. This 'pull movement' is in contrast to the push movement on the rims as is shown by subjects with a thoracic or lumbar lesion.

This raised the expectation that the begin angle of lesion group I would be larger; more specifically, that the hands would be placed further behind Top-Dead-Center. However, no significant differences were found for begin angle. The results are supposed to be influenced by the fact that not all subjects in lesion group I had a reduced triceps function. Lack of significant differences between the groups could also be caused by the small group sizes. Future research might show a larger stroke angle for subjects with a cervical lesion, when larger experimental groups will be investigated and a distinction in triceps function is made.

On the basis of this study, it can be concluded that the averaged trunk movement (TA) was small for all subjects. This is in agreement with previous research (15–17). However, the trunk was found to be more vertical than under comparable conditions in other studies (16,17). This may have been the consequence of the stationary test situation and the height of the ergometer. The high back rest could also have limited the rearward trunk movement of the subjects. Brubaker (18) previously showed that trunk angle can be affected by the height of the back rest.

Propulsion technique appeared to be intra-individually consistent. This is reflected in the consistency of the force curves, power output curves, and in the movement patterns of the hand (see **Figures 2, 3,** and **4**).

The intra-individual consistency is remarkable in the recovery phase, since the arms are free to choose one out of many possibilities to return to the handrims. Sanderson and Sommer (19) had similar findings after a kinematic analysis of athletes with a paraplegic lesion in a submaximal test. Subjects with a thoracic or lumbar lesion showed a large inter-individual variation in propulsion technique. Besides the assumption that propulsion technique will be

affected by differences in lesion level, an adaptation in propulsion technique can be expected when the velocity conditions are changed. This conforms with the observations in earlier studies by Lees (15), Veeger, et al. (17), and Coutts (20).

CONCLUSION

Subjects with a cervical spinal cord lesion had a low anaerobic power output, that was significantly lower than for subjects with a thoracic or lumbar lesion. No differences were found between the groups of SCI subjects with a thoracic or lumbar lesion.

The kinematics of subjects with a cervical lesion differed strongly from those of subjects with a lower lesion. Surprisingly, this was not reflected in the stroke parameters or the fraction of effective force. To what extent power output is influenced by a limited arm functionality, and to what extent power output can be enhanced by changes in wheelchair dimensions, should be subjects of further research.

REFERENCES

1. Woude van der LHV. Manual wheelchair propulsion: an ergonomic perspective (PhD thesis). Amsterdam, The Netherlands: Free University Press, 1989.
2. Hoffman MD. Cardiorespiratory fitness and training in quadriplegics and paraplegics. Sports Med 1986:3:312-8.
3. Hjeltnes N. Physical exercise and physiological exercise testing in patients with spinal cord injuries—a short review. Scan J Sports Sci 1988:10:55-64.
4. Burkett LN, Chisum J, Stone W, Fernhall B. Exercise capacity of untrained spinal cord injured individuals and the relationship of peak oxygen uptake to level of injury. Paraplegia 1990:28:512-21.
5. Veeger HEJ, Hadj Yahmed M, Van der Woude LHV, Charpentier P. Peak oxygen uptake and maximal power output of Olympic wheelchair dependent athletes. Med Sci Sports Exerc 1991:23(10):1201-9.
6. Janssen TWJ, van Oers AJM, Hollander AP, van der Woude LHV, Rozendal RH. Physical strain during activities of daily living in spinal cord injured subjects. In: Proceedings of the International Workshop on Wheelchair Research: Ergonomics of manual wheelchair propulsion, state of the art. Amsterdam: COMAC-BME, 1991:113.
7. Coutts KD, Stogryn JL. Aerobic and anaerobic power of Canadian wheelchair track athletes. Med Sci Sports Exerc 1987:19(1):62-3.
8. Lees A, Arthur S. An investigation into anaerobic performances of wheelchair athletes. Ergonomics 1988:31(11):1529-37.
9. Niesing RF, Eijskoot F, Kranse R, et al. Computer controlled wheelchair ergometer. Med Biol Eng Comput 1990:28:329-38.
10. Meys PJM, van Oers CAJM, Veeger HEJ, van der Woude LHV. The effect of seat height on the physiological response and propulsion technique in wheelchair propulsion. J Rehabil Sci 1989:2(4):104-13.
11. Woude van der LHV, Veeger HEJ, Rozendal RH. Propulsion technique in handrim wheelchair ambulation. J Med Eng Technol 1989:13(1/2):136-41.
12. Veeger HEJ, van der Woude LHV, Rozendal RH. Within-cycle characteristics of the wheelchair push in sprinting on a wheelchair ergometer. Med Sci Sports Exerc 1991:23(2):264-71.
13. Veeger HEJ, Lute EMC, Roeleveld K, van der Woude LHV. Differences in performance between trained and untrained subjects during a 30-s sprint test on a wheelchair ergometer. Eur J Appl Physiol 1992:64:158-66.
14. Veeger HEJ, van der Woude LHV, Rozendal RH. Effect of handrim velocity on mechanical efficiency in wheelchair propulsion against constant power output. Med Sci Sports Exerc 1992:24(1):100-7.
15. Lees A. The effect of speed and load on the pushing technique used by racing wheelchair athletes (Abstract). In: Proceedings of XIth International Congress on Biomechanics. Amsterdam: 1987:187-91.
16. Ridgeway M, Pope C, Wilkerson J. A kinematic analysis of 800-Meter wheelchair-racing techniques. Adapt Phys Act Q 1988:5:96-111.
17. Veeger HEJ, van der Woude LHV, Rozendal RH. Wheelchair propulsion techniques at different speeds. Scand J Rehabil Med 1989:21:197-206.
18. Brubaker CE. Wheelchair prescription: an analysis of factors that effect mobility and performance. J Rehabil Res Dev 1986:23(4):19-26.
19. Sanderson DJ, Sommer HJ. Kinematic features of wheelchair propulsion. J Biomech 1985:12(6):423-6.
20. Coutts KD. Kinematics of sport wheelchair propulsion. J Rehabil Res Dev 1990:27(1):21-6.