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Seat height in handrim wheelchair propulsion

Dr. Luc H.V. van der Woude; Dr. Dirk-Jan Veeger; Prof. Dr. Rients H. Rozendal; Prof. Dr. Tony J. Sargeant
Faculty of Human Movement Sciences, Free University, Van der Boechorststraat 9, 1081BT Amsterdam, The Netherlands

Abstract—To study the effect of seat height on the cardiorespiratory system and kinematics in handrim wheelchair ambulation, nine non-wheelchair users participated in a wheelchair exercise experiment on a motor-driven treadmill. The subjects conducted five progressive exercise tests. After an initial try-out test, four tests were performed at different standardized seat heights of 100, 120, 140, and 160 degrees elbow extension (subject sitting erect, hands on the rim in top-dead-center = 12.00 hrs; full extension = 180 degrees). Each test consisted of four 3-minute exercise blocks at speeds of respectively 0.55, 0.83, 1.11, and 1.39 $m.s^{-1}$ (2-5 $km.hr^{-1}$).

Analysis of variance revealed significant effects of seat height ($P<0.05$) on gross mechanical efficiency (ME), oxygen cost, push range, and push duration, and on the ranges of motion in the different arm segments and trunk. Mean ME appeared higher at the lower seat heights of 100 and 120 degrees elbow extension. This is reflected in an enhanced oxygen consumption at seat heights of 140 and 160 degrees elbow extension. Simultaneously, the push range showed a 15 to 20 degree decrease with increasing seat height, which is reflected in a decreased push duration. In the push phase, decreases in retroflexion and abduction/adduction of the upper arm were seen. The trunk shifted further forward, and the motion range in the elbow joint shifted to extension with increasing seat height. No shifts in minimum and maximum angular velocities were seen with increasing seat height.

The results showed an interrelationship between wheelchair seat height and both cardiorespiratory and kinematic parameters. With respect to the cardiorespiratory system, the optimization of the wheelchair geometry, based on functional characteristics of the user, appears beneficial.

Key words: *cardiorespiratory system, wheelchair ambulation kinematics, wheelchair propulsion, wheelchair seat height.*

INTRODUCTION

Conventional handrim wheelchairs do not meet essential ergonomic requirements in terms of seat comfort and locomotion. They impose relatively high loads on the cardiorespiratory system and are remarkably inefficient in terms of energy cost of locomotion (2,4,5). A major problem in wheelchair locomotion is that optimum physical performance at the lowest energy cost can only be attained when the wheelchair-seat configuration and the propulsion mechanism comply in an optimum manner to the functional characteristics of the user.

Due to a growing interest in wheelchair sports, the development of sports wheelchairs led to task-specific devices, based on contemporary knowledge of product design and vehicle mechanics. Although in wheelchair sports, optimum fitting of the wheelchair to the physical characteristics of the user has been recognized as a prerequisite for success (13), no

Address all correspondence and requests for reprints to: Dr. Luc H.V. van der Woude, Department of Functional Anatomy, Faculty of Human Movement Sciences, Free University, Van der Boechorststraat 9, 1081BT Amsterdam, The Netherlands.

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experiment-based fitting criteria are available. Trial-and-error procedures are applied to fit the individual to the wheelchair in terms of fore/aft position, height, angulation of the seat, and considerations of overall vehicle mechanics.

Experiments have been performed that related design features of different (contemporary) wheelchairs to energy cost and kinematics. Handrim wheelchairs in general (24), and especially racing wheelchairs (26), appeared to be not highly efficient devices for locomotion. Veeger, *et al.* were unable to confirm the generally-accepted notion that increased rear-wheel camber will elicit lower cardiorespiratory responses (21).

Woude, *et al.* showed that handrim diameter is relevant in wheelchair racing in terms of energy cost, mechanical efficiency, and velocity requirements (26). However, it appeared that seat height might also be a relevant factor, which should be optimized in conjunction with the rim diameter. In terms of muscle functioning, optimization of the arm/shoulder complex and trunk with respect to the propulsion mechanism may alleviate fatigue, increase maximum torque and power output, and suppress cardiorespiratory strain. This notion is based primarily on practical experience; no experiment-based guidelines relating wheelchair-seat configuration to body dimensions are available.

In tool design, it has been widely accepted that optimum fitting of the dimensions of tools to the anthropometry of the user will prevent overuse injuries to the musculoskeletal system (7,10,20). In cycling, the optimum saddle height in terms of energy cost has been shown to be related to the length of the leg (19). Despite previous research on seat position in wheelchair propulsion, no useful relationship between body and wheelchair dimension(s) has been formulated (4,5,6,16,22). This may have been due to the absence of standardization of the height adjustment on the basis of body dimensions.

In the current study, results are presented of a combined physiological and kinesiological analysis of effects of four different seat heights, during handrim wheelchair propulsion, at four different velocities. Seat height was standardized between individuals with respect to arm and trunk lengths.

METHODS

Subjects

Nine able-bodied male subjects participated in this study on a voluntary basis, and gave written informed consent. Anthropometric data were determined according to the procedures described by Clauser, *et al.* (8), and Weiner and Lourie (23), and are summarized in **Table 1.**

Table 1.
Mean and SD of anthropometric and personal data of the subject group (N = 9).

	Age (y)	Body Weight (kg)	Body Length (m)	Shoulder Height (m)	Biacrom. Width (m)	Arm Length (m)	Arm Volume (l)	Arm Mass (kg)
1	22	66.4	1.78	0.625	0.411	0.779	2.58	2.82
2	21	68.5	1.81	0.624	0.386	0.831	2.98	3.48
3	23	78.4	1.79	0.624	0.419	0.78	3.61	3.66
4	27	69.3	1.85	0.605	0.328	0.81	3.03	3.36
5	21	79.1	1.85	0.637	0.383	0.848	3.34	3.83
6	24	80.7	1.85	0.625	0.423	0.827	3.39	3.94
7	24	85.6	1.88	0.649	0.395	0.805	3.49	3.79
8	28	84.0	1.80	0.623	0.396	0.765	3.72	4.05
9	23	82.5	1.83	0.64	0.436	0.821	3.59	3.85
Mean	22.7	82.7	1.83	0.628	0.397	0.807	3.30	3.64
SD	1.6	2.5	0.03	0.011	0.029	0.026	0.37	0.38

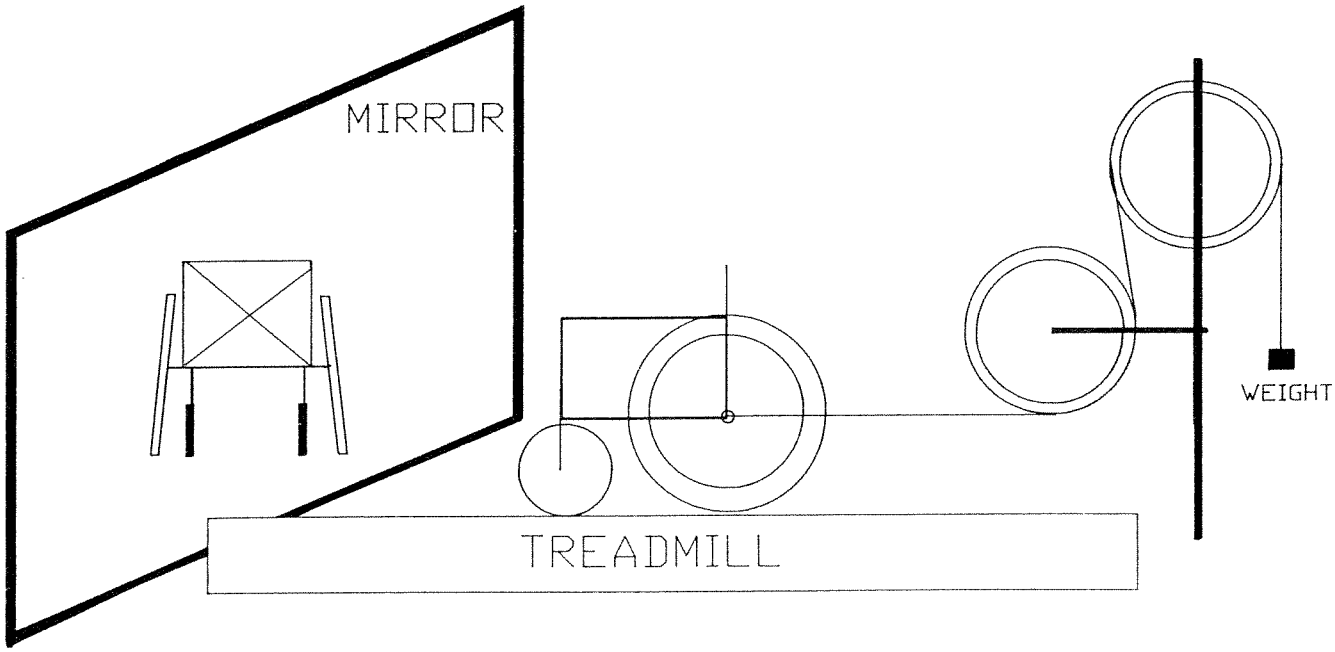

Figure 1.
Camera view of the sagittal and frontal plane of motion. The wheelchair on the treadmill encounters rolling resistance (F_r) and the force acting via the pulley system (F_{ad}).

Experimental procedure

All subjects propelled a solid-frame basketball wheelchair on a motor-driven treadmill. The wheelchair (Morien Tornado: weight—14.5 kg; hard castor wheels—0.08 m diameter; rear wheels—0.61 m diameter; tire pressure—3.10^5Pa) had an adapted wooden seat, which was continuously adjustable in height. The fore/aft position of the seat with respect to the wheels and camber angle (2 degrees) of the rear wheels were fixed. A standard handrim (0.52 m handrim diameter) was mounted to the wheels.

Each subject conducted five progressive wheelchair exercise tests. Each test was preceded by a 10-minute warm-up: the first test was a try-out test to get acquainted with the experimental set-up. The subsequent four exercise tests were conducted in random order at different seat heights. Seat height was adjusted with respect to the elbow angle (respectively 100, 120, 140 and 160 degrees elbow extension), and determined with the subject sitting immobile in a standardized posture with the hands on the rims in top-dead-center (12.00 hrs). For practical reasons, the elbow angle was defined as the internal angle between the upper and lower arm (full extension = 180 degrees).

Each test consisted of four subsequent 3-minute workloads. The treadmill velocity increased every three minutes with 0.28 $m.s^{-1}$ (1 $km.hr^{-1}$), starting at 0.55 $m.s^{-1}$ (2 $km.hr^{-1}$), and ranging up to 1.39 $m.^{-1}$ (5 $km.hr^{-1}$). Power output increased respectively from 0.17, 0.25, 0.33 to 0.44 $W.kgTW^{-1}$ (watts per unit weight of the subject + wheelchair). This load was imposed on the wheelchair-user combination through rolling drag (F_r), and an additional external force (F_{ad}) acting via a pulley system on the wheelchair (see **Figure 1**) (21). Mean external power output (P_o) was thus defined as the product of the sum of these drag forces ($F_r + F_{ad}$) and the mean belt speed (V). F_r for a given wheelchair-user combination was determined with a drag test, as previously described (2,24).

Since wheelchair ambulation under steady state conditions leads to a cyclic (repetitive) motion pattern, a power balance can be applied (14). Thus, from P_o and the cycle frequency (f), the amount of work/cycle (A) produced by the user can be determined, according to $A = P_o/f$(W).

Moreover, under submaximal conditions, gross mechanical efficiency (ME) can be calculated from the ratio between P_o and the rate of energy expenditure E_n, as derived from oxygen uptake: $ME = P_o/E_n \times 100\%$ (24).

Physiology

During each exercise test, expired gases were collected continuously with an OXYCON (Mijnhardt, Ox-4). The analyzers for oxygen and carbon dioxide were calibrated prior to each session with a known reference gas mixture. Minute ventilation ($l.min^{-1}$, BTPS), oxygen uptake ($l.min^{-1}$, STPD), carbon dioxide output ($l.min^{-1}$, STPD), and respiratory exchange ratio (RER) were determined every third minute of each workload, and used for further analysis (N = 9). Heart rate (HR) was monitored according to the method described by Woude, *et al.* (24).

Kinematics

During every third minute of each workload, subjects were filmed for a full cycle (60 $f.s^{-1}$). The movement pattern in the sagittal and frontal plane of the left upper arm and the trunk was recorded directly, and via a mirror with a high-speed camera (DBM-55, Teledyne; 16 mm Kodak 4XR-7277). With this synchronized sagittal and frontal view (**Figure 1**), a double 2D analysis was applied. Markers were placed on the anatomical landmarks C7, articulatio acromioclavicularis, epicondylus lateralis, articulatio manus, caput ossis metacarpalia III, the rear wheel axis, and a fixed point on the treadmill frame.

In accordance with previous descriptions (25), push time (duration of hand-to-rim contact, as derived from the film: PT), recovery time (termination of hand-to-rim contact to the next push phase: RT), cycle time (the sum of PT and RT and the inverse of the cycle frequency (*f*): CT), work/cycle (A) and push angle (the angular trajectory of the hand on the rim during the push phase: PA) were determined for each experimental condition (N = 9).

In order to obtain an impression of angular displacement and angular velocity of the upper and lower arm and trunk, film data were analyzed for the push phase of a complete cycle in each experimental condition (N = 4).

Marker positions in each frame were analyzed with a Summagraphics Supergrid digitizer (accuracy: 0.025 mm). Data points were smoothed with a digital low-pass filter (second-order recursive Butterworth filter, cut-off frequency 6 Hz) (15). Non-planar movements of the upper and lower arm were calculated from the mirror data in order to determine the actual elbow angle. Thus, angular displacement in the elbow joint was determined in terms of maximum ("extension") and minimum ("flexion") values. The minimum and maximum excursion of the upper arm in the frontal plane ("ad-/abduction") were determined from the mirror data with respect to a vertical reference line. The minimum ("retroflexion") and maximum angle of the upper arm ("anteflexion") in the sagittal plane were determined with respect to the trunk. Trunk excursion was derived from a line between C7 and a fixed point on the wheelchair seat (in a vertical position over the wheel axis). Thus, the maximum ("flexion") and minimum ("extension") angular displacement of the "trunk" during the push phase were determined with respect to a vertical reference line. Angular velocity of the described parameters was determined using a 5-point Lanczos differentiating filter (15).

Muscle activity of the left arm and trunk (mm. erector trunci pars lumbale, rectus abdominis, trapezius pars descendens, pectoralis major, deltoideus pars anterior, serratus anterior, latissimus dorsi, biceps brachii caput longum, and triceps brachii caput laterale) was studied qualitatively in the same four subjects, using surface electromyography (Sentry 1000 electrodes; DISA, type 15c01, band filter 10-500 Hz). Data were synchronized with the film data and expressed as a percentage of the maximum amplitude. A muscle was termed active when it exceeded 50 percent of the maximum amplitude.

Statistics

A three-factor analysis of variance with repeated measures on both "seat height" and "velocity" was conducted to the physiological parameters (N = 9). The parameters of propulsion technique and movement pattern were analyzed for N = 4 subjects. The level of significance was $P < 0.05$. To study the effects of anthropometric measures on oxygen cost, a multiple regression analysis was conducted (N = 9).

RESULTS

Physiology

The effects of seat height on the mean cardiorespiratory responses are shown in **Table 2** and **Figure 2.**

Table 2.
Results of analysis of variance with respect to main effects (seat height and velocity) and interaction.

Physiology		SEAT HEIGHT df (3,24)	VELOCITY df (3,24)	INTERACTION df (9,72)
Oxygen cost ($l.min^{-1}$)		3.04*	125.2**	1.6 ns
Ventilation ($l.min^{-1}$)		2.8 ns	36.5**	2.0*
Heart rate		1.7 ns	103.2**	1.3 ns
Gross ME		3.5*	18.3**	0.8 ns
RER		0.6 ns	3.8*	1.4 ns
Propulsion technique				
Cycle time		2.5 ns	50.9**	1.9 ns
Push time		5.8**	272.0**	2.1*
Push time (%CT)		2.6 ns	85.2**	0.9 ns
Recovery time		1.9 ns	10.6**	0.7 ns
Push angle		14.9**	4.4*	1.3 ns
Start angle		5.4**	15.9**	1.2 ns
End angle		8.1**	14.4**	1.4 ns
Work/cycle		2.7 ns	61.5**	1.2 ns
Kinematics push (N = 4)		**df (3,9)**	**df (3,9)**	**df (9,24)**
Trunk flexion	min	3.6 ns	6.0*	1.5 ns
	max	7.8**	16.3**	2.2 ns
Ang. velocity	min	1.1 ns	1.2 ns	0.9 ns
	max	0.6 ns	1.8 ns	0.8 ns
Anteflexion	min	120.0**	1.5 ns	1.2 ns
	max	0.4 ns	20.0**	0.7 ns
velocity	min	0.1 ns	4.3 ns	1.5 ns
	max	2.1 ns	8.7*	1.7 ns
Abduction	min	18.0**	6.1*	0.4 ns
	max	9.3**	2.4 ns	2.3*
velocity	min	0.8 ns	28.3**	1.5 ns
	max	1.1 ns	6.2*	1.1 ns
Elbow ext.	min	110.3**	0.5 ns	1.3 ns
	max	30.0**	3.0 ns	0.3 ns
velocity	min	1.1 ns	1.1 ns	1.3 ns
	max	0.6 ns	14.0**	1.2 ns
Ang.vel.hand	min	0.0 ns	57.6**	0.8 ns
	max	7.9**	220.6**	0.9 ns

*P <0.05
**P <0.01
ns: not significant

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In general, the cardiorespiratory responses appeared to increase with increasing seat height. The trends in oxygen cost, ventilation, and heart rate showed a minimum value at the conditions 100 and 120 degrees elbow angle. Mean heart rate increased with 4 to 6 $b.min^{-1}$ with seat height. Statistical analysis revealed a significant effect for oxygen cost between the lowest and highest seat positions. This difference ranged between 5 and 15 percent over the different velocities. Moreover, ME was significantly affected by seat height (**Table 2**), showing a difference of +0.5 to +1.0 percent between the lowest and highest seat position. The trends in the cardiorespiratory data are irrespective of the speed, as is indicated by the limited tendency for statistical interaction between the factors of "seat height" and "velocity" (**Table 2**).

As could be expected, all physiological parameters appeared highly dependent on velocity (i.e., P_o; **Table 2**). The relation between ME and speed is typically curvilinear.

Kinematics

Increments in seat height led to several shifts in the propulsion technique parameters (**Figure 3**). An expected decrease in push range was seen from a mean 1.7 radians at 100 degrees to 1.4 radians at 160 degrees. This is caused by a shift in both the start and end position of the hand on the handrim (**Table 2**). Simultaneously, a significant decrease in the push duration was found (**Figure 3**).

Although cycle time showed a slight decrease with increasing seat height, this trend was not significant. Since work/cycle is defined as the product of P_o and CT, no significant shifts in the amount of work per cycle can be expected with varying seat height. As was the case with cycle time, the recovery time did not show a significant trend with seat height position.

A kinematic analysis of seat height-dependent phenomena revealed several significant shifts in maximum and minimum values of the segment excursions (see **Figure 4** and **Table 2**). The mean maximum forward excursion of the trunk increased significantly with increasing seat height from 0.17 radians at $V = 1.39\ m.s^{-1}$ and 100 degrees to a mean maximum of 0.45 radians at 160 degrees. In contrast, the maximum retroflexion, and minimum and maximum values for abduction of the upper arm, decreased significantly with increasing seat height, whereas the range of elbow joint excursion showed a marked shift to an increased extension for both the minimum and maximum angles (**Figure 4**). Despite certain trends, no significant changes in the minimum and maximum angular velocities were found (**Figure 5**), although the maximum angular velocity of the hand on the rim decreased with increasing seat height (e.g., 5.15 $rad.s^{-1}$ to 4.65 $rad.s^{-1}$ at $V = 1.39\ m.s^{-1}$).

The moment in time at which maximum angular velocities were attained in the push phase showed a subsequent pattern for trunk flexion, upper arm anteflexion, and elbow extension, as is depicted in **Figure 6**. A decrease of these timing values is seen with increasing seat height.

The qualitative analysis of muscle activity during the push phase showed that both erector trunci and rectus abdominis muscles seem to prolong their activity with increasing seat height. As a muscle elevating the shoulder complex, mm. trapezius pars descendens tended to have a shorter active period with increasing seat height. M. serratus anterior did not show major shifts in activity, and can be viewed as a stabilizer and rotator of the scapula. Initial elbow flexion is accompanied by activity of the m. biceps and subsequent extension by m. triceps activity. With increasing seat height, the m. triceps tends to start sooner in relative terms, and prolong its activity. Activity of the mm. pectoralis major and deltoideus pars anterior tends to start sooner in the push phase over a shorter period.

As could be expected, the movement pattern showed marked shifts with increasing velocity (**Table 2**), leading to highly significant decreases in cycle time and push time, both in absolute (mean 0.81 to 0.34 s) and relative values (percent CT). Recovery time also showed a decrease with increasing speed (mean 1.29 to 1.05 s), resulting in an increase from 60.4 percent to 74.5 percent of the cycle time. The push range showed a minor but significant increase (1.5 to 1.6 radians at 120 degrees elbow angle), which was due mainly to an increase in the angular position of the hand at termination of the push phase. Work/cycle increased with speed from a mean 33 J at the lowest velocity, to 54 J at the highest (**Figure 3**).

The kinematic pattern showed several significant effects with velocity as well. The trunk showed a significant shift forward, and both the minimum abduction angle, and the maximum angle of

anteflexion, increased significantly. No changes were seen in the elbow joint (**Table 2**). As could be expected, the angular velocities increased in both the ab-/adduction and ante-/retroflexion range of motion of the upper arm and with respect to the maximum elbow extension velocity (3.7 $rad.s^{-1}$ to 5.8 $rad.s^{-1}$ at $V = 1.39\ m.s^{-1}$) (**Table 2**).

In general, the physiological and kinematic parameters showed seat height-dependent trends which were independent of the speed (in almost all parameters), as is stressed by the interaction values in **Table 2**.

DISCUSSION

Seat height

In the current study, seat height was individually adjusted and standardized in an experiment in which the effects of anthropometry-based seat height adjustment on physiology, kinematics, and muscle activity of steady-state wheelchair propulsion were evaluated. The philosophy of this approach is that proper fitting of the wheelchair to the user will lead to a reduction of physiological stresses of wheelchair propulsion, and subsequently will benefit the range of action of wheelchair users in general. It must be stressed that our findings with respect to the wheelchair as a mobility device may be out of line with the requirements of the wheelchair as a seating device. Differently-designed wheelchairs (with different ranges of adjustment) may result as most convenient solutions to both areas of wheelchair use. In the process of developing an anthropometry-based fitting model of the wheelchair-user interface for optimum conditions of wheelchair propulsion, it is recognized that other factors, such as the fore/aft position, and the width of the wheels, must be studied in the future.

The procedure of seat height adjustment studied here was based on the elbow angle in a standard sitting posture as an indicator of seat height. This procedure takes into account inter-individual variation in trunk and arm length. Probably due to this procedure for seat height standardization (and possibly standardization of power output in units of total weight), clear "seat height"-dependent trends were found in the physiological data, leading to significant effects for ME and oxygen cost. Previous studies of dynamic arm work (1,11,17) gave thought to the idea that optimization of arm position could diminish cardiorespiratory responses. Cummins and Gladden were unable to confirm an experimental effect of different height positions of a crank wheel in synchronous arm cranking (9). However, in this study, the reach of the arms was kept constant, which will have led to a more or less constant elbow excursion between different height conditions.

In handrim wheelchair propulsion, seat position has been previously studied by a number of investigators (4,5,6,16). These studies were unable to statistically confirm a "seat height"-related trend. However, inter-individual anthropometric variations in trunk and arm length were not taken into account. Results reported by Nordeen-Snyder stress the relevance of anthropometric standardization of saddle height in cycling (19). In the current results, the influence of anthropometry on oxygen cost was verified with a multiple regression analysis. The distance between the acromioclavicular joint and the wheel axis (AC-WA—which varies between subjects and with seat height) significantly improved the multiple correlation coefficient between power output and oxygen cost (Multiple R = 0.881) with 1.6 percent:

$$V_{O2}\ (l.min^{-1}, STPD) = 0.03307(P_o) + 0.009448(AC\text{-}WA) - 0.6495$$

Multiple R = 0.897; R-Square = 0.804; F(2,141) = 290.4 (P = 0.0000)

$t_{Po} = 23.6$ (P = 0.0000); $t_{AC-WA} = 4.47$ (P = 0.0009).

A minor significant improvement of the multiple R could be attained by adding biacromial width into the analysis (R = 0.901), which stresses the interdependency between wheelchair and body dimensions. However, these equations are based on a small group, and cannot merely be generalized for other groups of subjects.

The current results indicate that an elbow angle of 100 to 120 degrees, within the perspective of the standardization procedure described, is most appropriate. Moreover, the trend in the physiological data seems to level off at low seat heights, and probably increases at still lower seat heights. This was confirmed in preliminary data on a group of five subjects propelling a wheelchair ergometer at seat heights of 70 to 90 degrees elbow angle. A rising trend in heart rate and oxygen cost, and a decreasing trend in ME, was again seen at seat heights going down from 90 to 70 degrees elbow angle (27). These results imply that seat height in daily-use and

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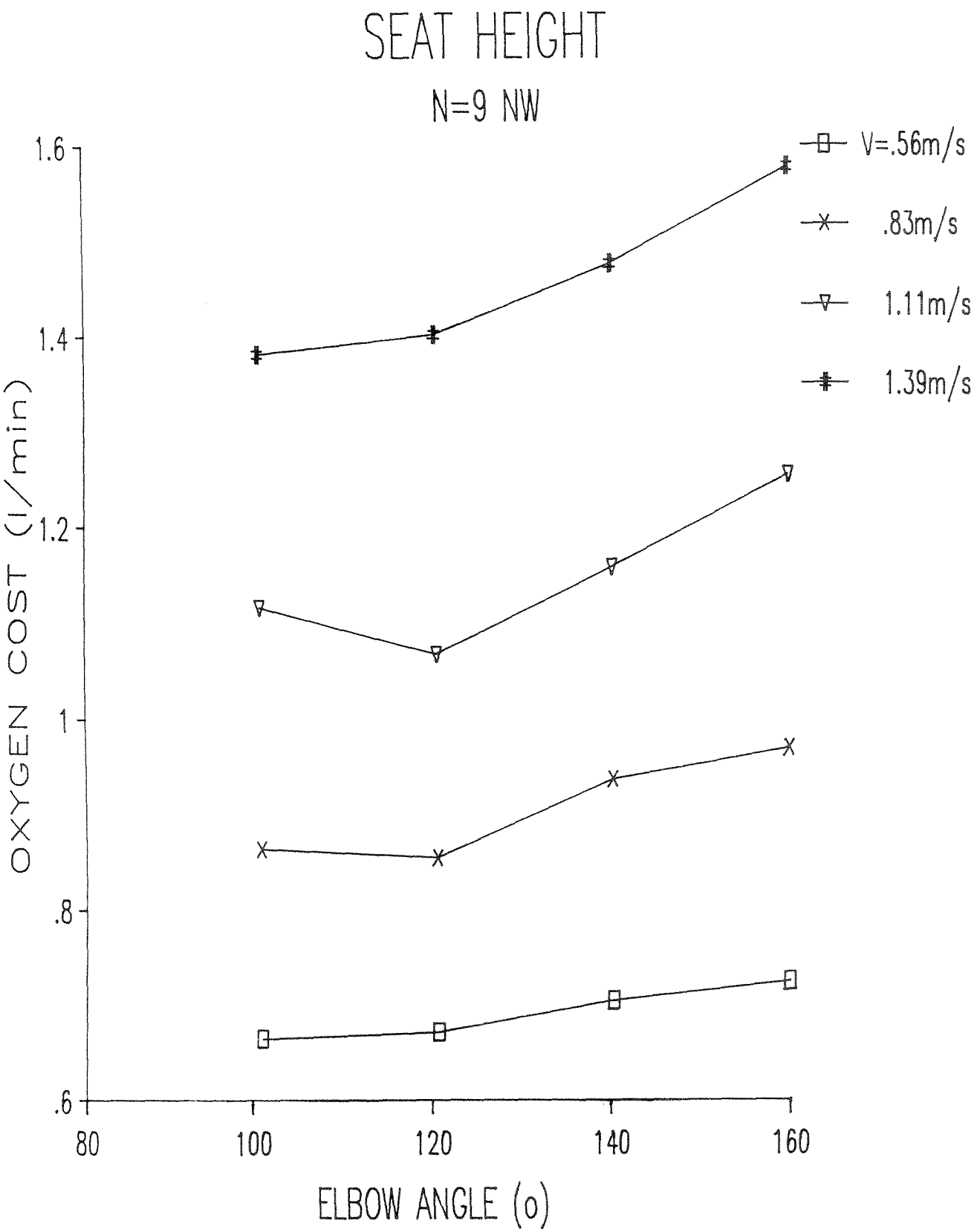

Figure 2a.
Mean values (N = 9) of oxygen cost, at four different seat heights (100 to 160 degrees elbow extension) and four velocities (0.55 to 1.39 $m.s^{-1}$).

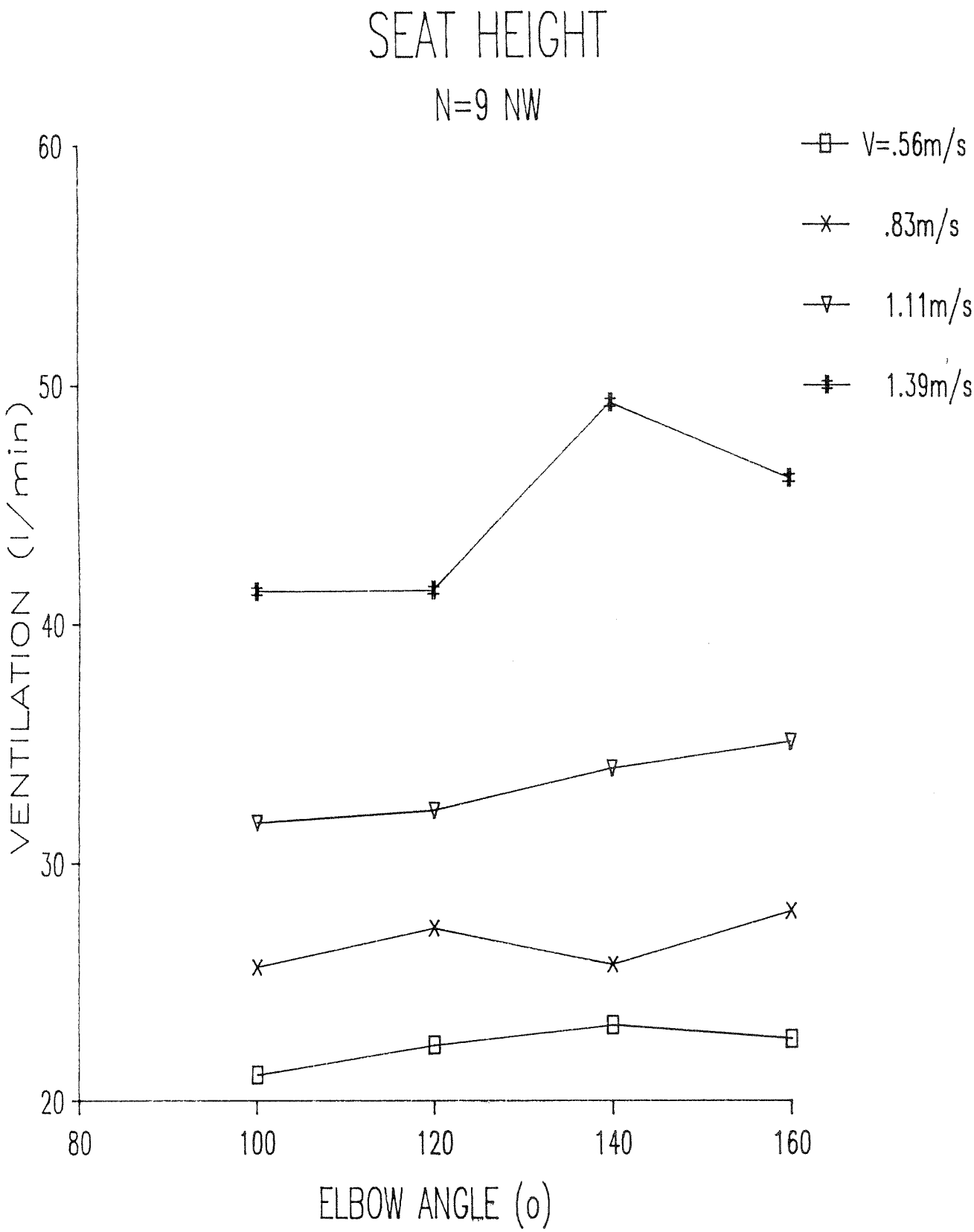

Figure 2b.
Mean values (N = 9) of ventilation, at four different seat heights (100 to 160 degrees elbow extension) and four velocities (0.55 to 1.39 $m.s^{-1}$).

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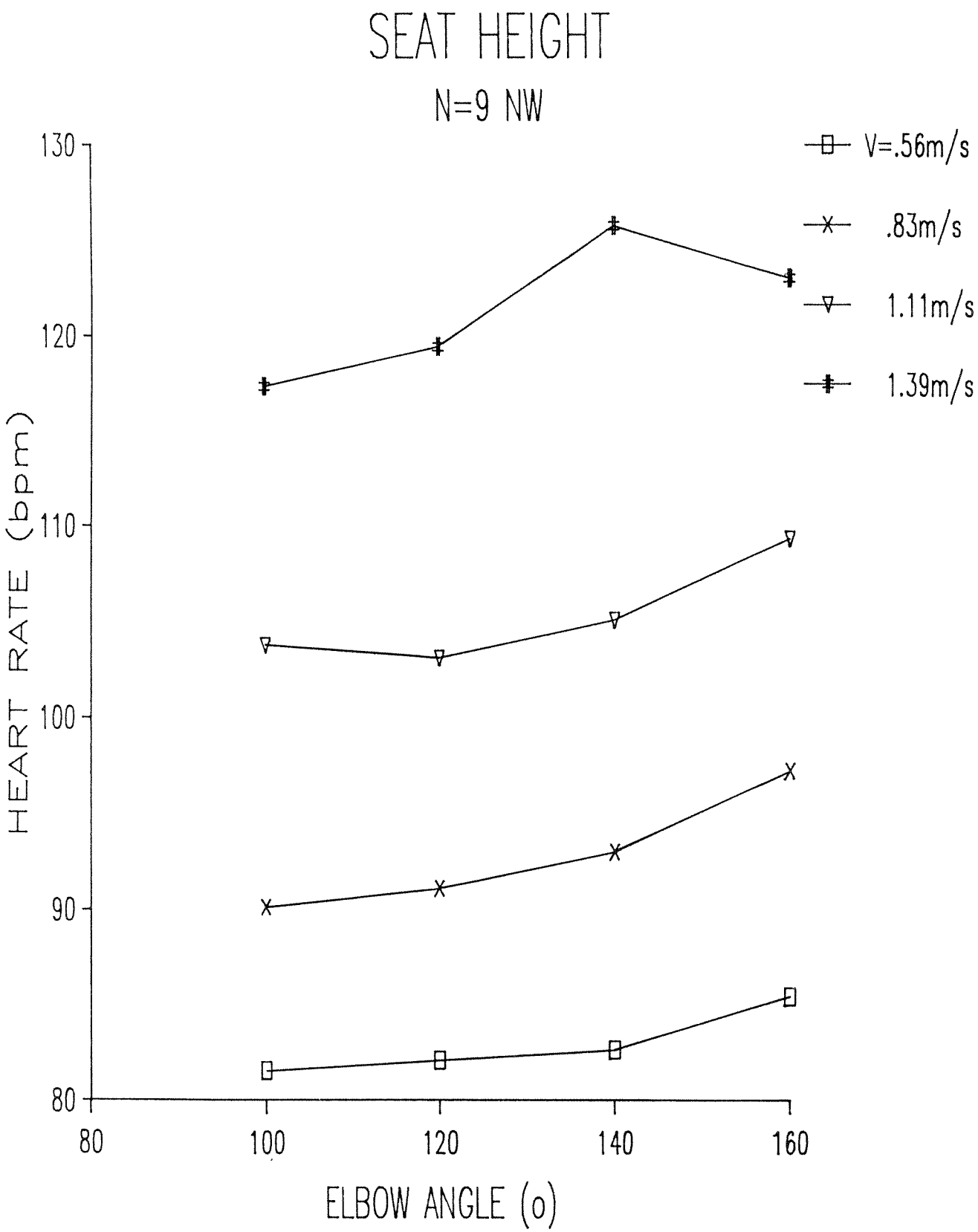

Figure 2c.
Mean values (N = 9) of heart rate, at four different seat heights (100 to 160 degrees elbow extension) and four velocities (0.55 to 1.39 $m.s^{-1}$).

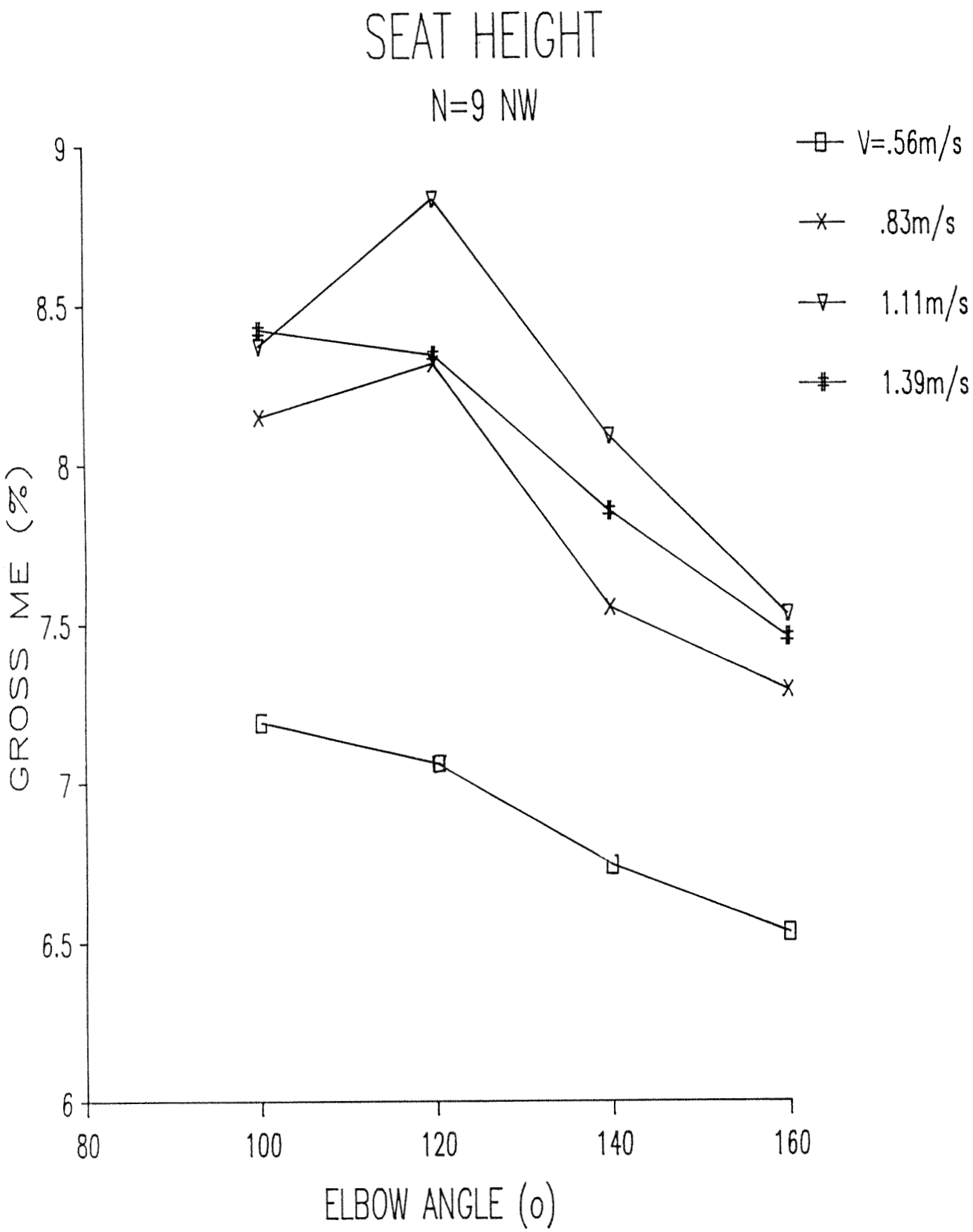

Figure 2d.
Mean values (N = 9) of gross mechanical efficiency at four different seat heights (100 to 160 degrees elbow extension) and four velocities (0.55 to 1.39 $m.s^{-1}$).

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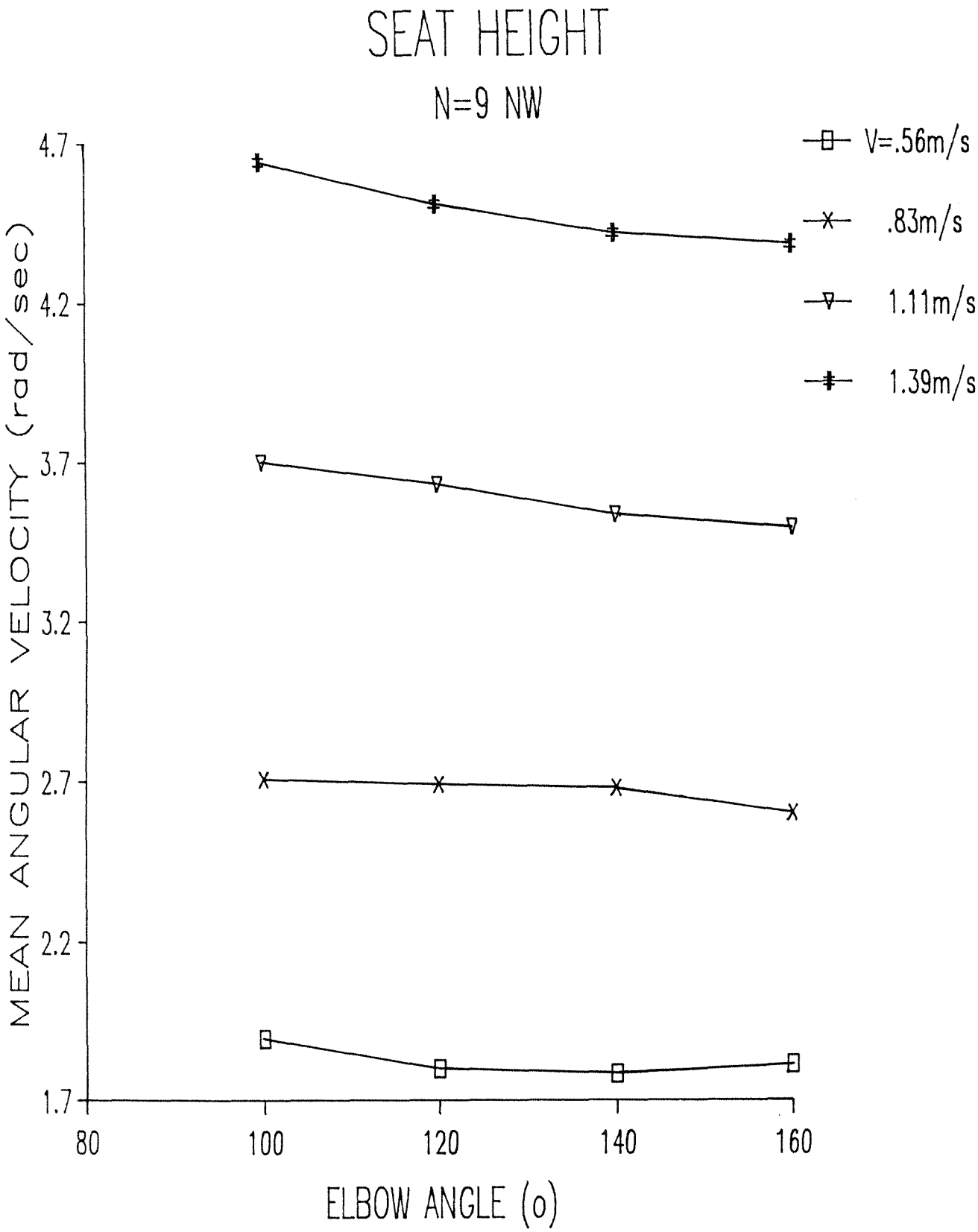

Figure 3a.
Mean values (N = 9) of mean angular velocity of the hand in the push phase at four different seat heights (100 to 160 degrees elbow extension) and four velocities (0.55 to 1.39 $m.s^{-1}$).

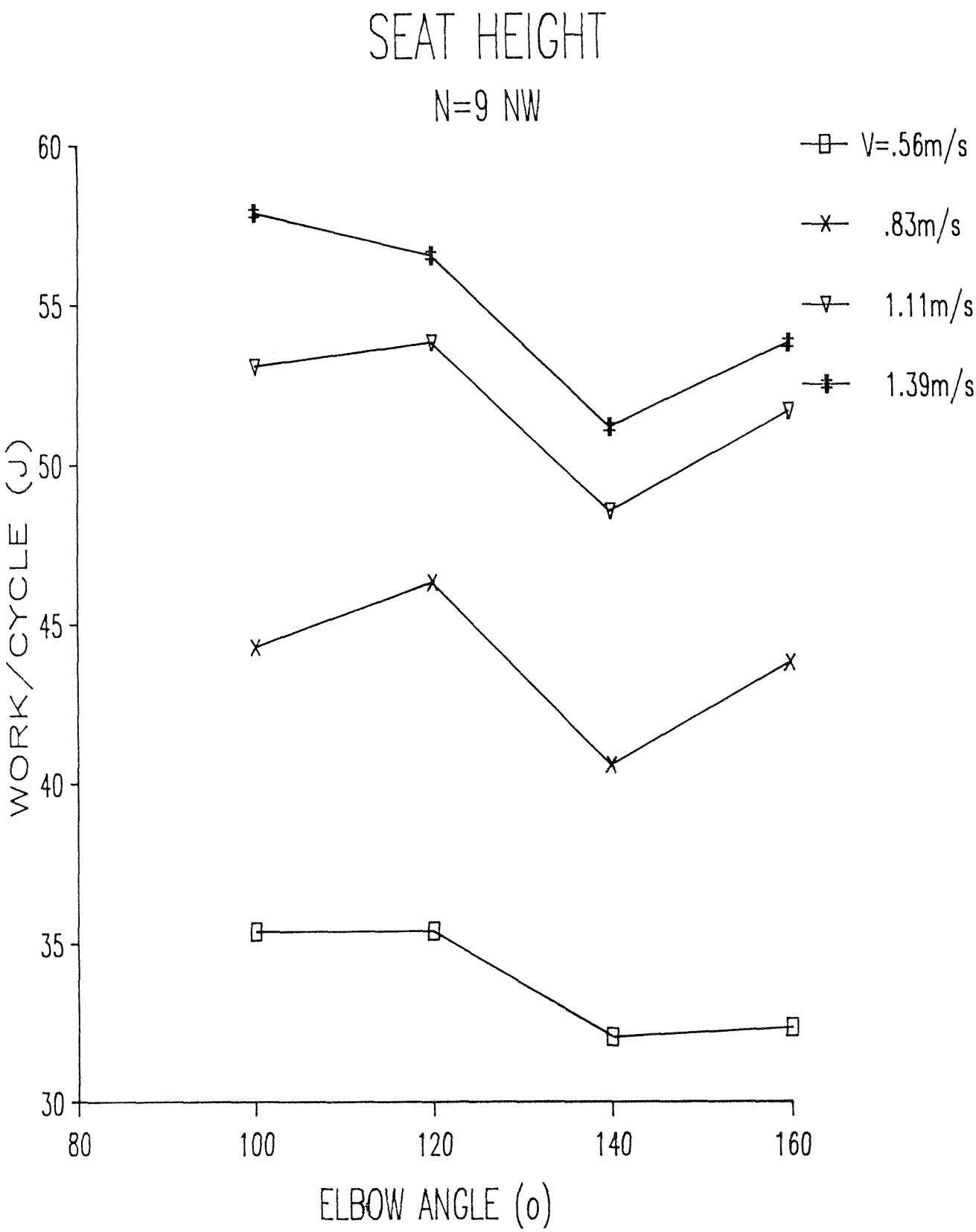

Figure 3b.
Mean values (N = 9) of mean angular velocity of the hand in the work/cycle at four different seat heights (100 to 160 degrees elbow extension) and four velocities (0.55 to 1.39 $m.s^{-1}$).

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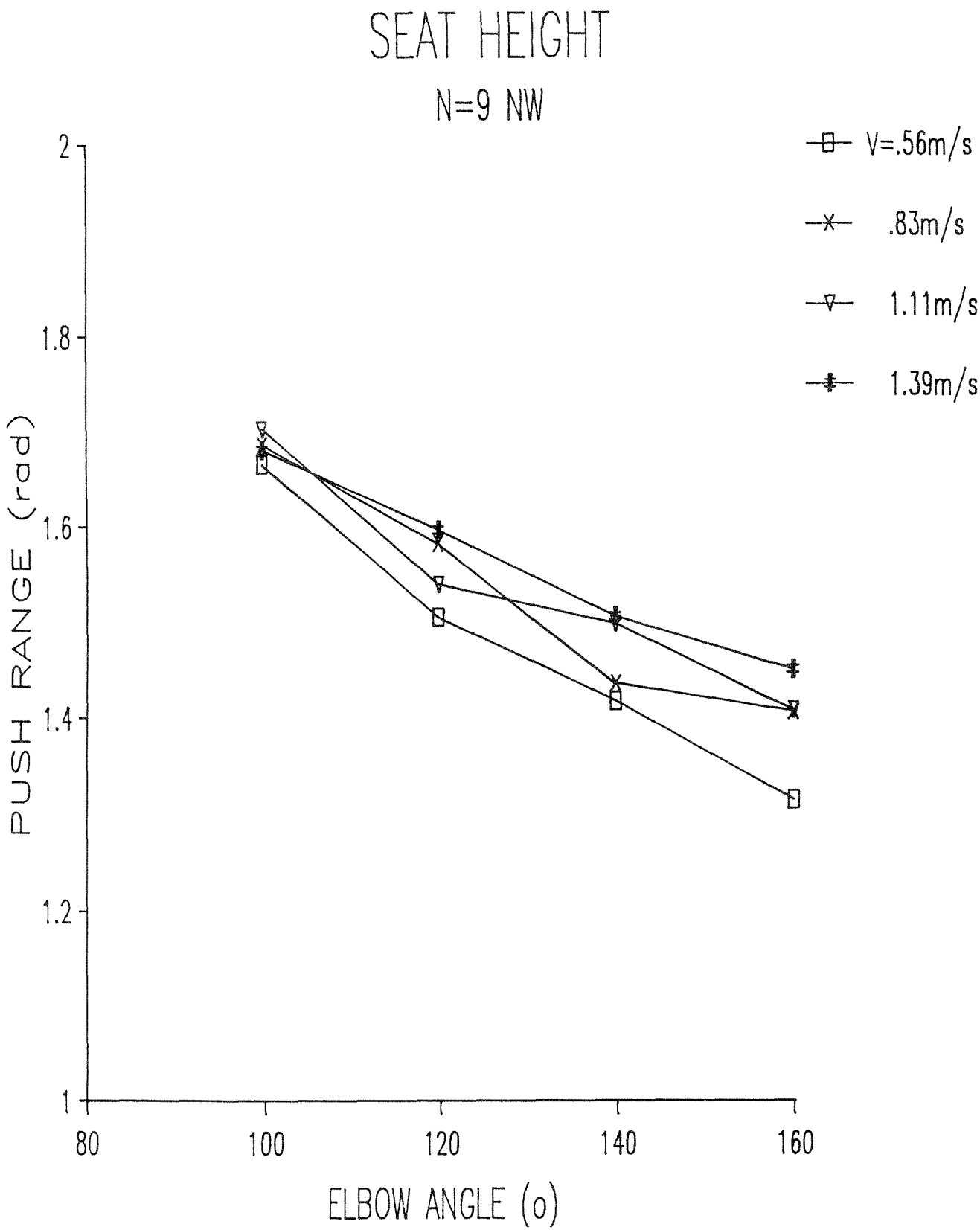

Figure 3c.
Mean values (N = 9) of mean angular velocity of the hand in the push range at four different seat heights (100 to 160 degrees elbow extension) and four velocities (0.55 to 1.39 $m.s^{-1}$).

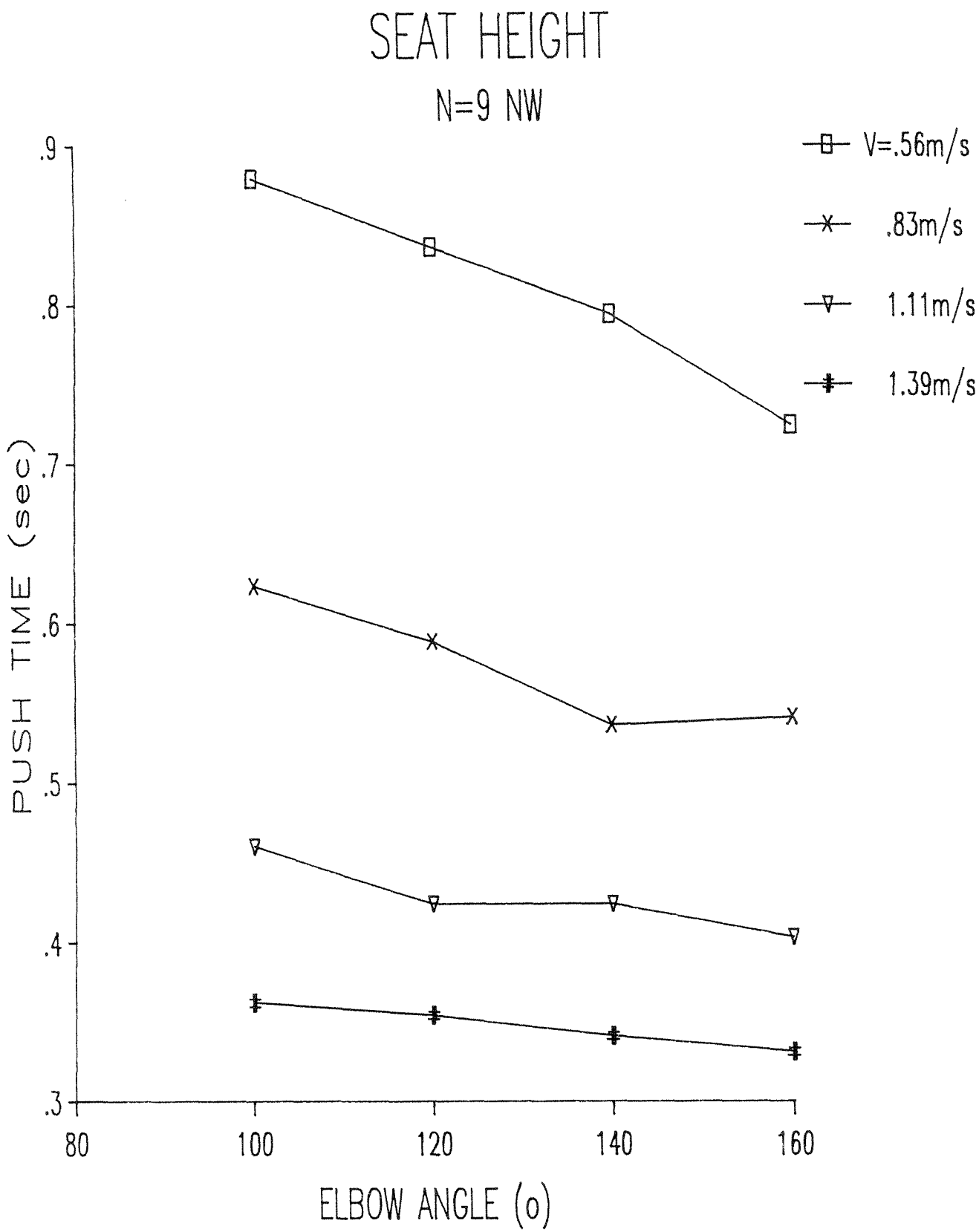

Figure 3d.
Mean values (N = 9) of mean angular velocity of the hand in the push time at four different seat heights (100 to 160 degrees elbow extension) and four velocities (0.55 to 1.39 $m.s^{-1}$).

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basketball wheelchairs is indeed curvilinearly related to cardiorespiratory parameters, with an optimum close to 100 to 120 degree elbow angle.

Although the trend in the physiological data with respect to seat height is quite clear, statistical evidence is not present for all physiological parameters. This may be due to the lack of experience and specific training in wheelchair propulsion of the subject group. Moreover, the method of anthropometric standardization for seat height is still an oversimplification of the actual (biomechanical) factors which seem important. The method is based on the assumption that anthropometric standardization of seat height is solely dependent on the length dimensions of arm and trunk, and that an optimum range of motion in the elbow joint alone can account for a decreased energy cost. Indeed, in this study, the mean range of flexion/extension during the push phase shifted from 94-143 degrees in the 100 degree condition, to 117-152 degrees at 160 degrees. This method does not take into account the inter-individual variation in, for example, shoulder-width and chest-depth, which factors influence the shoulder position, and thus affect the trajectory of the shoulder complex in both the frontal and sagittal plane. As is clearly indicated in the kinematic results, definite shifts in trunk and upper arm trajectories took place. The trajectories of trunk and wrist muscles should be optimized in conjunction with those acting at the elbow joint and shoulder

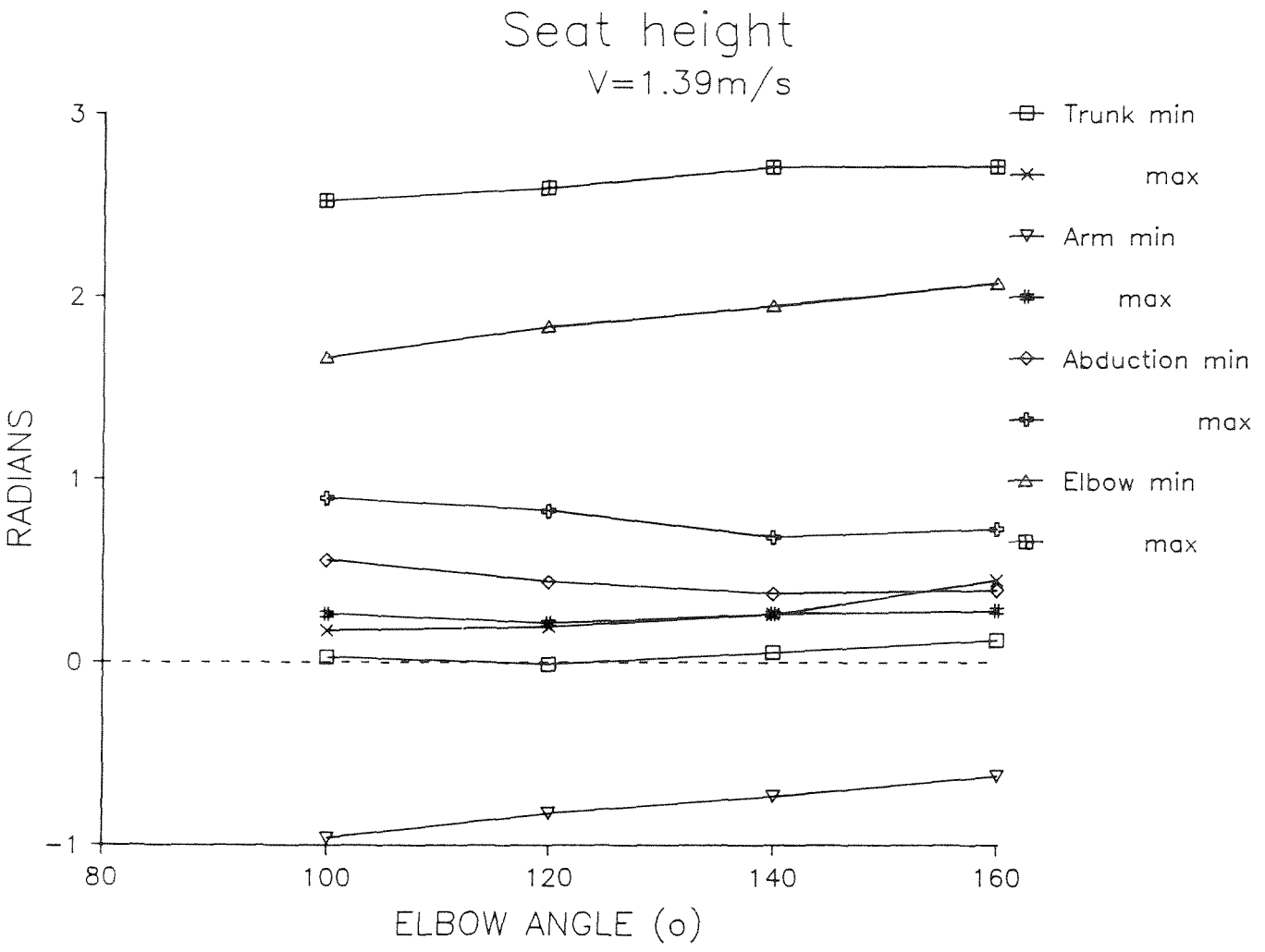

Figure 4.
Mean values (N=4, V=1.39 m.s^{-1}) of minimum (min) and maximum (max) joint excursions (rad) during the push phase of the trunk, upper arm in both sagittal (arm) and frontal plane (abduction), and of the elbow in relation to seat height, expressed in degrees elbow extension.

complex. It becomes even more complicated when one considers the possible role of mono-/multi-articular muscles and the possibility of energy transfer from proximal to distal joints (12). In accordance with vertical jumping (3), a sequential timing pattern of maximum angular velocities during the push phase was seen (**Figure 5**), which may indicate the possible presence of a similar mechanism of energy transfer in arm work. It is clearly seen in **Figure 5** that subjects tend to keep this timing sequence intact despite complications associated with increased seat height. Analysis of this complicated problem in arm work requires a thorough study of applied forces, and timing of muscular activity in combination with three-dimensional movement analysis. This study is difficult to conduct with complete wheelchairs on a treadmill. For this purpose, a specific computer-controlled wheelchair ergometer has been designed (18).

Movement analysis during the push phase revealed a decrease of the maximum values of elbow flexion, shoulder extension, and shoulder abduction with increasing seat height; maximum elbow extension and trunk flexion increased. The reason for these changes in movement pattern seems self-evident: the shifts in segment excursions compensate for the loss of reach of the hands with respect to the rim with increasing seat height. As a consequence, an increased activity of the trunk extensors was found. The compensatory mechanism of increased

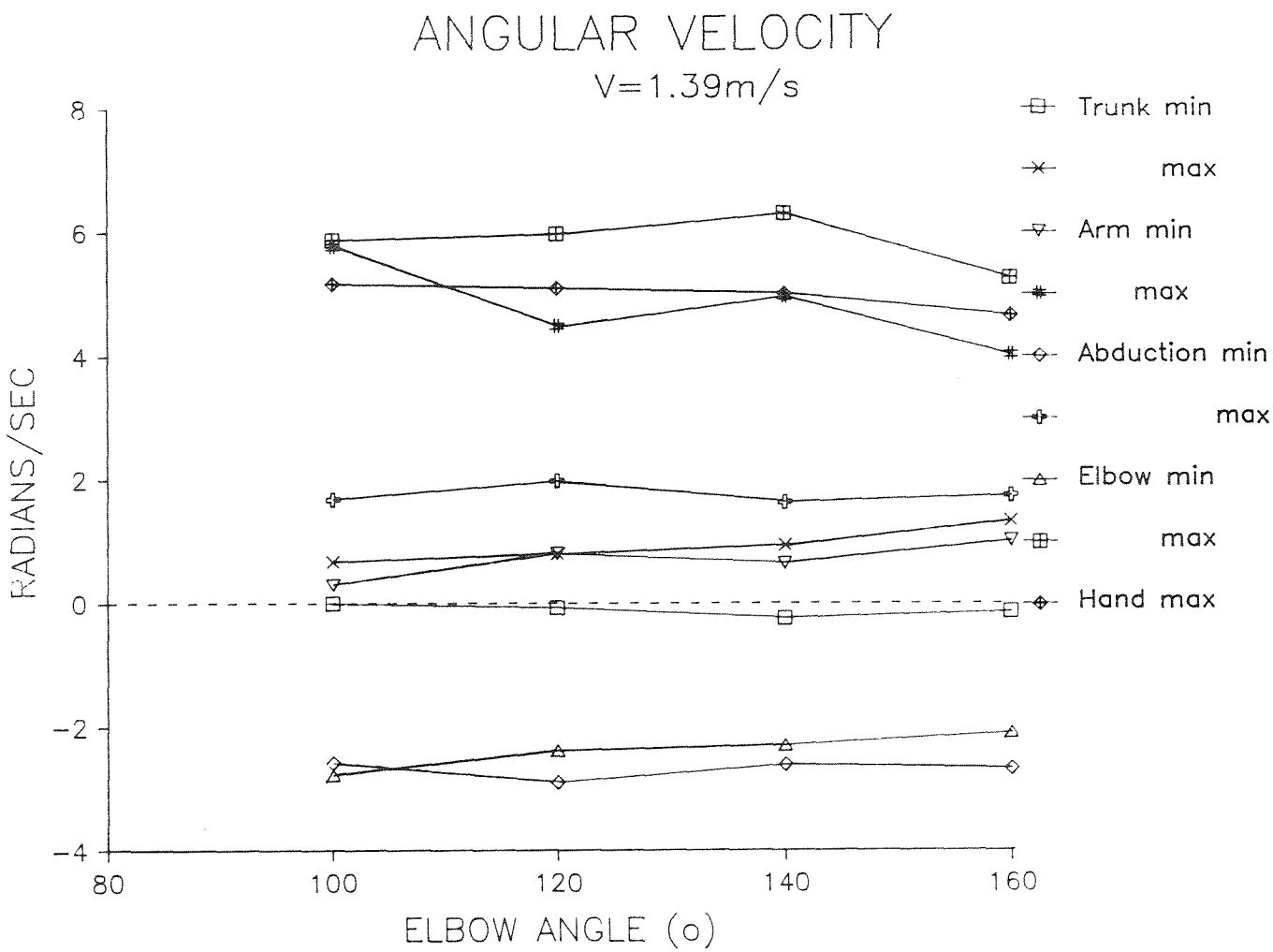

Figure 5.
Mean values (N = 4; V = 1.39 m.s^{-1}) of minimum and maximum angular velocity (rad.s^{-1}) during the push phase of the trunk, upper arm in both sagittal (arm) and frontal plane (abduction), and of the elbow and push angle (hand) in relation to seat height, expressed in degrees elbow extension.

trunk flexion cannot prevent a shift in the elbow trajectory and a decrease in the push range and duration. This will affect the effective time of torque generation, and thus may have influenced energy cost. In terms of technique, subjects tended to keep the push range intact. Similarly, the timing pattern of peak angular velocities remained intact over the different seat heights (**Figure 5**).

As was indicated by the physiological data, a seat height of 100 to 120 degrees elbow angle appeared most efficient under the conditions studied. The question remains as to where the increased cardiorespiratory responses originate from. In this respect, different factors may appear relevant:

- an increased trunk flexion during the push phase increases the reach of the arms with respect to the rims. Moreover, the weight of the trunk may assist in generating torque in the propulsion phase. But these aspects (will) lead to increased muscle activity of the trunk muscles, and thus enhance energy cost. In addition, a periodic increase in trunk flexion with increasing seat height will lead to a forward shift of the center of gravity, which leads to a periodic increase in rolling drag (Fr).
- higher seat positions lead to decreased levels of abduction, flexion, and extension of the upper arm, which in turn may affect the effectiveness of the (shortened activity of) mm. pectoralis major and deltoideus pars anterior as prime movers of the upper arm in the sagittal plane (21).
- an increase in the shoulder-to-rim distance will lead to an increased torque at the shoulder joint at

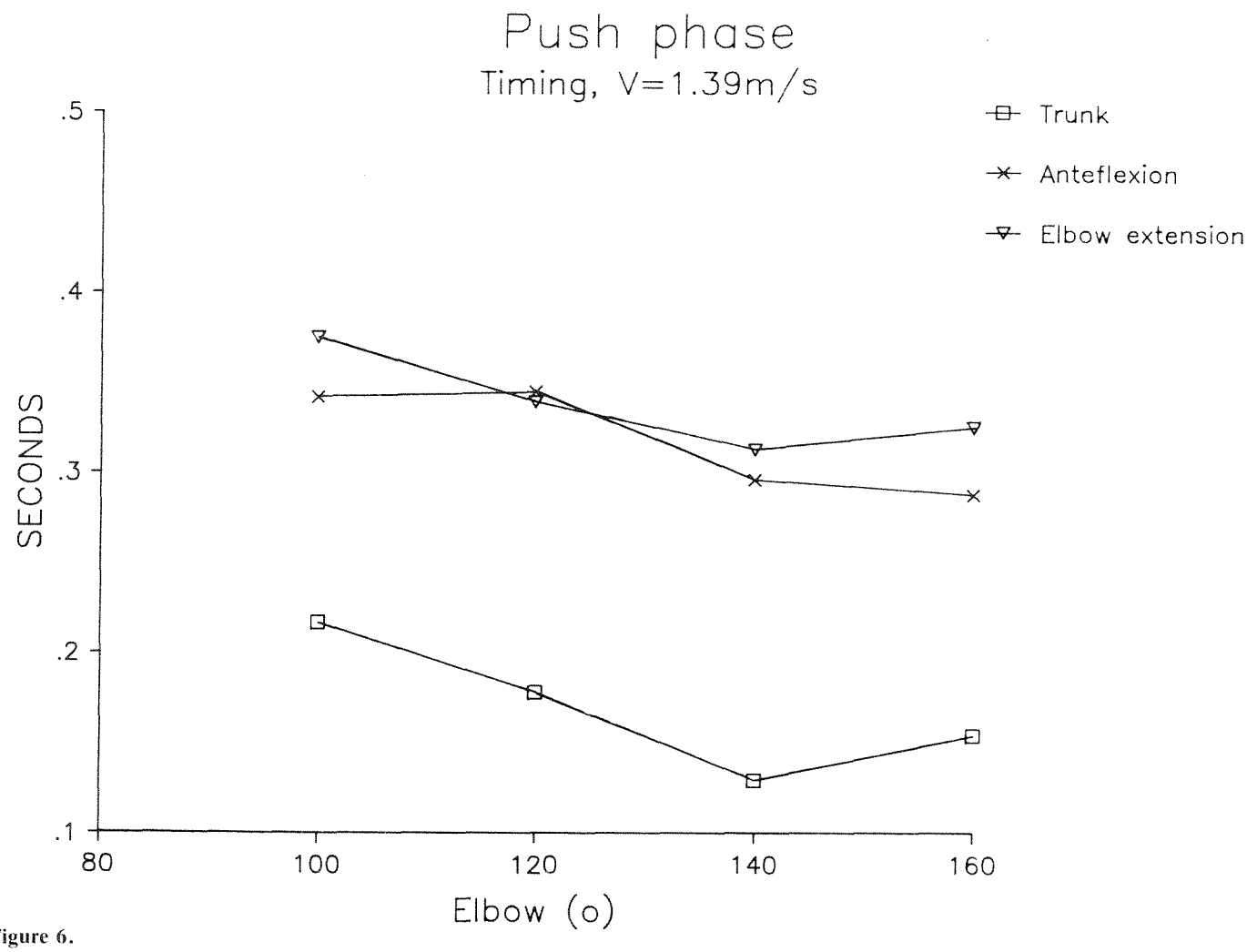

Figure 6.
Mean (N = 4; V = 1.39 m.s^{-1}) of the moment of occurrence (s) of the maximum angular velocity during the push phase for trunk flexion, upper arm flexion, and elbow extension in relation to seat height.

equal conditions of power output, which influences the muscle mechanics concerned, and consequently, may lead to a decreased efficiency.

- the shifts in elbow extension in the push phase will lead to a shift in the trajectory of the contracting flexor and extensor muscles, which again may enhance energy cost.
- the decreased push range and push time at higher seat heights must coincide with adaptations in the pattern of torque applied to the handrims, since an equal level of power output has to be generated in a shorter time span, and over a shorter trajectory. Much to our surprise, the maximum angular velocity (and acceleration) of the hand during the push phase decreased. One may speculate that these findings in propulsion technique with increasing seat height may lead much more to a change in the form (width) of the curve of the applied torque to the rim than to a change in the peak value alone. This may influence energy cost.

Previous studies focused on the hemodynamic effects of different arm positions (1,9). Although these factors may play a role, it is more likely that they become relevant in movement conditions with a large static or postural component, or where hydrostatic effects may be expected due to different positions with respect to gravity. A clear static aspect of handrim propulsion is the grip of the hand on the rim, which is, however, relatively short (0.85-0.35 s), and more or less identical between the different seat positions.

Velocity

As was described before, increased speed leads to major decreases in CT and PT, whereas work/cycle increases (25,26). A minor decrease in recovery time was found as well, together with an increase in the push range. The latter seems to coincide with an increased forward position of the trunk, and a reduced retroflection of the upper arm, which enables an increased reach to the rims (**Table 2**). Moreover, an increase in the abduction values is seen with velocity. This may, together with clear increments in the minimum and maximum angular velocities of the upper arm in both the frontal and sagittal plane, and the maximum velocities of the elbow, explain the required increase in power production with increasing wheelchair speed. As was previously suggested by Veeger, *et al.*, the increased abduction angle and angular velocity in the push phase, with increasing velocity, appears much more a side-effect of the activity of the m. pectoralis major in the act of anteflexion of the upper arm in combination with a closed-chain system between handrim, elbow, and shoulder complex (21).

CONCLUSIONS

The current results lead to a number of implications. First, further research of the wheelchair-user interface is required to generate a generally applicable fitting model of the wheelchair-user interface: lower and smaller intervals of seat height, the fore/aft position of the seat, the width of the wheels, and the angulation of seat and backrest. Secondly, the effect of different seat heights in groups of disabled wheelchair users requires careful study, for instance, regarding the suggested role of the trunk. And thirdly, other wheelchair designs, such as racing wheelchairs, may require other optima in terms of seat configuration. Understanding the relation between physiological processes and biomechanical phenomena requires analysis of effective torque and reaction forces to the handrim, and analysis of reaction forces of the seat and backrest, all in conjunction with movement pattern, electromyography, and physiology.

In conclusion, seat height adjustment is critical, and related to anthropometric dimensions. Optimum seat height in terms of cardiorespiratory responses is near 100 to 120 degrees elbow angle (procedure as described) for daily-use and basketball wheelchairs. Clearly, kinematic and technique parameters are associated with shifts in energy cost, and will be dealt with in more detail in future research. Moreover, overuse injuries in the musculoskeletal system may be prevented with proper fitting of the subject to the wheelchair. Optimum fitting of the wheelchair is especially important for persons with lower-body disabilities, because they have to rely on their arms and the intact part of the trunk for locomotion.

The geometry requirements presented here may contradict similar geometry requirements of the wheelchair in light of other activities of daily living (i.e., sitting at table or desk, working in the kitchen). It seems most appropriate to have different wheelchairs for these different tasks.

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