BIOMECHANICS OF WHEELCHAIR RACING

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INTRODUCTION

Performance capacity in wheelchair racing depends upon overall work capacity, propulsion technique, functionality of the trunk and the hand-am-shoulder mechanism, the fitting of the wheelchair to the athlete and finally the vehicle mechanics of the wheelchair design.

Obviously, an additional assistive device - such as a wheelchair or a prosthesis forms an integral and functional part of the athlete and as such is a prerequisite for mobility of lower limb disabled persons. The use of such a tool in high performance sports situations however, sets typical demands to the performance capacity of the individual athlete. The study of the wheelchair-user coinbination sets typical demands to the methods and techniques of biomechanical research. Apart from environmental factors optimization of performance in wheelchair racing must focus upon three areas of interest: (1) the mechanics of the wheelchair, (2) the performance capacity of the athlete, including propulsion technique and finally (3) the interfacing of the wheelchair-user cornbination (Woude, 1989).

Firstly, wheelchairs should obey to the laws of (vehicle) mechanics. Obviously, rolling resistance, air drag and internal friction must be minimized (Hedrick et al., 1990; Frank & Abel, 1991), since these forces detennine the external power output for the human engine. The contemporary high performance racing wheelchair is a light ($\pm$7.5kg) modular wheelchair, non-foldable, equipped with quick release rear axles, for/aft adjustment, fixed bucket/cage seat, rear wheel camber up to 15°, a non-adjustable front wheel and high pressure tubes (Davis, 1992). 'Lightweight' and 'high tech' are keywords, the design is task-specific and tuned to the racer. In all, track wheelchairs conform to the intrinsic requirement of a low external power output.

Secondly, understanding the functioning of the human engine in wheelchair racing requires a careful combined biomechanical and physiological approach of wheelchair arm work under standardized experimental conditions. In recent years an increasing number of studies focused upon the biomechanics of wheelchair track performance (Lees & Arthur, 1988; Coutts, 1990; Davis et al., 1990; Cooper, 1990a,b; Bakker et al., 1992). The majority of studies concentrated on kinematic aspects (Sanderson & Sommer, 1985; Woude et al., 1988; Brown et al., 1990; Kobayashi et al. 1991). The complexity and costs of wheelchair ergometry within a biodynamic perspective probably has limited the number of research efforts dedicated to a more thorough exploration of the biomechanics of wheelchair racing. However, the biodynamics of manual wheelchair propulsion in general was studied in a number of recent publications (Lesser, 1986; Traut, 1989; McLaurin & Brubaker, 1991; Haghpanahi et al., 1991; Woude et al., 1990a, 1991; Veeger et al., 1991a,b; 1992a,b).

Thirdly, the wheelchair must be tuned to the individual physical demands. A careful analysis of human power production and propulsion technique with respect to design features of the interface will serve the development of proper guidelines to help the fitting of the wheelchair to the functional characteristics of the user (Woude, 1989).

Next to performance. some authors stressed the health impact of (high performance) wheelchair use upon the musculo-skeletal system (Haghpanahi et al., 1991). Repetitive strain injuries (RSI) in the upper limb (bursitis, tendinitis to hand and shoulder) have been reported as a, serious problem in wheelchair athletes (McCormack et al., 1991) and in relation to (long term) daily wheelchair use (Nichols et al., 1979). Apart from the cardio-vascular risks of a sitting and inactive life-style (LaPorte et al., 1984), skin tissue problems may emerge as a consequence of pressure concentrations, temperature, local humidity and the absence of intermittent pressure relief (Barbenel, 1991). Unknown - as far as we know - are however the effects on skin tissue of physical activity itself, extreme sitting postures, 'whole body vibrations', high propulsion forces and the long term use of the specialized bucket/cage seats in racing wheelchairs. Although lumbar disc pressure was studied in relation to standardized daily wheelchair activities (Andersson & Örtengren, 1974), the risks of racing with respect to the mechanical condition of the spine in lower limb disabled are unknown as well.

METHODOLOGY

Biomechanics may thus contribute in various ways to the problems of wheelchair mobility. Wheelchair propulsion and (track) performance can adequately be studied in lab-experiments on a *motor driven treadmill* (Sanderson & Sommer, 1985; Woude et al., 1988; Veeger et al., 1991d). Measurement of external power output is possible through a standardized drag test (Woude et al., 1988). Three-dimensional kinematics and electromyography were studied in conjunction with overall physiology and proved to lead to a valuable description of some biomechanical aspects of wheelchair mobility. To enable a more detailed biodynamic analysis of wheelchair propulsion specialized egorneters have been designed (Lesser, 1986; Traut, 1989; Niessing et al. 1990) as well as instrumented wheelchairs (Haghpanahi et al., 1991; Kobayashi et al., 1991). A *computer-controlled wheelchair ergometer* was designed (Niessing et al., 1990; Woude et al. 1990a; Veeger et al. 1992c) to study torque and force generation in all interfacing units of the wheelchair-user combination: seat, backrest and rims. The ergometer allows simulation of a variety of wheelchair dimensions (geometry, rim size/form, camber angle) and wheeling conditions, whereas three-dimensional motion analysis, force and power analysis, electromyography and cardio-respiratory parameters can be combined. Through modelling, the net torques and net power production around the arm joints and in-depth study of efficacy of force generation under (sub-)maximal performance conditions is possible. A comparison between wheelchair propulsion on the treadmill and the ergometer revealed no significant differences in kinematics or in overall physiology data (with the exception of the trunk angle; Veeger et al., 1992c).

Preferably, (track) performance of wheelchair athletes and wheelchair confined subjects is evaluated in standardized lab-experiments: an aerobic ([supra-] maximal) exercise test and a 30-sec sprint test, indicative for endurance capacity and short term power output, respectively.

THE HUMAN ENGINE

The *athlete* is the motor of the wheelchair-user combination. The cardio-respiratory work capacity and propulsion technique eventually determine the performance of the motor. Wheelchair arm work is an inefficient form of locomotion and gross mechanical efficiency in hand rim wheelchairs hardly ever exceeds 10% (Woude et al., 1988). The combined study of functional anatomy, biomechanics and physiology will help explain these low levels of efficiency. In general, mobility in hand rim propulsion is limited as a consequence of the small muscle mass involved, the complex push phase in a discontinuous movement pattern of the hand-arm-shoulder system and the complex 'joint' structure of the *shoulder complex*, which makes it highly versatile, but is assumed to require extensive additional (static) muscular activity. Obviously the functional status of the athlete is a major overall determinant in arm work. To study the 'absolute' limits of wheelchair hand rim propulsion with respect to disability a group of wheelchair athletes was studied during the World Games and Championships for the Disabled (Assen, 1990). Eventually 68 athletes participated voluntarily in a sprint and aerobic maximum exercise test on the wheelchair ergometer. Applied hand-to-rim forces, torque and power production were thus evaluated for a group of athletes varying in age, sex, functional status and - to some degree - sports discipline. Preliminary results indicate a considerable variation in maximum power production as well as in technique parameters between groups of subjects with different disabilities and/or functional status (Bakker et al., 1992). Variation in aerobic capacity indeed is strongly associated with functional status and sports discipline, as has been described in different previous studies (Veeger et al. 1991d). The variation in more technique oriented parameters has not been described to a large extent however.

As is shown in Figure 1. the mean *(maximum) power output* in the last step of the maximum exercise test remained well below 150W whereas the mean power production in the sprint test (P_{30}: power over 30 seconds) did not exceed 200W. Due to the discontinuous character of wheelchair propulsion, peak power within the push phase reached values of over 1100W in the *sprint* test. Power output as measured on the left and right hand rim showed a high correlation (R=0.97), especially when hemiplegic subjects (n=2) were excluded. The extremely low values for power output in Figure 1. were seen in subjects with cerebral palsy (CP) or with a cervical spinal cord injury (ISMGF Functional classification code: T1). The maximum exercise test and the sprint test are seen as indicators for endurance capacity and short term anaerobic power output respectively. The majority of track athletes participated in both the short as well as the long distance races and the test results indicate a strong association between power output in the sprint and maximum exercise test (R=0.85). This may imply that there is not a very high degree of specificity among the wheelchair athletes with respect to track disciplines. However, correlations between the 200m sprint time and marathon time of 22 track athletes and their sprint test power output showed values of R=-0.79 and -0.43 respectively. Track results here are however confounded by major differences in wheelchair quality! Also, among the track athletes classified according to the new ISMGF system (T1-T4) a high degree of accordance is seen between test outcome and functional classification.

As is to be expected, within an individual exercise test high correlations are found between mean and peak hand forces F_x (for-aft), F_z (vertical) and torque M and the mean and peak power (R=0.8-0.99) as is shown in Figure 2. for the maximum exercise test. An individual exception may be found for the medio-lateral acting hand force F_y, which generally shows a high level of variability. Within the group of 68 athletes high associations are seen between mean maximum power (final stage of the test) and force parameters as well. This may be related to the high interindividual variation with respect to functionality, which is reflected in both power and technique parameters. When looking into a smaller but much more homogeneous group of male athletes (T4; n=22) no correlations were seen between power output and force parameters. Other more subtle parameters must be responsible. Some of them will be discussed below.

In mechanical terms ideal force exertion upon the rim should be directed tangential to the rim. In 'the sprint test the mean Fraction Effective Force (FEF) was determined to evaluate whether experienced athletes tend to follow this simple mechanical principle (Figures 3. & 4.). FEF is the ratio between the tangential force applied upon the hand rim [torque/rim radius] during the push phase and the total force vector ($[F_{tot}=\sqrt{(F_x^2+F_y^2+F_z^2)}]$: $FEF= F_m * F_{tot}^{-1} * 100\%$). Figure 3. shows the FEF_{push} in combination with uni-lateral power output and indicates a large variability in FEF along with the power output. Values of little less than 100% are contrasted with values below 20 and even near zero, indicating no or little effective force production. It is important to see the evident relation between a technique parameter and power output. Increasing functional ability ($T1_{male}$ --> $T4_{male}$) indeed showed a consistent increase in FEF_{30} from 36 to 51% within the male subject group (Figure 4.). FEF_{30} ranged from 32% for a group of 4 athletes with cerebral palsy to a mean of 56% for a group of three female spinal cord injured subjects ($T3_{female}$). In the last step of the maximum exercise test FEF_{push} appeared consistently higher for the $T4_{male}$ group (N=23) and ranged between 56 and 80 % with an overall mean value of 69%. This difference in FEF between the tests will be related to differences in velocity which was on average 2.6m.s^{-1} in the sprint test and 1.31m.s^{-1} in the maximum exercise test for the $T4_{male}$

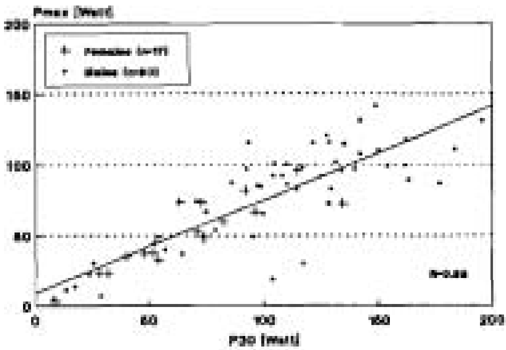

Fig. 1 Left and right

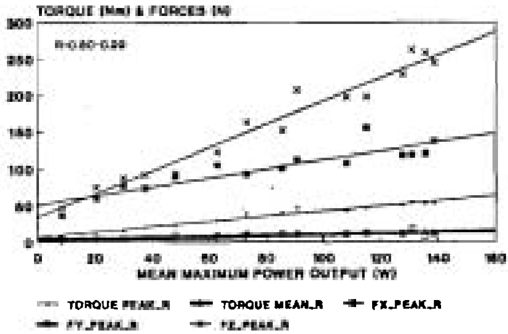

Fig. 2 SL20 max Lest

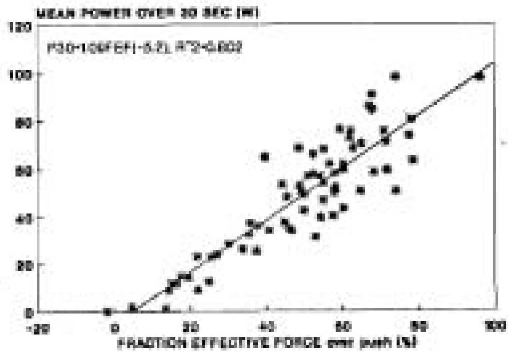

Fig. 3 N=67, 30 sec, right side

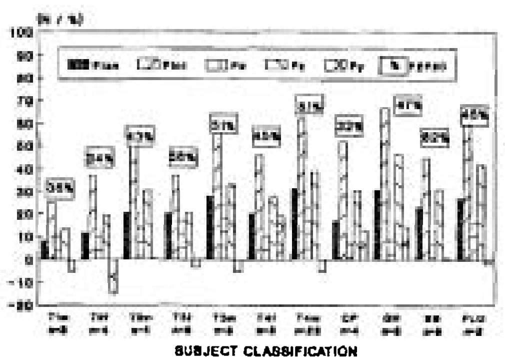

Fig. 4 Subject classification

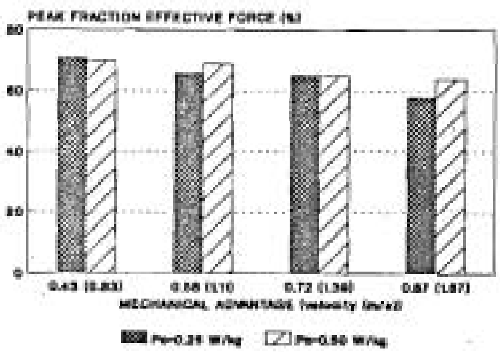

Fig. 5 N=9 non-wheelchair users

group (Woude et al. 1991; Bakker et al. 1992). A clear interindividual variation is also seen in the T4 group, which may be associated with wheeling expertise, training status and smaller variations in functionality. FEF increases with decreasing speed and power levels. It may -in part - explain the low levels of mechanical efficiency seen in hand rim wheelchair propulsion and racing, but on the other hand may be inevitable in maximum performance!

Other factors within in the push may be responsible for the ***low efficiency*** of wheeled mobility. The individual push in wheelchair propulsion requires a complex guided motion of the hand along the rim. The torque is characterized by a phasic pattern with a ***negative deflection*** at the start and end of the push phase and a dip in the midphase. The negative deflections reflect a braking torque leading to energy loss of the system. Results show that these dips increase with the speed of the rim -stressing the 'grab' problem in the push phase- both in sprint tests (Veeger et al. 1991a) and in submaximal wheeling (Veeger et al. 1992b). The fluency of the mid-push phase seems associated with the speed of motion as well and generally seems to be related to the transition of biceps to triceps activity, initiating a shift from flexion to extension.

The ergometer allows calculation of the effective or tangential force directly from both the torque applied to the rim and from the force transducers in the frame of the ergometer. In the latter case kinematic information of the orientation of the hand upon the rim and force data are combined. Although it evidently is rather complicated to determine a realistic point of force application in the hand, the results indicate that a (negative) ***misting torque*** M_h of the hand onto the rim surface, which is not measured in the force transducers, is included in the overall torque measured in the wheel axle. Especially in static force exertion, M_h may contribute positively to the torque, whereas in dynamic experiments it will usually exert a braking force which for instance varied up to -9.4Nm in a series of sprint tests of a group of experienced wheelchair users (Veeger et al. 1992a). Clearly, this will contribute to the 'inefficacy' of propulsion. On the other hand it may well be an inevitable aspect of the complex interfacing between the hand and the rim, where a sufficiently large friction force between hand and rim surface is clearly required to transfer large propulsion forces, especially in high performance track events.

The interfacing between the user and the wheelchair surely seems 'to be of influence on performance as well. Detailed analysis of both biomechanics and physiology may lead to a proper theoretical framework in this respect and to improved design and fitting criteria. Within. that framework special attention must he dedicated to the functional anatomy and biomechanics of the shoulder mechanism. This versatile interface between trunk and arm is crucial in power production. Detailed dissection studies (Helm et al. 1992; Veeger et al. 1991c) will firstly help clarify the role of the different components with respect to propulsion technique in wheelchair propulsion, will secondly clarify differences in performance between groups of disabled and will help determine wheelchair fitting guidelines. This integrated approach also may help explain the human potential and limitations in arm work, and the mechanisms leading to overuse injuries to the arm-shoulder complex.

WHEELCHAIR-USER INTERFACE

The design of the propulsion mechanism (rim form, size) and the fitting of the wheelchair to the athlete - the wheelchair-user interface - is expected to be of influence on performance (Woude, 1989). Veeger et al. (1989) however showed that the effect on performance of an increasing ***rear wheel*** *camber* angle from 0 to 9" - during wheelchair propulsion on a motor driven treadmill in a basketball wheelchair - was negligible with respect to efficiency and energy cost. A small decrease in the maximum abduction angle during the push phase appeared significantly related to camber angle. Electromyographic tracings of the medial head of the deltoid however revealed an unexpected absence of activity during the greater part of the push phase and an absence of any substantial variation between different camber angles. This suggests that variation in abduction angle during the push phase is not an active process but a forced motion as a mere consequence of the closed chain between hand rim, hand, shoulder complex and trunk and the anteflexion torque of the greater pectoral muscles. The latter muscle will additionally force the upper arm to endorotate and adduct. These results however not necessarily apply to racing wheelchairs, where the sitting posture, the rim size/form, but also power requirements and propulsion technique are quite more extreme. Camber itself obviously has practical advantages: an improved 'reach' of the hands to the rims, improved lateral stability especially in racing, a decreased downward turning moment on lateral slopes. Recent results on the effect of an absolute shift in *width* of the rims upon technique and power production in a group of 8 male spinal cord injured subjects showed no significant effects in the torque, total applied force and FEF or cardio-respiratory parameters. Within the ranges studied the arm-shoulder system seems to adapt effectively to the task. Under more extreme conditions the tuning may be more critical.

Rim and tube diameter varied with the years in wheelchair racing. Woude et al. (1988) stressed the significance of rim diameter (0.3 - 0.56m in diameter; mechanical advantage: 0.43-0.77) for cardio-respiratory responses during submaximal performance of skilled athletes on a motor driven treadmill during steady state propulsion ($v=0.83$ - $4.17 m.s^{-1}$; mean $P=$ 8.5 -37W) in a racing wheelchair. The smallest rim appeared the most efficient (+2%) and less straining (heart rate: -10/20b.min^{-1}). Kinematic analysis showed no significant shifts in timing and angular hand displacement. Linear speed of the hand increased significantly with rim diameter of course, while simultaneously the shoulder joint showed a significantly increased excursion during the push phase. Angular velocity for elbow, shoulder and trunk excursions increased with rim size. Rim size seems to affect the human engine through the force/velocity ratio (rim size/linear rim velocity) and length condition of active muscle groups, shown in shifts in segmental de-/accelerations (shoulder-rim distance).

The role of linear rim velocity in the complicated -guided- hand-arm motion during hand rim propulsion was subsequently studied with respect to *mechanical advantage* (MA: output force/input (hand) force). Veeger et al. (1991b) showed the effect of linear rim velocity upon both physiological and biomechanical measures during submaximal propulsion at constant levels of power output (0.25 and 0.5 $W.kg^{-1}$; $V=0.83$ -$1.67 m.s^{-1}$; MA=0.43-0.87) on the ergometer. Significantly increasing values for heart rate and oxygen uptake, and a decreasing mechanical efficiency with increasing rim velocity, but constant power output were accompanied by significantly increasing values for peak power production in the push phase, a constant push angle and peak torque value, a decreasing push time and a decreased fraction effective force FEF (peak and average). FEF_{peak} dropped from 71 to 58% for an increasing mechanical advantage ($P=0.25 W.kg^{-1}$) and thus increasing linear rim velocity (Figure 5.). The negative deflections in the individual power curves during the push phase increased simultaneously. The increase in energy cost with increasing MA seems to be related to the decrement in FEF.

Scar height will not only influence air drag at high velocities, but also may influence the functioning of the hand-arm system. Results of two subsequent experiments (Woude et al., 1990b) on the treadmill and ergometer showed a significant effect of seat height (20 to 110" elbow angle in standardized sitting posture, hands on top of rim; full extension=0°), indicating an optimum seat height of 60-80° elbow angle during submaximal wheelchair propulsion in a basketball wheelchair. Push time, push range and angular hand velocity decreased with increasing seat height, whereas simple kinematics indicated a decrease in abduction, flexion/extension of the upper ann, flexion/extension in the elbow and an increase in trunk flexion with increasing shoulder-rim distance. No shifts in peak angular velocities were seen. Qualitative analyses of patterns of muscle activation indicated their role in accommodating to different requirements in joint torques and the varying kinematic conditions. Power and torque measures (N=5) in the submaximal tests on the first-stage design of the ergometer did not indicate significant differences at seat heights of 90, 100 and 110" elbow angle. Recent results of an absolute variation in the *for/aft position* in a group of 8 male spinal cord injured subjects revealed no significant effects upon total force, torque and fraction effective force FEF or the cardio-respiratory parameters under the submaximal conditions studied.

CONCLUSIONS

It may be concluded that combined bioniechanical and physiological research of wheelchair racing and wheelchair propulsion is of utmost importance to develop a thorough theoretical framework of arm work. This can lead to the iinprovement of wheelchair sports performance and daily wheelchair ambulation as a consequence of a better understanding of the human engine and its interfacing with the wheelchair. Moreover, such a knowledge base will lead to a better understanding of the hand-arm-shoulder system in general, but also with respect to functional disabilities ànd in conjunction with overuse injuries to the musculo-skeletal system.

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