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Biomechanical analysis of wheelchair propulsion for various seating positions

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Abstract—The pattern of propulsion was investigated for five male paraplegics in six seating positions. The positions consisted of a combination of three horizontal rear-wheel positions at two seating heights on a single-purpose-built racing wheelchair. To simulate wheelchair propulsion in the laboratory, the wheelchair was mounted on high rotational inertia rollers. For three trials at each seating position, the subjects propelled the designed wheelchair at 60 percent of their maximal speed, which was determined at the beginning of the test session. At each trial, the propulsion technique of the subject was filmed at 50 Hz with a high-speed camera for one cycle, and the raw electromyographic (EMG) signal of the biceps brachii, triceps brachii, pectoralis major, deltoid anterior, and deltoid posterior muscles were simultaneously recorded for three consecutive cycles. The digitized film data were used to compute the angular kinematics of the upper body, while the EMG signals were processed to yield the linear envelope (LE EMG) and the integrated EMG (IEMG) of each muscle. The kinematic analysis revealed that the joint motions of the upper limbs were smoother for the Low positions—since they reached extension in a sequence (wrist, shoulder, and elbow), when compared to the High positions. Also, the elbow angular velocity slopes were found to be less abrupt for the Backward-Low position. It was observed that in lowering the seat position, less IEMG was recorded and the degrees of contact were lengthened. Among the seat positions evaluated, the Backward-Low position had the lowest overall IEMG and the Middle-Low position had the lowest pushing frequency. It was found that a change in seat position caused more variation in the IEMG for the triceps brachii, pectoralis major, and deltoid posterior. The trunk angular momentum was not found to be affected by a change in seat position which may be related to the variability among the subject's technique of propulsion or to a posture compensation.

Key words: *electromyography, kinematics, paraplegics, propulsion pattern, seat adjustment, wheelchair propulsion, wheelchair racing design.*

INTRODUCTION

In the past decade we have witnessed tremendous improvements in wheelchair racing records (1). The changes in performance observed over the years may, in part, be attributed to the development of better wheelchairs. Since the introduction of wheelchair racing in the 1940s, drastic changes have occurred in the design of wheelchairs. Some of the modifications have included: lowering the seat, cambering the rear wheels and moving them forward, changing the diameter of the rear wheels and the pushrims, and finally, modifying the frame (2). The athletes themselves are responsible for most of the improvements observed in wheelchair design. They have slowly modified the conventional wheelchair into a streamlined, lightweight racing machine according to their individual needs. Despite the

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changes that have occurred over the years, a large amount of uncertainty still exists as to the design of the ideal racing wheelchair.

There are several factors which may affect the performance of the athlete[1] including: the strength of the upper body, the physical capacity of the individual, the level of neurological lesion, and the interaction between the user and the wheelchair (3). Wheelchair design also plays an important role in optimizing the performance of an athlete. Different seat positions alter the athlete's pattern of propulsion and consequently affect his performance. One of the problems facing the athlete today is to define the ideal seat position needed to achieve an optimal propulsion technique (2). Therefore, research in wheelchair design may enable athletes to improve their performance through a better technique of propulsion.

Research related to the biomechanics of wheelchair propulsion is fairly recent with most of the published research appearing after 1980 (4). Experiments have been performed that related the seat position of different wheelchairs with the kinematics of wheelchair propulsion (5). Some experiments also have investigated the kinematic features of wheelchair propulsion using a racing wheelchair (6) or a conventional one (7). Brubaker, McLaurin, and Gibson studied the effect of varying the seat position on the mechanical efficiency of the chair and observed that both a Middle-Middle and a Middle-Forward seat position have higher mechanical efficiency (8). Both lower pushing frequency and smaller energy expenditure were associated with higher mechanical efficiency. Higgs' (9) static analysis of wheelchair racing, used at the 1980 Olympic Games for the Disabled, revealed that a Low-Backward seat position was highly correlated with success for long-distance athletes. Walsh, *et al.* (2) studied the effect of seat position on the maximal linear velocity of wheelchair sprinting and found that no significant differences existed in the maximal linear velocities with a change in seat position. Van der Woude, Veeger, Rozendal, and Sargeant (5) have shown a relationship between wheelchair seat height and both cardiorespiratory and kinematic parameters which were independent of the speed of propulsion. A lower mechanical efficiency was found at lower seat heights which corresponded to an elbow extension of 100 to 120 degrees. The push range showed a 15 to 20 degree decrease with an increase in seat height, resulting in a decrease in push duration. Some discrepancies seem to exist among the studies cited here; this may have resulted because the subjects' levels of neurological lesions were not taken into account by Higgs (9) and Walsh, *et al.* (2), while able-bodied individuals were used in Brubaker, *et al.* (8) and Van der Woude, *et al.* (5). The discrepancies may also have been caused by the use of different types of wheelchairs in these studies. Most importantly, the test conditions evaluated by each of these authors are quite different, with the conditions of the Brubaker group being far more extreme, thereby making a comparison of the studies even more difficult.

Muscle response is also influenced by the position of the user in relation to the pushrim, the resistive forces, and the level of disability (10). Brubaker, McLaurin, and McClay (11) related the seat position to electromyographic (EMG) activity and efficiency using lever arm propulsion. Their results indicated that the Middle-Middle and Middle-Backward seat positions had an overall lower EMG activity, which was reflected in higher efficiency. Brubaker, McClay, and McLaurin (12) also found that the lever arm propulsion[2] is not as sensitive to a change in seat position as is the pushrim propulsion. In using pushrim propulsion, Van der Woude, *et al.* (5) found that a higher seat position leads to a decreased level of abduction, flexion, and extension of the upper arm. This may in turn reduce the activity of the pectoralis major and the deltoid anterior muscles as prime movers of the upper arm during wheelchair propulsion.

At the present time, doubt remains as to the identification of the ideal seating position. This study was conducted in an attempt to provide more information specific to a change in seating position during wheelchair racing. Therefore, the purpose of this investigation was to examine the influence of six racing wheelchair seating position changes (consisting of a combination of three horizontal rear wheel positions at two sitting heights) on the

[1] L.H.V. Van der Woude, H.E.J. Veeger, and R.H. Rozendal, *Ergonomics of Manual Wheelchair Propulsion*. Free University, Amsterdam, The Netherlands, 1986 (unpublished).

[2] The lever arm propulsion will not be considered further, even if it seems to present some mechanical advantages, since this mode of propulsion is not used in wheelchair racing and does not truly represent the real pattern of pushrim propulsion.

Table 1.
Subject information.

Subject	Sex	Age (yrs)	Mass (kg)	Level of lesion*	Date of accident	Arm length (cm)	Trunk length (cm)
1	M	19	70.4	T11	87/31/10	68.1	34.0
2	M	22	47.7	T12	84/03/09	69.7	34.4
3	M	36	68.1	T12-L1	71/01/08	72.8	38.1
4	M	23	84.0	T12	82/18/10	72.1	40.1
5	M	37	50.0	T12-L1	69/26/12	70.4	45.8
$\overline{X}$		27.4	64.1			70.6	38.5
SD		7.5	13.6			1.68	4.31

*The neurological lesion at the spinal cord
SD = standard deviation $\overline{X}$ = mean

kinematic and electromyographic parameters of the upper limbs during wheelchair propulsion. The seating positions were contrasted in terms of criteria which are assumed to be related to the ideal seating position. The following criteria were assumed to be related to the ideal seating position: a low EMG activity and pushing frequency, and a longer recovery phase and recovery time. In using these criteria, the authors assumed that a low IEMG and a low pushing frequency combined with both a longer recovery phase and recovery time would result in a more efficient pushing pattern. A change in seat position refers to a change in the position of the seat in relation to the main axle, which is not necessarily obtained by actually moving the seat but may be brought about by moving the rear wheels.

METHODS

Five male paraplegics served as subjects for this investigation. **Table 1** provides a summary of information about each subject. They were selected from the Ottawa region on a voluntary basis and were free of any known pathological disorders of the upper extremities. All of the subjects who participated had to be classified as class IV (a neurological lesion from T11 to L2). Subjects within that class have full use of their upper extremities and possess good abdominal musculature as well as spinal extensors with a possibility of some hip flexion and adduction (13,14). This class was chosen because the technique of propulsion should not be affected by the level of lesion, since the upper body is fully functional. The level of physical fitness of the subject was also used as a selection criterion in order to avoid large variation among the subjects. The subjects used in this investigation were all physically active; either involved in wheelchair racing, wheelchair basketball, or actively played sled hockey.

Apparatus

All subjects were tested on an adjustable racing wheelchair which was built by Advance Mobility System Corporation according to our specifications. The following adjustments were possible with this chair: seat height; horizontal position of the rear wheel; wheel camber; and seat base and back rest inclination. The wheel camber is defined as the angle formed by the rear wheel and the positive vertical axis in the frontal plane. The seat *base* inclination is the angle of the seat made with the positive horizontal axis, while the seat *back* inclination is the angle formed by the seat base and the back seat of the wheelchair. The wheels and pushrims of the wheelchair had a diameter of 55.88 cm (22 inches) and 30.48 cm (12 inches), respectively. [For further details about the testing wheelchair see Mâsse (15).]

The subjects were evaluated in six experimental conditions consisting of three horizontal positions of the rear wheels (Forward, Middle, and Backward) and two seat heights (High and Low). The horizon-

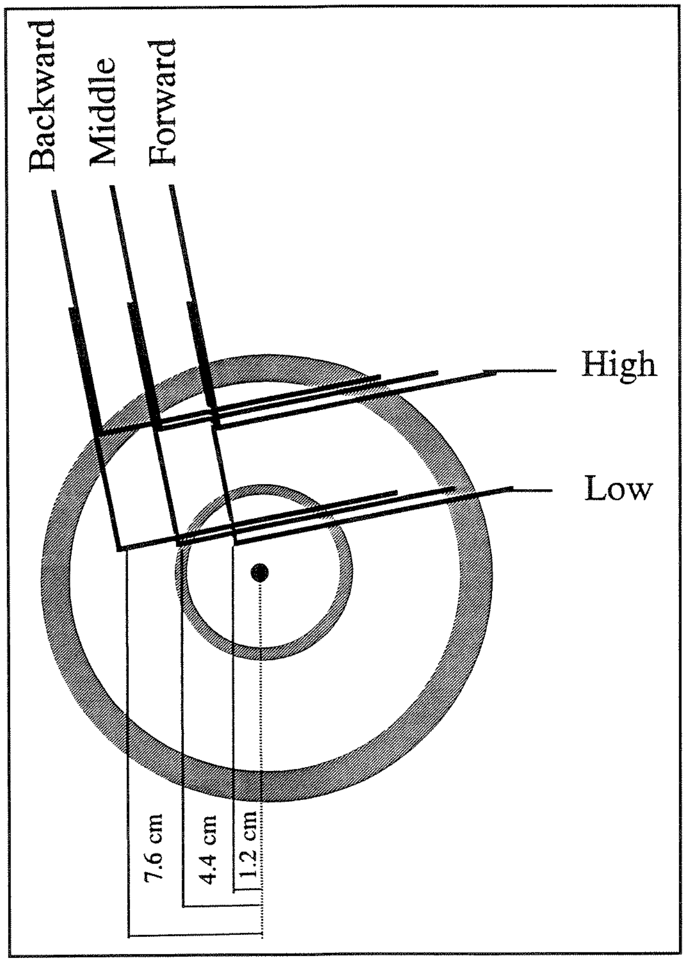

Figure 1.
Vertical and horizontal seating positions.

tal positions consisted of moving the rear wheels horizontally, which affected the position of the center of mass (CM) of the wheelchair. For each of the horizontal positions, this corresponded to a seat base and back rest intersection being located at 1.2 cm (Forward), 4.4 cm (Middle), and 7.6 cm (Backward) behind the main axles (see **Figure 1**). A modified reaction board technique (16) was used to determine the CM of the wheelchair for these horizontal positions. The CM for the Forward, Middle, and Backward seat positions was found to be located at 10 cm, 6 cm, and 3 cm in front of the main axles, respectively. These selected positions were believed to represent a range between the conventional and racing wheelchair which was reported by Peizer, Wright, and Freiberger (17). The High and Low positions were established by moving the seat. Both positions depended on the arm and trunk lengths of the subject. The Low position represented the position at which the distal phalanges of the second fingers of the subject's hands were aligned with the lowest portion of the pushrims.

In some cases, the location of the Low position depended on the design of the wheelchair, but for all of the subjects it was found to correspond closely with the desired position (having the distal phalanges of the second fingers aligned with the lowest portion of the pushrims). The High position was set at 10 percent of the subject's arm length above the Low position. The Low and High positions were selected because they represented a range of positions used by wheelchair athletes (9,18). For each experimental condition, the wheel camber, seat base, and back rest angle remained constant. These positions were determined according to Higgs (9), and York and Kimura (18) as being those positions most often used by wheelchair athletes and which correlated with a high level of success. The wheel camber was set at 8 degrees from the vertical; the seat base at 12.5 degrees from the horizontal; and the back rest at 90 degrees from the seat base.

To simulate wheelchair propulsion in the laboratory, the testing wheelchair was mounted on rollers. Because the rollers were hollow, and in order to simulate actual locomotion, each roller mass was modified by adding two iron rings to increase the rotational inertia of the system. To monitor the speed of propulsion, a tachometer mounted on the side of the rollers was calibrated to an accuracy of more than 95 percent. [For a more detailed description of the rollers see Mâsse (15).]

Procedures

Prior to testing, informed consent was obtained from each subject. Markers were placed on the appropriate joint center locations to facilitate the ensuing kinematic data reduction of the upper body segments. The anthropometric measurements of the subject's upper body (see **Table 1**) were recorded according to Jette's procedure (19). The Low sitting position was measured with the shoulders aligned with the main axles. The subject was asked to keep his back against the back rest, keep his head straight, and have his arms lay against the pushrims. The seat was then moved vertically until the subject's distal phalanges of the second fingers were aligned with the lowest portion of the pushrims.

EMG surface electrodes were fixed over the following muscles: the biceps brachii (long head), the triceps brachii (lateral head), the pectoralis major, the deltoid anterior, and the deltoid posterior. These muscles were investigated since the triceps brachii, the pectoralis major, and the deltoid anterior are believed to be the prime movers in wheelchair propulsion (4) and the biceps brachii and deltoid posterior were analyzed because they are presumed to be involved during wheelchair propulsion (10,20). Disposable silver/silver chloride surface electrodes (Medi-trace) were placed over the motor points of each muscle in the manner described by Delagi, Perotto, Iazzetti, and Morrison (21). Before affixing the electrodes, the skin was rubbed with alcohol, shaved, and rubbed with electrolyte paste. Skin resistance was then measured with an ohmmeter. If the impedance exceeded 2 kΩ, the area was cleaned again and new electrodes were applied. The electrode wires were taped to the skin or clothing to reduce movement artifacts and to allow freedom of movement.

After a warm-up period, the maximal speed of propulsion for the subject was obtained by incrementing his speed until he felt that maximal speed was reached. Three trials of the maximal speed were recorded for each subject, the maximal speed average was calculated, and 60 percent of that speed was used in the remainder of the experiment (see **Table 2**). The maximal oxygen uptake was not used to set the intensity of exercise because the physiological capacity of a paraplegic differs from that of a nondisabled person (22). In order to make comparisons among seating positions, it was important that the speed remain constant for all seat positions being tested. The maximum voluntary isometric contraction (MVC) for each muscle was recorded according to the procedure used by Delagi, *et al.* (21). For each muscle, the subject was asked to perform a maximum voluntary isometric contraction for a brief period of 5 sec. Recordings were made using a data acquisition system (23) and the data were stored on a microcomputer (Compaq 386, 16 MHz). This procedure was performed before and after testing for each experimental condition.

Finally, the subject was asked to propel the designed wheelchair at each experimental condition [Forward and High (FH), Forward and Low (FL), Middle and High (MH), Middle and Low (ML), Backward and High (BH), and Backward and Low (BL)], at 60 percent of his maximal speed of propulsion. At each condition, the subject was asked to propel the designed wheelchair for three trials at a constant speed of propulsion for a period of 90 sec. For each trial, one cycle was filmed at 50 Hz with a high-speed camera (Locam II) placed perpendicular to the sagittal plane of the subject (approximately 11 meters away). Simultaneously, the muscular activity was recorded for at least three cycles at each trial. The raw electromyographic (EMG) signals were recorded at 1,000 Hz for a period of 5 sec for each trial. The signals were fed to a bioamplifier (University of Ottawa, input impedance of 10 MΩ, 10-700 Hz bandpass), digitally converted by a data acquisition system, and stored in the memory of the microcomputer using the BIOAD system (23). In between trials, the subjects had approximately 2 minutes of rest.

Table 2.
Speed of propulsion (m/s).

Subject	Max	(SD)	60%
1	5.36	(0.47)	3.22
2	6.13	(0.22)	3.59
3	6.14	(0.29)	3.59
4	6.36	(0.26)	3.82
5	5.77	(0.25)	3.47
$\overline{X}$	5.92		3.54
SD	0.35		0.19

SD = standard deviation $\overline{X}$ = mean

Data reduction and analysis

The phases of propulsion were established from the speed of the rear wheels as measured by a tachometer. The pushing phase represented the time when an increase in the speed of the rear wheels was observed. The recovery phase was characterized by a decrease in speed. The cycle time was defined as the amount of time required to perform one cycle of propulsion. The degree of contact was calculated from the film data: it represented the angle formed by the point at which the hand contacted the pushrim, the main axle of the rear wheels, and the point at which the hand released the pushrim. The pushing frequency was defined as the number of strokes per minute.

All EMG data were processed on the microcomputer with the BIOPROC program (23). The raw

EMG signal bias was removed by zeroing the mean, the signal was then full-wave rectified and filtered at 6 Hz (single-pass, critically dampened, digital filter) to obtain a linear envelope (LE EMG). For each cycle, LE EMG signal for each muscle was normalized over time and was also normalized by amplitude using the MVC values. The LE EMG ensemble average of each condition was calculated from 45 cycles (5 subjects, 3 trials, and 3 cycles).

Eight markers for each frame were digitized on a Hewlett Packard 9874A digitizing system. The marker coordinates were transferred to a mainframe computer (Amdahl IBM VM/SP, University of Ottawa) for complete data processing. The filmed data were projected at 9.3 percent of the life-size image and digitized to an accuracy of 1.98 mm. The digitized coordinates were filtered (second-order, Butterworth, low-pass filter) at a cutoff frequency of 6 Hz. The segmental angular velocities, accelerations, and momenta were calculated for a complete wheelchair cycle by processing the digitized cinefilm coordinates with the BIOMECH package (University of Ottawa) as defined by Winter (24). The data were normalized over time for each condition and subject and ensemble-averaged for three complete cycles of propulsion. Thus, three complete cycles of propulsion ($n=3$) were used to compute within-subject ensemble averages which consisted of taking one cycle of propulsion from each trial (3 trials were recorded per condition). These ensemble averages were in turn averaged across all subjects ($n=5$) to yield a grand ensemble normalized average (3 trials per subject $\times$ five subjects, or $n=15$) for each condition.

Both descriptive analyses and descriptive statistics were performed on the EMG and kinematic data, to examine the differences among the seat positions. The descriptive statistics used for the EMG consisted of integrating the within-subject ($n=9$) normalized average LE EMG using trapezoidal integration, which yield the integrated EMG (IEMG). These IEMG were ensemble-averaged across subjects ($n=5$) to give a grand ensemble normalized IEMG (3 cycles per trial $\times$ 3 trials per subject $\times$ 5 subjects, or $n=45$) for each condition. For the kinematic data, the descriptive statistics comprised the comparison of the cycle time, pushing phase, recovery phase, pushing time, recovery time, pushing frequency, and degree of contact. A grand ensemble average (3 trials per subject $\times$ 5 subjects, or $n=15$) was computed on each of the above kinematic parameters for each condition. Also, a visual inspection of the averaged angular velocities, accelerations, and momenta, and LE EMG curves was used to further contrast the seating positions.

Nonparametric statistics were used to help identify the ideal seating position based on the following criteria which were assumed to be associated with an ideal seating position: a low IEMG activity and pushing frequency, and a high recovery time and recovery phase. These criteria were used since smaller energy expenditures are believed to be linearly related to IEMG (25,26) during cycle ergometry and are also associated with lower pushing frequency during wheelchair propulsion (8). A high recovery phase and time were presumed to be associated with an ideal seating position. Only these criteria were used for the statistical analysis as they were the only criteria which could be easily quantified. The joint analysis of these criteria was made possible by ranking the mean values of the criteria across the seating positions. This was done in order to transform the criteria on the same scale. A Friedman two-way analysis of variance (ANOVA) by ranks was then performed to test the null hypothesis that the median of the criteria were the same across the seating positions. The Friedman analysis may only be used in this situation if we assume that the criteria were "biomechanical samples" on which the positions may be contrasted. It is realized that this assumption may not be totally tenable; however, in the absence of better statistical techniques, this may provide a standard against which the positions may be contrasted given the small number of subjects. This analysis further assumed that all the criteria were equally important in the identification of the ideal seating position. If more subjects were available, the data would be better analyzed through a two-way repeated multivariate analysis on the ranks using Puri and Sen L statistics (27).

RESULTS AND DISCUSSION

The mean ($n=15$) speed of the rear wheels, pushing frequency, and degree of contact for each seating position are presented in **Table 3. Figure 2** shows the mean cycle time, pushing time, and recovery time for all the trials and subjects at each seating position. It was observed that lowering the

Table 3.
Means (n = 15, 3 trials and 5 subjects) and standard deviations of the speeds, pushing frequency and degree of contact for each seating position.

Kinematic parameters		Positions BL	ML	FL	BH	MH	FH
Speeds (m/s)	$\overline{X}$	3.72	3.65	3.62	3.63	3.66	3.58
	SD	(0.26)	(0.28)	(0.29)	(0.28)	(0.29)	(03.0)
Pushing freq. (Strokes/min)	$\overline{X}$	77.6	73.5	77.9	82.6	78.6	78.9
	SD	(23.5)	(26.2)	(27.2)	(32.0)	(34.0)	(28.8)
Degree of contact (Degrees)	$\overline{X}$	108.5	108.6	111.0	97.1	100.	95.7
	SD	(14.8)	(12.1)	(15.1)	(11.6)	(11.6)	(13.5)

BL = Backward-Low
ML = Middle-Low
FL = Forward-Low
SD = Standard Deviation
BH = Backward-High
MH = Middle-High
FH = Forward-High
$\overline{X}$ = Mean

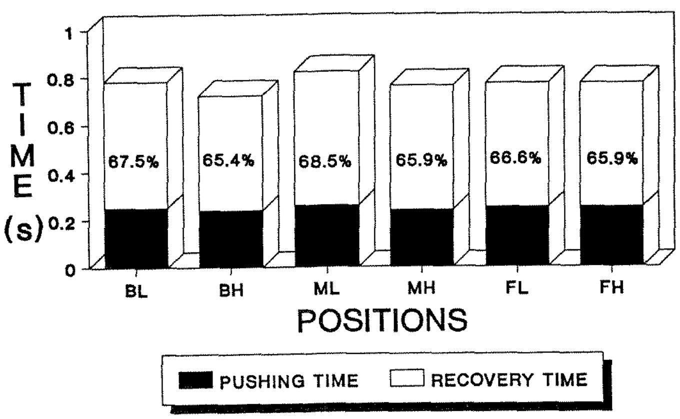

Figure 2.
Mean cycle time, pushing time, and recovery time for all the seat positions: Backward-Low (BL), Backward-High (BH), Middle-Low (ML), Middle-High (MH), Forward-Low (FL), and Forward-High (FH). The percentage of time spent in recovery (recovery phase) is also indicated.

seat position caused an increase in cycle and recovery times, and a decrease in pushing frequency for the Backward and Middle positions. By contrast, no such variation was found among these parameters for the Forward positions. Also, the percentage of time spent in each of the phases of propulsion (pushing and recovery phases) did not differ with changes in seat position. This implied that the Backward- and Middle-Low positions would not be as physiologically demanding as their corresponding High positions, since the subjects had to stroke less often and spend more time in recovery. However, for the Forward positions, the level of exertion should be similar since there were no differences observed in the cycle time and pushing frequency.

It was seen that greater degrees of contact and cycle time resulted in slower pushing frequencies for the Backward- and Middle-Low positions when compared to their corresponding High positions. Therefore, a change in seat position might affect the speed of propulsion during a wheelchair racing event since an inverse relationship was found between the pushing frequency and both the cycle time and the degree of contact. However, Walsh, *et al.* (2) did not find any significant difference in speed with a change in seat position, which might have resulted from the high variability in the levels of lesion of the subjects used in their study.

In moving the seat down, the proportion of the pushrim accessible for pushing was lengthened since greater degrees of contact were found with lower seat positions (see **Table 3**). This was in agreement with the results of Brubaker, *et al.* (8) who found that the Middle positions had longer degrees of contact than did the other positions, whereas this only occurred for the High positions in our study. In our study of the Low positions, the longest degree of contact was observed for the Forward position. The Middle-Low and the Backward-Low degrees of contact were similar, but the pushing frequency was lower for the Middle-Low position. This implies that in moving the seat position, the location and orientation at which the hands contact the pushrims

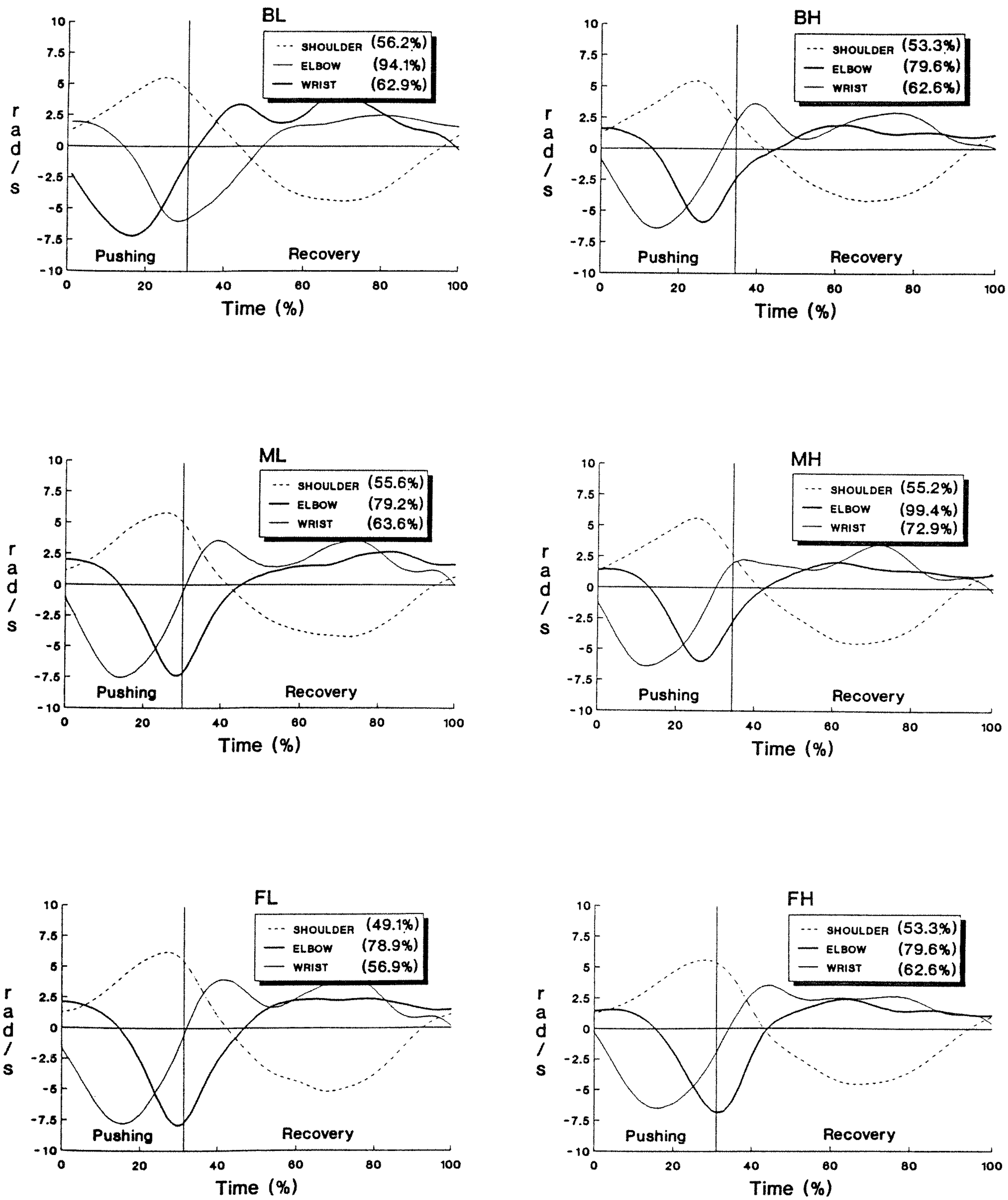

Figure 3.
Mean angular velocities of the upper limb joints for all the seat positions: Backward-Low (BL), Backward-High (BH), Middle-Low (ML), Middle-High (MH), Forward-Low (FL), Forward-High (FH). Coefficients of variations are indicated in parentheses.

was varied, which may have resulted in a change in the pushing frequency. Consequently, the location and orientation with which the hands contact the pushrims can affect the speed of propulsion during a wheelchair racing event through varying the angle at which the forces are applied to the pushrims. This would be in agreement with Brubaker and McLaurin (3) and Sanderson and Sommer (7).

The mean angular velocities of the upper limb joints for all the trials and subjects at each seating position are shown in **Figure 3**. The peak angular velocity of extension for the elbow was found to correspond to the end of the pushing phase, except for the Backward- and Middle-High positions. Therefore, for the Backward- and Middle-High positions, the timing at which the peak angular velocity of extension for the elbow occurred was off, since it was not timed with the end of the pushing phase. This indicated a poor synchronization in the joint motion at these seating positions (Backward- and Middle-High). It was also observed that at all the Low positions, the change of motion of the upper limb joint occurred in a sequence (approximately 45 percent of the cycle). The sequence was as follows: the first joint to reach flexion was the wrist, followed by the shoulder extension, and then elbow flexion. At the High positions, the change of motion of the shoulder and elbow joints occurred at the same time (approximately 40 percent to 45 percent of the cycle). Consequently, the motion of the upper limb joints was flowing more smoothly for the Low positions since the changes of motion of the joints were more sequential in comparison to the High positions.

Some differences in the peak angular velocities during the pushing phase were observed among the seat positions (see **Figure 3**). The peak angular velocities of extension for the wrist and elbow were found to be slightly higher for the Low positions in comparison to their High positions. Therefore, the elbow and wrist were not moved as much for the High positions because lower seat positions allow greater extension of the elbow and wrist joints. This occurred because with higher seat positions, the elbow and wrist assume a more extended position to reach the top of the pushrims than for lower positions (i.e., the top of the pushrims are located lower). The observed variation in the peak angular velocities during the pushing phase might explain some of the differences which were earlier found in the pushing frequency. Lower pushing frequencies may have resulted from greater angular velocities of the elbow and wrist during the pushing phase. This can result in a stronger pushing phase for the lower seat positions, allowing the subjects to spend more time in recovery, consequently decreasing their pushing frequency. However, it should be noted that the differences in the peak angular velocities and pushing frequencies were not large.

The Backward-Low position peak angular velocities during the pushing phase were lower than for the other Low positions, mainly for the elbow joint. Also, the Backward-Low position change of motion of the elbow was less abrupt than for the other positions tested as shown by the smoothness of the flexion and extension slopes. It was found that the lower peak angular velocities observed for the Backward-Low position did not translate into a greater pushing frequency as did the High positions. Therefore, at this position (Backward-Low), the work during the pushing phase might be more evenly distributed as the motion of the joints was less abrupt and the joints were subjected to less flexion and extension for an equivalent speed of propulsion. The brisker extension of the elbow and wrist was not needed to maintain a given speed of propulsion, which implies that the angle at which the hands contact the pushrims might present some mechanical advantage with the Backward-Low position.

In **Figure 4**, the mean angular acceleration of the elbow and wrist joints are presented for all the trials and subjects at each seating position. The angular acceleration curves of the elbow and wrist further stress the less abrupt change in the motion for the Backward-Low position. For the elbow joint, the angular deceleration (20 percent to 40 percent of the cycle) slopes were not as steep for the Backward-Low position when compared to all the other positions. This again stressed a less abrupt change in the elbow motion for this position. The peak positive and negative acceleration of the wrist during the pushing phase was found to be lower for the Backward-Low position, showing less hand acceleration. Also, less wrist acceleration could help decrease the amount of slipping of the hands on the pushrims at the beginning and end of the push phase for the Backward-Low position.

The mean angular momentum of the trunk for all the trials and subjects at each seat position

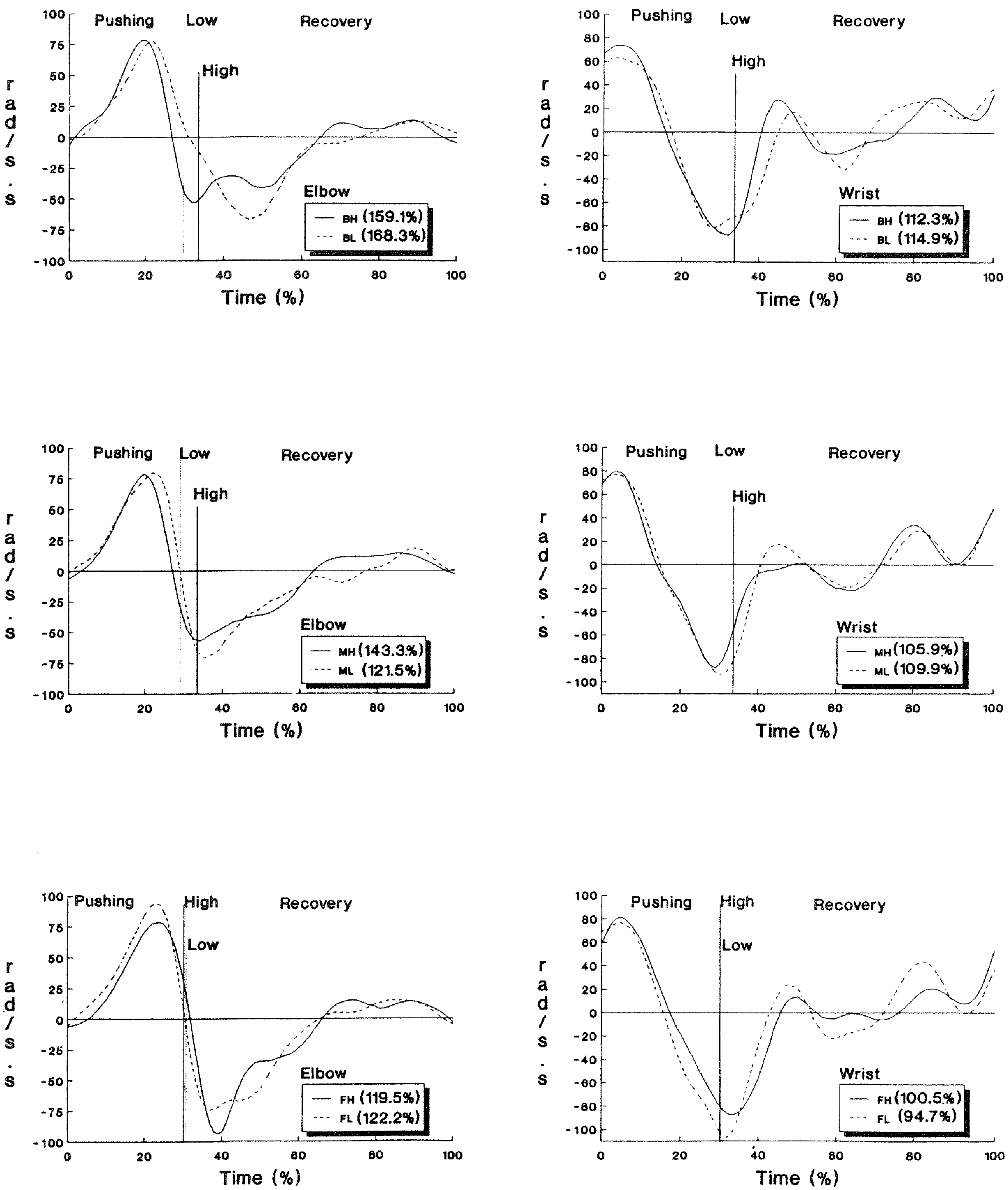

Figure 4.
Mean angular accelerations of the elbow and wrist joints for all the seat positions: Backward-Low (BL), Backward-High (BH), Middle-Low (ML), Middle-High (MH), Forward-Low (FL), Forward-High (FH). Coefficients of variations are indicated in parentheses.

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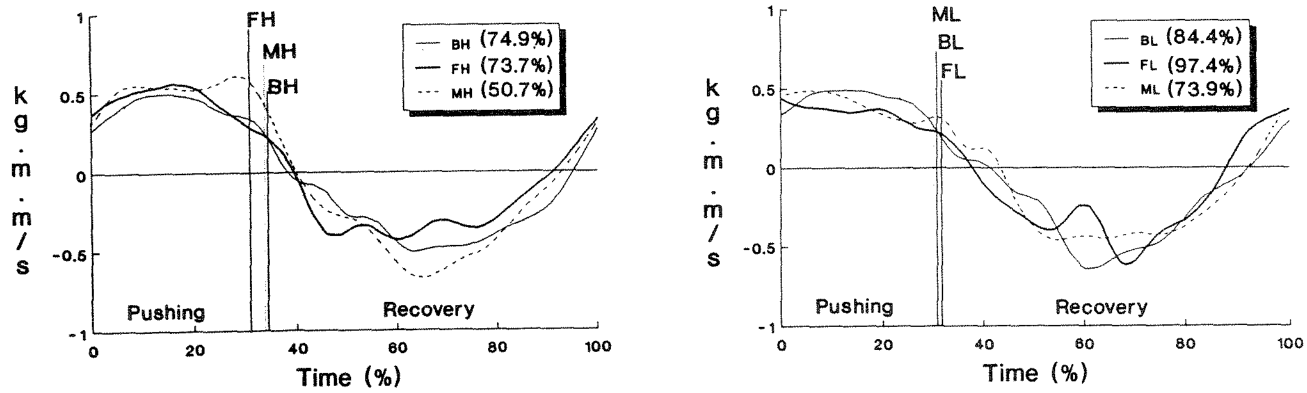

Figure 5.
Mean angular momentums of the trunk for all the seat positions: Backward-High (BH), Backward-Low (BL), Middle-High (MH), Middle-Low (ML), Forward-High (FH), and Forward-Low (FL). Coefficients of variations are indicated in parentheses.

evaluated is shown in **Figure 5**. These curves were evaluated on the basis that an increase in the mean angular momentum of the trunk during, or at the beginning of, the pushing phase would be associated with a more efficient pushing pattern. This was assumed since the motion of the trunk is believed to help increase the ability to transfer angular velocity to the pushrims (7). From this investigation, the momentum of the trunk was found to be constant during the pushing phase for all the seat positions. During the first part of the recovery phase, the trunk was brought backward and was followed by a forward motion of the trunk (which was believed to help transfer angular velocity to the pushrims). As a result of the height of the seat playing an important role in stabilizing the posture (28), it can be expected that a change in seat position would affect the motion of the trunk. However, as shown in **Figure 5**, a clear relationship does not seem to exist between a change in seat position and the trunk angular momentum. The lack of differences between the trunk angular momentum and changes in seat position can be explained in terms of the variability among and between the subjects (mean variation of approximately 75.8 percent). The high variation in trunk momentum among the subjects might have resulted from weak hip flexors and abductors associated with class IV lesions (13). Similarly, even though the abdominal muscles are said to be fully functional within class IV, a subject who is confined to a wheelchair for an extended period of time might suffer a decrease in abdominal strength. Also, the lack of difference observed might be the result of a posture compensation. The subjects were found to change their posture with higher seat positions: for example, some slouched their back in order to overcome the change in position. However, the lack of difference does not prove that a greater number of subjects would not elicit a significant relationship in trunk momentum. Therefore, further research should be conducted in this area to estimate the extent and influence on the angular momentum of the trunk on a change in seat position.

A distinct pattern of EMG activity was observed for all the seat positions. The LE EMG curves of **Figure 6** represent an ensemble average of five subjects and nine trials for all of the seat positions and muscles. For each muscle, the coefficient of variation is indicated in parentheses. The coefficient of variation was calculated according to Giroux and Lamontagne's equation (29). The LE EMG activity indicated that the biceps brachii was active during the initial part of the pushing phase (pull motion) and during the latter part of the recovery phase. Thus, the biceps served as a forearm flexor to pull during the pushing phase and was used to flex the arm at the end of the recovery phase. The triceps brachii showed a burst of activity in the latter part of the push phase (push motion) and therefore appeared to function as a forearm extensor. Thus, the biceps brachii and the triceps brachii activity showed that a pull-push type of motion was used during the pushing phase. This stroke pattern was also observed by Mâsse and Lamontagne (30); Ross and Brubaker (10); and Steadward (31). The pull motion started when the hands initially contacted

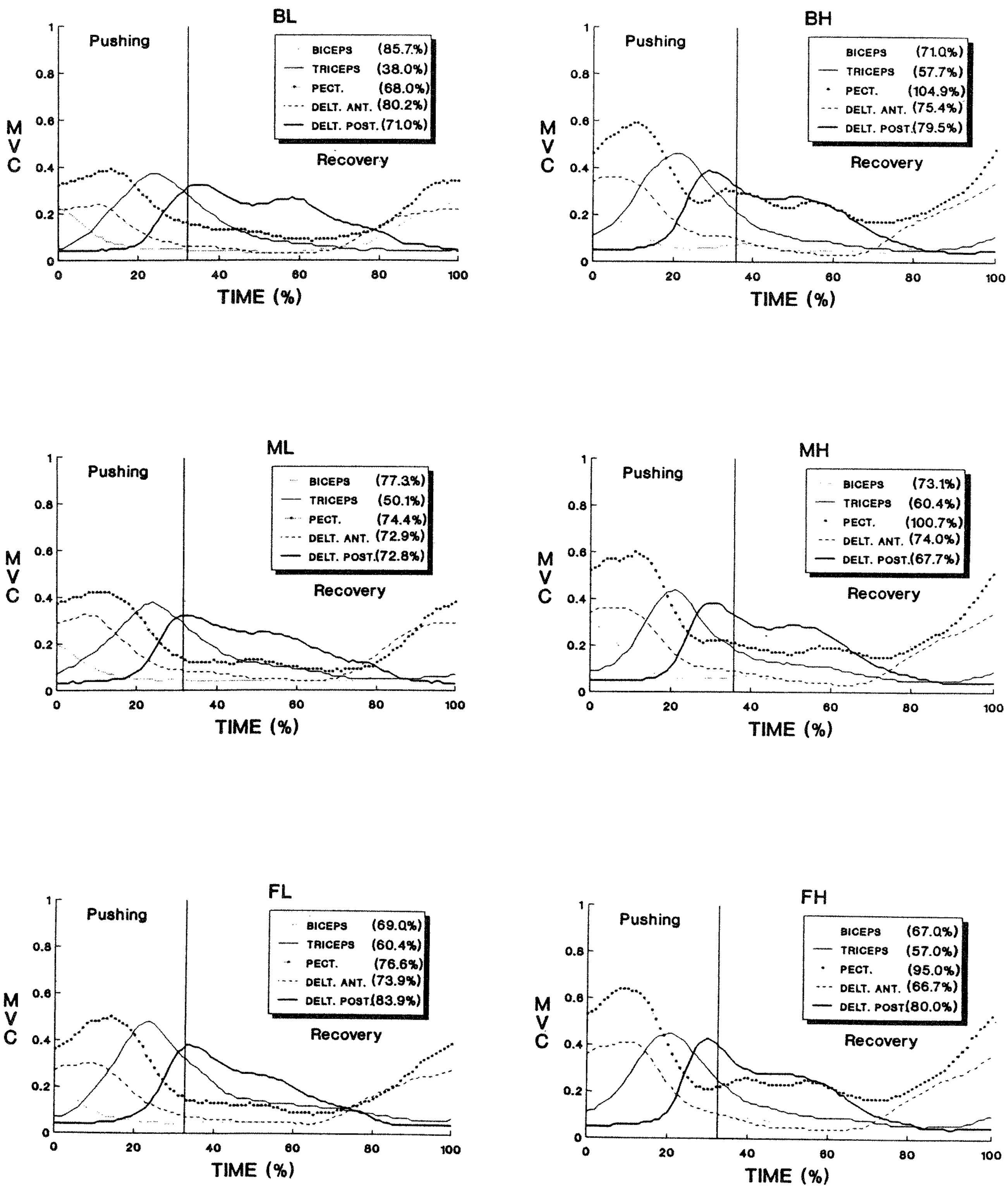

Figure 6.
Mean linear envelopes for the biceps brachii, triceps brachii, pectoralis major, deltoid anterior, and deltoid posterior muscles for all the seat positions: Backward-Low (BL), Backward-High (BH), Middle-Low (ML), Middle-High (MH), Forward-Low (FL), Forward-High (FH). Coefficients of variations are indicated in parentheses.

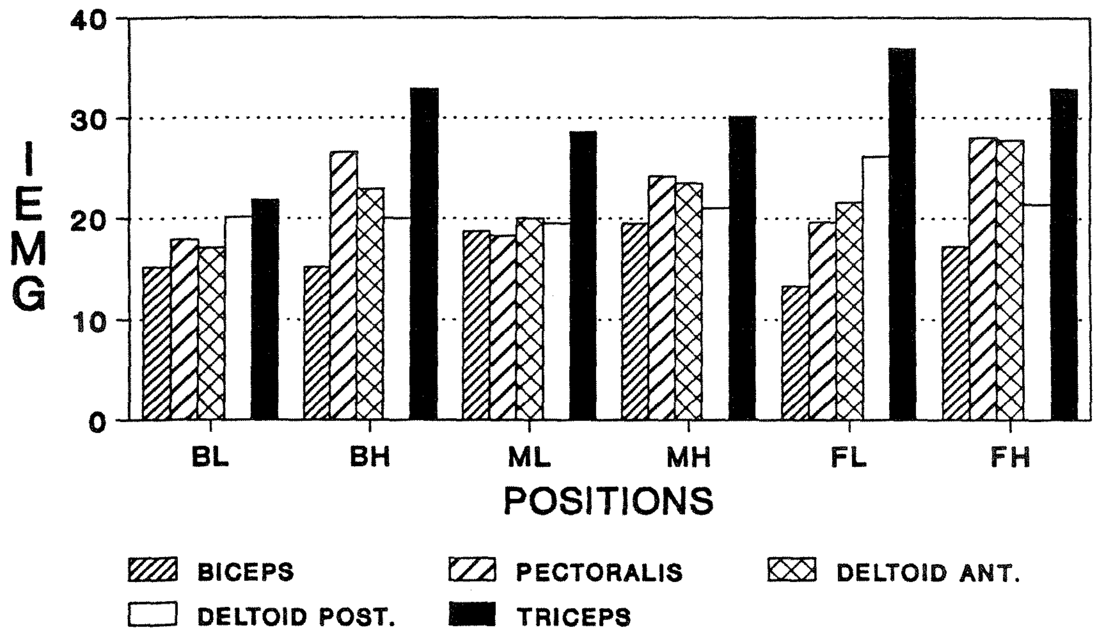

Figure 7.
Mean IEMG of each muscle for each seating position: Backward-Low (BL), Backward-High (BH), Middle-Low (ML), Middle-High (MH), Forward-Low (FL), and Forward-High (FH).

the pushrims and carried on until they reached the top of the pushrims. The pushing motion started as the hands crossed the top of the pushrims and finished at the end of the pushing phase.

The flexion and extension of the arm and forearm used for propulsion are shown by the activity of the pectoralis major, the deltoid anterior, and the deltoid posterior. The pectoralis major and the deltoid anterior activity were similar (both being active during the pushing phase), and a burst of activation was observed in the latter part of the recovery phase. The pectoralis major and the deltoid anterior muscles appeared to function as arm flexors during the pushing phase and at the end of the recovery phase. The pectoral muscle also may have been used as a stabilizer for medial rotation which might have occurred at the end of the pushing phase and during the recovery phase. Finally, the deltoid posterior activity was observed in the latter part of the pushing phase and in the beginning of the recovery phase. Therefore, it probably functioned as a stabilizer for the lateral rotation and as an arm extensor. The observed patterns of LE EMG recruitment were similar to those obtained by Ross and Brubaker (10) and Veeger, Van der Woude, and Rozendal (32).

Although the patterns of LE EMG activity were similar at all of the seat positions evaluated, the amplitude of LE EMG activity varied with a change in seat position as shown by **Figure 6**. The mean IEMG for all the trials and subjects at each seating position are illustrated in **Figure 7**. More variations in the IEMG were observed for the triceps brachii, the pectoralis major, and the deltoid anterior muscles with a change in seat position. These muscles were more affected by a change in seat position since they are thought to be responsible for the pushing phase. This would be in agreement with Cooper (4) who stated that the deltoid anterior, the pectoralis, and the triceps muscles are the prime movers for wheelchair racing. Cooper based his statement on the work of Taylor, McDonnel, Roger, Loiselle, Lusch, and Steadward (33) who observed hypertrophied lateral triceps for track and field wheelchair athletes, and Tesch and Karlsson (34) who demonstrated hypertrophied middle deltoid in wheelchair basketball athletes. The biceps brachii was also observed to be active during the pushing phase; however, since it was minimally recruited (approximately 30 percent of MVC), it did not significantly contribute to the motion and therefore was not greatly affected by a change in seat position. Harburn and Spaulding (20) observed that the muscles most active for pushrim propulsion were the middle deltoid, posterior deltoid, and, in some subjects, the triceps brachii. However, we observed that the triceps brachii, the pectoralis major, and the deltoid anterior and posterior were used for propulsion, which differed from the results of Harburn and Spaulding. Since the shoulder muscle complex offers greater range of motion than any other articulations, it offers greater abilities to compensate (35). This may explain some of the differences we observed.

The qualitative analyses of **Figure 6** and **Figure 7** clearly indicate that less EMG activity was found with lower seat positions, mainly for the Backward-Low and the Middle-Low positions. Therefore, these positions are expected to require less energy. This would be in agreement with Bigland-Ritchie and Woods (25), and Henriksson and Bonde-Petersen (26) who reported a linearity between the IEMG and the rate of oxygen uptake during cycle ergometry. The lower IEMG recorded for the lower seat positions was thought to be caused by greater degrees of contact with the pushrim, since it was previously observed that a greater degree of contact was found with lower seat positions (see **Table 3**). Also, a better orientation of the segments on the pushrim during wheelchair propulsion, as well as the degrees of contact and the pushing time, may have affected the magnitude of the IEMG recorded. Van der Woude, *et al.* (5) found that a seat height corresponding to an elbow range of 100 to 120 degrees appeared to be most efficient under their conditions studied. However, they stated that they are not able to indicate if the increase in cardiorespiratory response originated from the change in muscle length. Although the overall amplitude in IEMG was lower for the Backward-Low position, it should be noted that the Middle-Low position would offer more stability than the Backward-Low position during wheelchair racing, because the longer the distance between the CM and the main axles, the greater the resistive moment arm will be (17). Therefore, if the CM is closer to the main axles the wheelchair stability is less (36).

The lower overall IEMG activity observed for the Backward-Low positions would indicate that in moving the CM, the amount of force applied to the pushrim was affected, since moving the seat backward would cause a decrease in the rolling resistance ratio (the ratio of the weight distribution of the back roller to the front roller). In real-life conditions, a similar decrease in rolling resistance ratio is found between the rear wheels and the front casters with a Backward seat position. This would be in agreement with Brubaker who found that moving the CM closer to the axle of the rear wheels resulted in a decrease in rolling resistance (1). Therefore, lower IEMG activity should occur at the Backward positions since less force is required to propel the chair. However, the Backward-High position had slightly higher IEMG activity than did the Backward-Low position, which indicated that the rolling resistance ratio was not the only important factor which affected the IEMG activity. The greater IEMG recorded for the Backward-High position in comparison to the Backward-Low position would result from poor contact (location and orientation of the hands) being made with the pushrims, mainly because the orientation and location with which the hands contact the pushrims differ. This would agree with Sanderson and Sommer (7).

Table 4 depicts the ranking of the biomechanical samples across the seating positions. The results of the Friedman two-way ANOVA on the ranks revealed that a $Fr = 21.48$ when $df = 5$ was significant at $p \leq 0.05$, therefore indicating the rejection of the null hypothesis. One may then conclude that the sum of the ranks of the biomechanical samples significantly differed across the seating positions. To locate this difference a *post hoc* analysis using multiple comparisons between seating positions was computed. The multiple comparisons revealed that the Backward-Low and the Middle-Low positions were significantly different from all the High positions and that all the other pairwise comparisons were not significant. Although the trend indicated that the Backward- and Middle-Low positions were slightly better than all the other seating positions (see **Table 4**), the statistical evidence did not find the Forward-Low position to be significantly different from the Backward- and Middle-Low positions. It should again be stressed that the validity of the statistical results are limited to the assumption that the biomechanical parameters are assumed to be a "biomechanical sample" on which the seating positions may be contrasted. This assumption implies that the criteria used to contrast the positions are distinct. In light of this assumption, the statistical results should be interpreted with

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Table 4.
Ranking of the biomechanical samples across the seating positions.

	Positions					
Biomechanical samples	**BL**	**ML**	**FL**	**BH**	**MH**	**FH**
Biceps m.	2	5	1	3	6	4
Pectoralis m.	1	2	3	5	4	6
Triceps m.	1	2	6	4.5	3	4.5
Deltoid anterior m.	1	2	3	4	5	6
Deltoid posterior m.	3	1	6	2	4	5
Pushing frequency	2	1	3	6	4	5
Recovery phase	2	1	3	6	4.5	4.5
Recovery phase	2	1	4	6	4	4
Sum of rank	14	15	29	36.5	34.5	39

BL = Backward-Low
ML = Middle-Low
FL = Forward-Low
BH = Backward-High
MH = Middle-High
FH = Forward-High

caution. It may then be concluded that the Backward-Low and the Middle-Low positions were slightly better seating positions, especially when compared with the High positions. It should be noted that the proposed conclusions were based on a small number of subjects, and that the high amount of variation found among the subjects warrants that these results be replicated to ensure the generalizability of the trends observed in this study to other groups. These conclusions would, however, be in agreement with Burk (37), Schuman (38), and Walsh (39). Walsh specified that the seat should be slightly behind the hip bone in order to align the shoulder with the main axle, which he believed would allow the forces to be applied directly downward. According to Higgs (9), a Backward-Low seat position was also found to be related to greater success for long-distance racers. However, some discrepancies exist with the findings of Brubaker, *et al.* (8) and Middle-Middle and Middle-Forward seat positions for the pushrim propulsion were observed to be correlated with higher efficiencies. The differences observed with our results are not difficult to explain since the seat positions tested by Brubaker, *et al.* (8) did not correspond to our positions and the subjects used for their study were able-bodied. Also, Walsh, *et al.* (2) did not observe any relationship between seat position and wheelchair velocity, but this is not to say that the kinematic or the EMG activity would not differ with a change in seat position, as observed in our study.

CONCLUSIONS

This investigation showed that with lower seat positions, less EMG activity was recorded than for higher seat positions. For the lower seat positions, the kinematics of the upper limb joints were observed to have a smoother motion since the upper limb joints were moved in sequence (wrist, shoulder, and elbow). Among the positions evaluated, the Backward-Low position had the overall lowest IEMG and the elbow and forearm acceleration slopes were less abrupt than with the other seat positions, thus indicating a smoother motion. The Middle-Low position was found to have the lowest pushing frequency. The location and orientation with which the hands contacted the pushrims was found to be very important, since less IEMG was recorded and greater degrees of contact were observed for the Low positions. Furthermore, the overall IEMG was found to be lower for the Backward-Low position. Based on these descriptive observations and the statistical analysis, it was concluded that the Middle-Low and the Backward-Low positions would be slightly better seating positions. A change in seat position was not found to affect the angular momentum of the trunk. This nonrelationship was explained by the variability among the subjects' technique of propulsion. Weak hip flexors are associated with class IV, and there is a possibility that some subjects adjusted their posture with a change of seat position. The kinematic and EMG analysis of wheelchair propulsion at different seat positions has provided useful information to enhance our understanding and development of wheelchair design. However, further studies should be conducted in this area to confirm our observations and to provide more information about the ideal seating position.

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