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Kinematic and Electromyographic Analysis of Wheelchair Propulsion for Various Seating Positions

by
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Dedicated

To my parents and Jeff

Who have encouraged me and

provided me with endless support

Much love and thanks

Preface

This thesis was written in article format style, which consists of an article and a technical note. In the appendices the traditionnal three chapters are joined as well as a copy of the consent forms and a list of anthropometric measurements.

Acknowledgements

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Finally, a special thank you is extended to the subjects without whom this investigation would not have been possible.

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Abstract

The pattern of propulsion for five male paraplegics was investigated for six seating positions, consisting of a combination of three horizontal rear wheel positions at two seating heights. To simulate wheelchair propulsion in the laboratory, the wheelchair was mounted on high rotational inertia rollers. For three trials at each seating position, the subject propelled the designed wheelchair at 60% of their maximal speed which was determined at the beginning of the test session. At each trial, the subject's propulsion technique 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 linear and 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 peak linear acceleration of the hand at the end of the recovery phase was lower, thus facilitating the contact of the hands on the pushrims at the point of grabbing since lower acceleration would reduce slippage of the hands on the pushrims. Also, the forearm linear velocity slopes and the elbow angular velocity slopes were 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 a posture compensation.

Key Words: Wheelchair design, Paraplegics, Propulsion pattern, seat adjustment, kinematics, Electromyography, and wheelchair propulsion.

Kinematic an Electromyograpphic Analysis of Wheelchair Propulsion for Various Seating Positions

In the past decade we have witnessed tremendous improvements in wheelchair racing records (Brubaker, 1986). 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 1940's, 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 (Walsh, Marchiori, & Steadward, 1986). The athletes themselves are responsible for most of the improvements observed in wheelchair design. They have slowly modified the conventional wheelchair into a streamlined light-weight racing machine according to their individual needs. Despite the changes that have occurred over the past 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 athlete's performance including: the strength of upper body, the physical capacity of the individual, the level of neurological lesion, and the interaction between the user and the wheelchair (Brubaker & McLaurin, 1982; van der Woude, Veeger, & Rozendal, 1986). Wheelchair design plays an important role in optimizing the athlete's performance, since different seat positions alter the athlete's pattern of propulsion and consequently affect the performance. One of the most prevalent problems facing the athletes today is to define the ideal seat position needed to achieve an optimal propulsion technique (Walsh et al., 1986). Therefore, research in wheelchair design may enable athletes to improve their performance through a better technique of propulsion.

The relationship between seating position and the pattern of propulsion has not yet been investigated, however, some authors have looked at the effect of varying seating positions on performance. Brubaker, McLaurin, and Gibson (1980) studied the effect of varying the seat position on the mechanical efficiency and they observed that both a middle-middle and a middle-forward seat position were found to have higher mechanical efficiency. As well, lower pushing frequency and smaller energy expenditure were associated with higher mechanical efficiency. However, their do not agree with those of Higgs' (1983) and Walsh et al. (1986). Higgs (1983) static analysis of wheelchair racing used at the 1980 Olympic games for the disabled revealed that a low-backward seat position was found to be highly correlated with success for long distance athletes. Walsh et al. (1986) studied the effect of seat position on the maximal linear velocity of wheelchair sprinting and found that no significant differences existed between the maximal linear velocities with a change in seat position. Since the subjects' levels of neurological lesion were not taken into account by Higgs (1983) and Walsh et al. (1986) and because a small number of subjects were used for these investigations (Brubaker et al., 1980 & Walsh et al., 1986), this might explain the discrepancies observed among these studies. According to Steadward (1979) subjects' having various level of lesions were observed to use different techniques of propulsion and were found to have a different electromyographic recruitment patterns. Therefore, the neurological level of lesion and the anthropometric variability of the subjects as well as a change in seat position can alter the technique of propulsion and consequently the performance.

Muscle response is also influenced by the position of the user in relation to the pushrim, the resistive forces, and the level of disability (Ross & Brubaker, 1984). Brubaker, McLaurin, and McClay (1985) related the seat position with the electromyographic (EMG) activity and the efficiency using lever arm propulsion. Their results indicated that the middle-middle and the middle-backward seat position had an overall lower EMG activity, which was reflected in higher efficiency. However, the lever propulsion does not truly represent the real pattern of propulsion, therefore, further research should be conducted using the pushrim type of propulsion.

At the present time, doubt still remains as to the identification of the ideal seating position. Therefore, further research should be conducted in this area to facilitate the process of optimizing wheelchair design. Hence, the purpose of this investigation was to examine the influence of seating position changes on the kinematic and electromyographic parameters of the upper limbs during wheelchair propulsion. The pushrim propulsion technique was investigated for six seat positions which consisted of a combination of three horizontal rear wheel positions at two sitting heights at 60% of the subject's maximal speed of propulsion.

Methods

Five male paraplegics served as subjects for this investigation. Table 1 provides a summary of information about each subject.

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
S.D.		7.5	13.6			1.68	4.31

*The neurological lesion at the spinal cord

The subjects were selected from the Ottawa region on a voluntary basis and were free of any known pathological disorders of the upper extremities. Also the subject level of physical fitness was as a criteria for the selection in order to avoid large variation among the subjects. The subjects used in this investigation were all physically active: they were either involved into wheelchair racing. wheelchair basketball. or activelly play sleigh hockey.

Apparatus

All subjects were tested on an adjustable wheelchair built by Advance Mobility System Corporation according to our specifications. The following adjustments were possible with this chair: the seat height. the horizontal position of the rear wheel; the wheel camber; and the seat base and back rest inclination (for further details about the testing wheelchair refer to the technical note). The subjects were evaluated at six experimental conditions consisting of three horizontal positions of the rear wheels (Forward. Middle. and Backward) at two seat heights (High and Low). The horizontal positions of the rear wheels consisted of moving the centre of mass (C.M.) of the wheelchair horizontally. A modified reaction board technique (Lemaire. Lamontagne & Barclay. 1989) was used to determine the C.M. of the wheelchair. For the Forward. Middle. and Backward position of the rear wheels. the C.M. was located at 10 cm. 6 cm. and 3 cm in front of the main axles. respectively. This also corresponded to the seat base and back rest intersection being located at 7.6 cm (Forward). 4.4 cm (Middle). and 1.2 cm (Backward) behind the main axles. These selected positions were believed to represent a range between the conventional and racing wheelchair which was reported by Peizer. Wright. and Freiberger (1964). The High and Low positions were established by moving the seat. Both positions depended on the subject's arm and trunk lengths. 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 the subjects it was found to closely correspond to the desired position (distal phalanges of the second fingers were aligned with the lowest portion of the pushrims). The High position was set at 10% of the subject's arm length above the Low position. The Low and High positions were selected because they represented a range of positions which were used by wheelchair athletes (Higgs, 1983; York & Kimura, 1987). For each experimental condition the wheel camber, the seat base, and the back rest angle remained constant. These positions were determined according to Higgs (1983); York and Kimura (1987) 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 was set at 12.5 degrees from the horizontal; and the back rest was set at 90 degrees from the seat base.

To simulate wheelchair propulsion in the laboratory, the testing wheelchair was mounted on rollers. Due to the hollow characteristic of the rollers and to better simulate actual locomotion each roller mass was modified by adding two iron rings to increase the system's rotational inertia. To monitor the speed of propulsion a tachometer was mounted on the side of the rollers (for a more detail description of the rollers refer to the technical note).

Procedures

Prior to testing, informed consent (see Appendix B) was obtained from each subject. Markers were placed on the appropriate joint centre locations to facilitate the ensuing kinematic data reduction of the upper body segments. The subject anthropometric measurements of the upper body (see Table 1 and Appendix A) were recorded according to Jette's procedure (1983). Then, the Low sitting position was measured with the shoulder aligned with the main axles. The subject was asked to keep his back against the back rest and to keep his head straight. The arms were lying against the pushrim. The seat was then moved 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 because they are believed to be mainly recruited during wheelchair propulsion (Harburn & Spaulding, 1986; Ross & Brubaker, 1984). Disposable silver/silver chloride surface electrodes (Medi-trace) were placed over the motor point of each muscle as described by Delagi, Perotto, Iazzetti and Morrison (1975). 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 kilohms 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. For each subject the maximal speed average was calculated and 60% of this speed was used in the remainder of the experiment (see Table 2).

Table 2

Speed of Propulsion

Subject	Max	(S.D.)	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
S.D.	0.35		0.19

The maximal oxygen uptake was not used to set the intensity of exercise because paraplegics physiological capacity differ from normal person (Coutts, Rhodes & McKenzie, 1983) and mainly because in order to compare among seating positions it is important that the speed remained constant. The maximum voluntary isometric contraction (MVC) for each muscle was then recorded according to Delagi et al. (1975) procedure. For each muscle, the subject was asked to perform a maximum voluntary isometric contraction for a brief period of 5 s and this was recorded with a data acquisition system (BIOAD system, Lamontagne, Bradley & Lemaire 1989) and the data were stored on microcomputer (Compaq 386, 16 MHz). This procedure was performed before and after the testing of each experimental condition.

Finally, the subject was asked to propel the designed wheelchair at each experimental condition FH (Forward and High), FL (Forward and Low), MH (Middle and High), ML (Middle and Low), BH (Backward and High), and BL (Backward and Low) at 60% 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 s. 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 metres away) for one cycle. Simultaneously the muscular activity was recorded for at least three cycles at each trial. The raw Electromyographic (EMG) signals were recorded at 1000 Hz for a period of 5 s for each trial. The signals were fed to a bioamplifiier (University of Ottawa, input impedance of 10 megohms, 10-700 Hz bandpass), digitally converted by a data acquisition system (Lamontagne et al. 1989), and then stored in the memory of a microcomputer (Compaq 386, 16 mHz) using the BIOAD system (Lamontagne et al., 1989). In between trials the subjects had approximately 2 min of rest.

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 where an increase in 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 contacts the pushrim, the main axle of the rear wheels, and the point at which the hand releases the pushrim. The pushing frequency was defined as the number of strokes per minute.

All the EMG data were processed on a microcomputer (Compaq 386, 16 MHz) with the BIOPROC program (Lamontagne, Bradley & Lemaire 1989). The raw EMG signal bias was removed by mean and then the signal was full-wave rectified and filtered at 6 Hz (single pass critically dampened digital filter) to obtain the linear envelope (LE EMG). For each cycle, LE EMG signal for each muscle was normalized over time (%) and was 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 then transferred to a mainframe computer (Amdahl IBM VM/SP, University of Ottawa) for complete data processing. The filmed data were projected at 9.3% of the life-size image and were 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 linear and angular, velocities, accelerations, and momentums were calculated for a complete wheelchair cycle by processing the digitized cinefilm coordinates with the BIOMECH package (University of Ottawa). Then, for each condition and subject, the data were normalized over time 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 (n=15, 3 trials per subject times 5 subjects) for each condition.

Both descriptive analyses and descriptive statistics were performed on the EMG and the 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. Then, these integrated EMG (IEMG) were ensemble averaged across subjects (n=5) to give a grand ensemble normalized IEMG (n=45, 9 trials per subject times 5 subjects) 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 averaged (n=15, 3 trials per subject times 5 subjects) was computed on each of the above kinematic parameters for each condition. Also, a visual inspection of the averaged linear and angular velocities, accelerations, and momentums, and LE EMG curves were used to further contrast the seating positions.

Results and discussion

The mean (n=15, 3 trials and 5 subjects) speed of the rear wheels, pushing frequency, and degree of contact for each seating position are presented in Table 3. Figure 1 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 seat position caused an increase in cycle time, recovery time, and a decrease in pushing frequency for the Backward and Middle positions while no variation was found among these parameters for the Forward positions (see Table 3 and Figure 1). Also, the percentage of time spent in each of the phases of propulsion (pushing and recovery phases) did not differ with a change in seat position (see Figure 1). This implied that the Backward and Middle Low positions would not be as tiring as their corresponding High positions, since the subjects had to stroke less often and spent more time recovery. However, for the Forward positions the level of exertion should be similar since there were no differences observed in the cycle time and the pushing frequency. This would agree with 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)	$\bar{X}$	3.72	3.65	3.62	3.63	3.66	3.58
	S.D.	(0.26)	(0.28)	(0.29)	(0.28)	(0.29)	(03.0)
Pushing freq. (Strokes/min)	$\bar{X}$	77.6	73.5	77.9	82.6	78.6	78.9
	S.D.	(23.5)	(26.2)	(27.2)	(32.0)	(34.0)	(28.8)
Degree of contact (Degrees)	$\bar{X}$	108.5	108.6	111.0	97.1	100.	95.7
	S.D.	(14.8)	(12.1)	(15.1)	(11.6)	(11.6)	(13.5)

findings of Brubaker et al. (1980) who stated that smaller energy expenditures were found with a low pushing frequency. Furthermore, it was seen that greater degree of contact and cycle time resulted in slower pushing frequencies for the Backward and Middle Low position in comparison to their corresponding High positions. Therefore, a change in seat position might affect the speed since an inverse relationship was found between the pushing frequency with the cycle time and the degree of contact. However, Walsh et al. (1986) did not find any significant difference in speed with a change in seat position, which might have resulted from the high variability among the subject's level of lesion which was used in their study.

In moving the seat down the proportion of the pushrim accessible for pushing was lengthened since greater degree of contact were found with lower seat positions (see Table 3). This was in agreement with the results of Brubaker et al. (1980). However, Brubaker et al (1980) found that the Middle positions had longer degree of contact than the other posittions whereas this only occured for the high positions in our study, while for 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 was varied and this 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 through varying the angle at which the forces were applied to the pushrims. This would be in agreement with Brubaker and McLaurin (1982) and Sanderson and Sommer (1985).

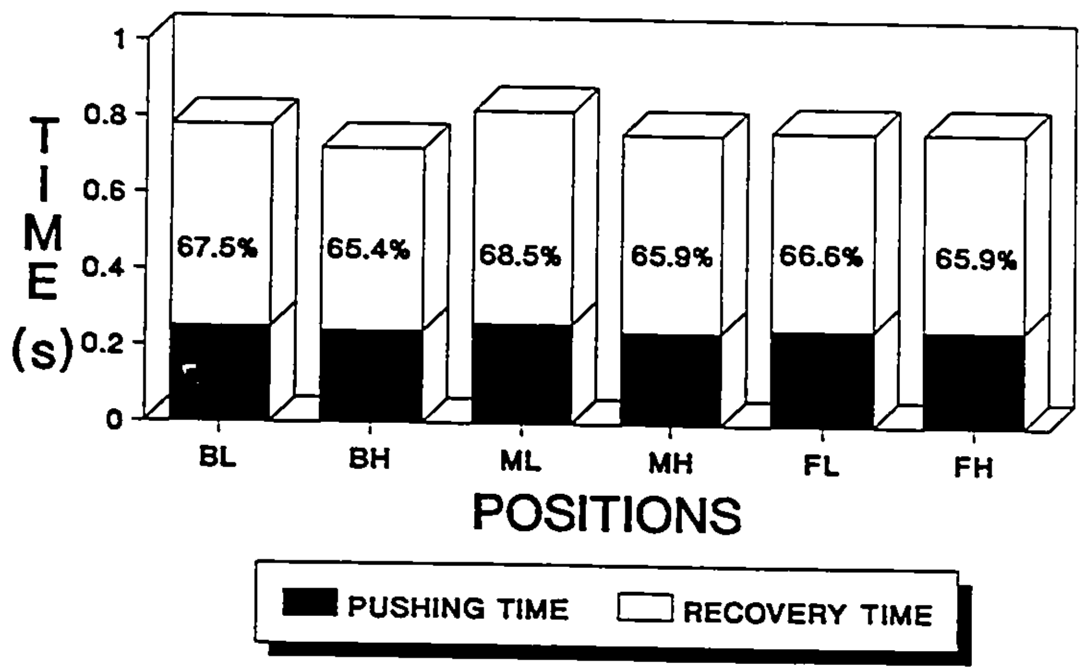

Figure 1: Mean cycle time, pushing time, and recovery time for all the seat positions: Backward-Low (BL), Backward-High (BH), Middle-Low (ML), Middle-High (MH), Forward-Low (FL), and Forward-High (FH). The percentage of time spent in recovery (recovery phase) is also indicated.

Figure 2 presents the mean linear velocities of the upper limbs for all trials and subjects at each seating position. The peak positive forearm linear velocity for the Backward and Middle Low positions was observed to occur near the end of the pushing phase, while the peak forearm linear velocity for the Backward and Middle High positions occured before the Backward-Low and the Middle-Low positions. Thus, the timing between the forearm motion and the end of the pushing phase appeared to be better for the Backward and Middle Low positions than their corresponding High positions. However, for the Forward positions such a difference was not observed. Also, it was observed that for all the High positions the backward change of direction (approximately 45% of the cycle time) for the upper limb segments occured at the same time. However, for the Low positions the hand was brought backward first followed by the forearm and upper arm segments. Thus, for the Low positions the upper limb segments were found to have a more continuous motion than for the High positions since the upper limbs backward motion occured in a sequence (hand, forearm, and upper arm).

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 positions). It was also observed that at all the Low positions, the upper limb joints change of motion occured in a sequence (approximately 45% 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 shoulder and elbow joints change of motion occured at the same time (approximately 40% to 45% of the cycle). Consequently, the motion of the upper limb joints were flowing more smoothly for the Low positions since the joints change of motion were more sequential in comparison to the High positions.

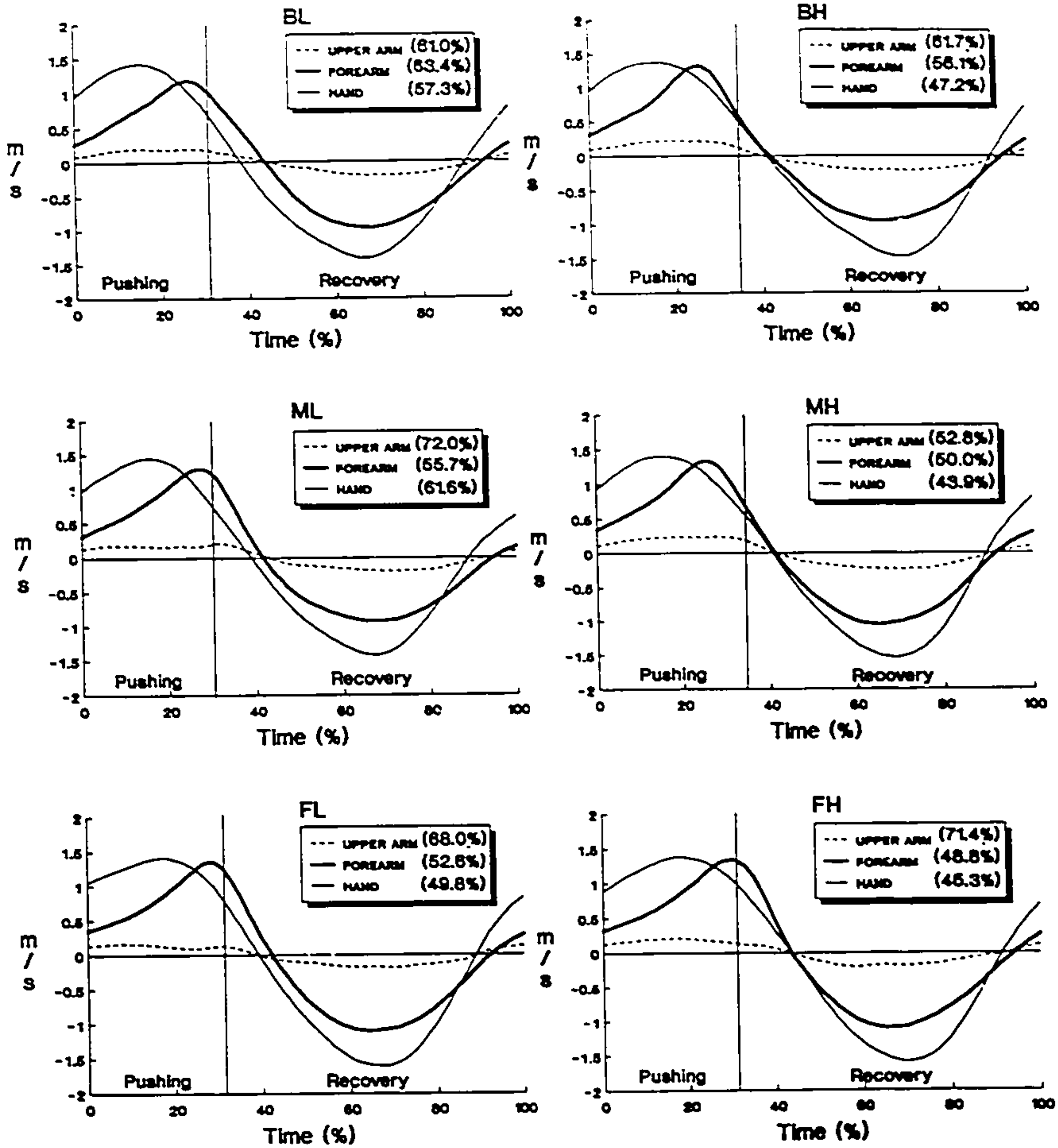

Figure 2: Mean linear velocities of the upper limb segments for all the seat positions: Backward-Low (BL), Backward-High (BH), Middle-Low (ML), Middle-High (MH), Forward-Low (FL), Forward-High (FH). The coefficients of variation are indicated in parentheses.

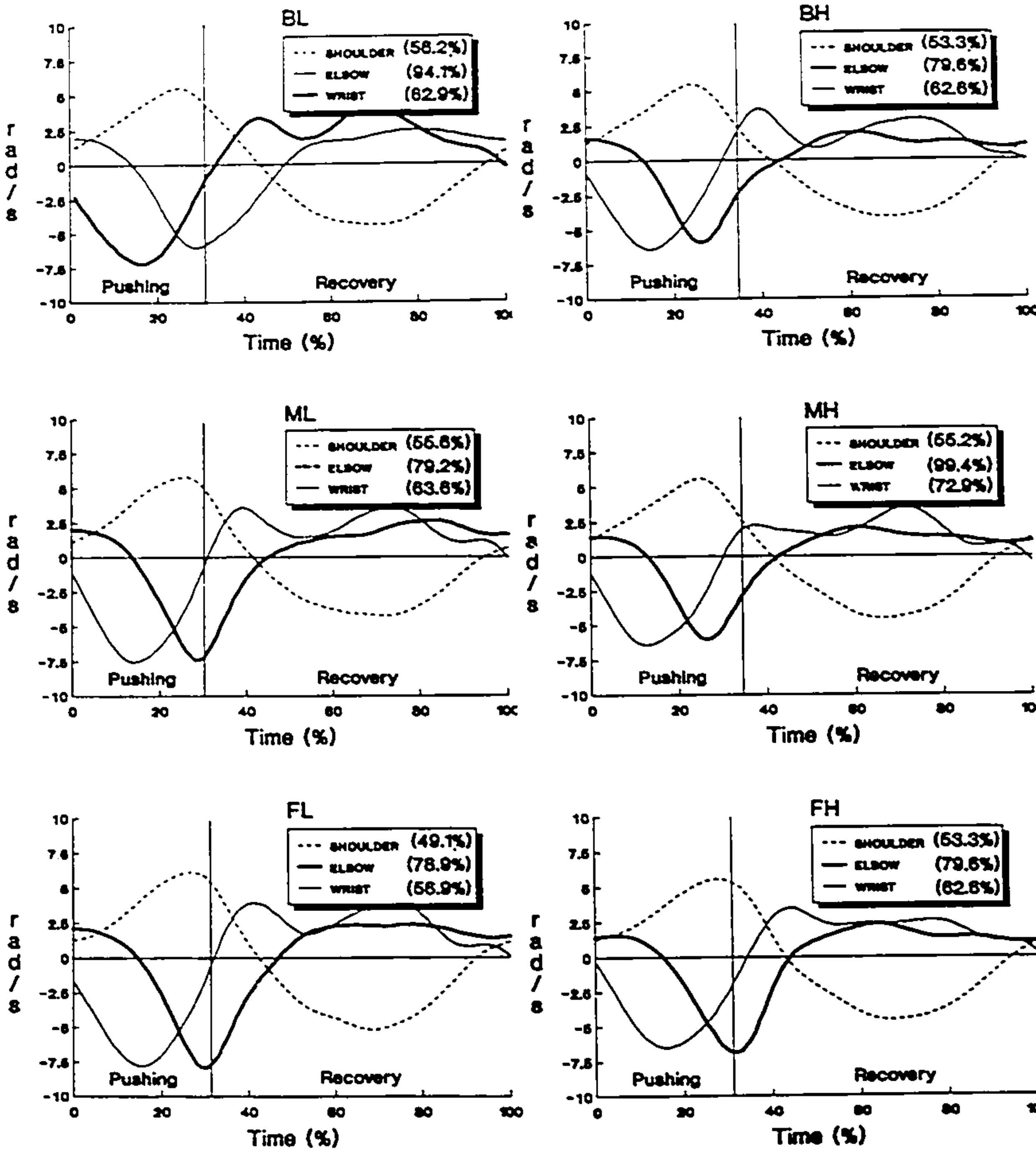

Figure 3: Mean angular velocities of the upper limb joints for all the seat positions: Backward-Low (BL), Backward-High (BH), Middle-Low (ML), Middle-High (MH), Forward-Low (FL), Forward-High (FH). The coefficients of variation are indicated in parentheses.

Somes 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 since lower seat positions allow greater extension of the elbow and wrist joints. This occurred since with higher seat positions the elbow and wrist assumed a more extended position to reach the top of the pushrims than for lower positions since 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 and allow the subject to spend more time in recovery and consequently decreased 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 the other Low positions, mainly for the elbow joint. Also, the Backward-Low position change of motion of the elbow was less abrupt than the other positions tested as shown by the smoothness of the flexion and extension slopes (see Figure 3). It was found that the lower peak angular velocities observed for the Backward-Low position did not translate into a greater pushing frequency like the High positions. Therefore, at this position (Backward-Low) the work during the pushing phase might be more evenly distributed since the motion of the joints were 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

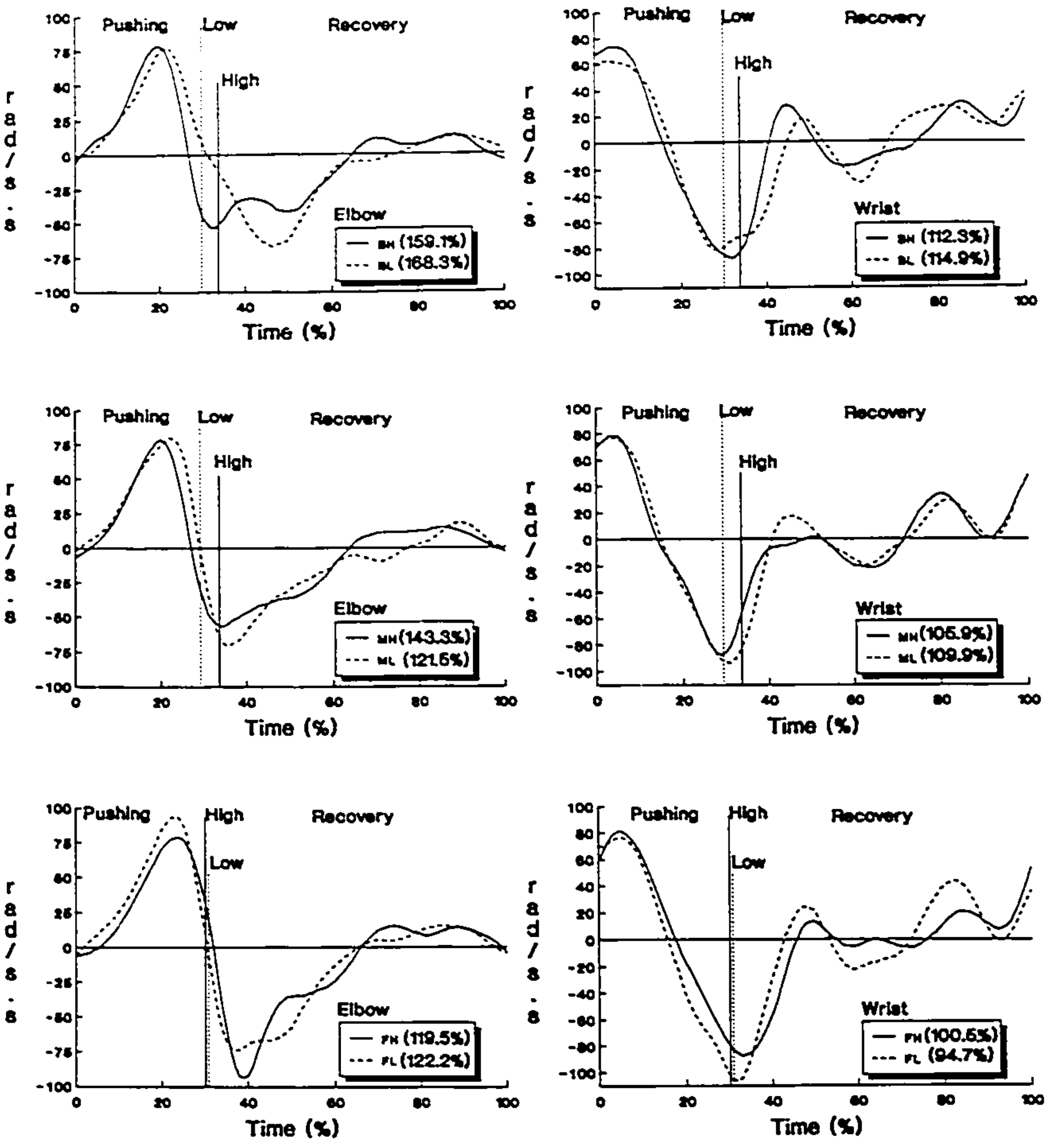

Figure 4: mean angular accelerations of the elbow and wrist joints for all the seat positions: Backward-Low (BL), Backward-High (BH), Middle-Low (ML), Middle-High (MH), Forward-Low (FL), Forward-High (FH). The coefficients of variation are indicated in parentheses.

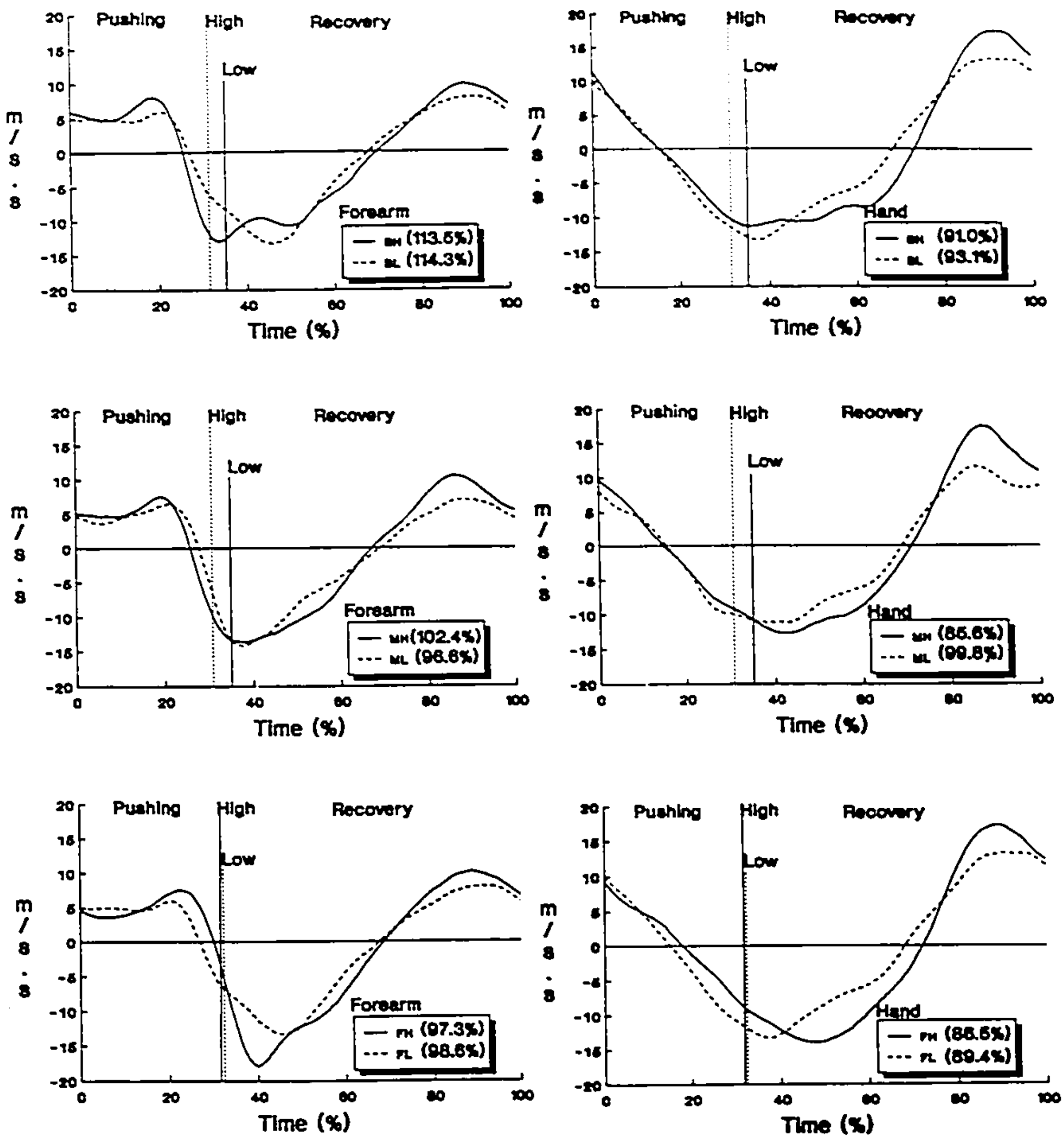

Figure 5: Mean linear accelerations of the forearm and hand segments for all the seat positions; Backward-Low (BL), Backward-High (BH), Middle-Low (ML), Middle-High (MH), Forward-Low (FL), Forward-High (FH). The coefficients of variation are indicated in parentheses.

acceleration curves of the elbow and wrist further stressed the less abrupt change in the motion for the Backward-Low position. For the elbow joint the angular deceleration (20% to 40% 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 the end of the push phase for the Backward-Low position.

The mean linear acceleration curves of the forearm and the hand for all the trials and subjects are shown in Figure 5. Some variations were observed in the linear acceleration of the forearm and the hand among the seat positions evaluated. The forearm linear deceleration slope during the pushing phase was found to be less abrupt for the Backward-Low position. This implies that the change in forearm direction was smoother (less abrupt) for the Backward-Low position in comparison to the other positions. Also the peak linear acceleration of the hand toward the end of the recovery phase (80% to 100% of the cycle) was found to be slightly lower for all the Low positions (see Figure 5). Therefore, the hand was not subjected to as much oscillation for the Low positions when compared to their corresponding High positions. Additionally, the Low positions may facilitate the contact of the hands on the pushrims at the point of grabbing since the hand was not oscillated as fast as the other positions. Too much acceleration of the hands could reduce the effectiveness of the pushing phase since there could be some slipping of the hands at the grabbing point and during the pushing phase (Sanderson and Sommer, 1985).

The motion of the trunk can help increase the ability to transfer angular velocity to the pushrims (Sanderson & Sommer, 1985). The mean angular momentum of the trunk for all the trials and subjects at each seat position evaluated is shown in Figure 6. From this investigation, the trunk angular momentum was found to be constant during the pushing phase for all the seat

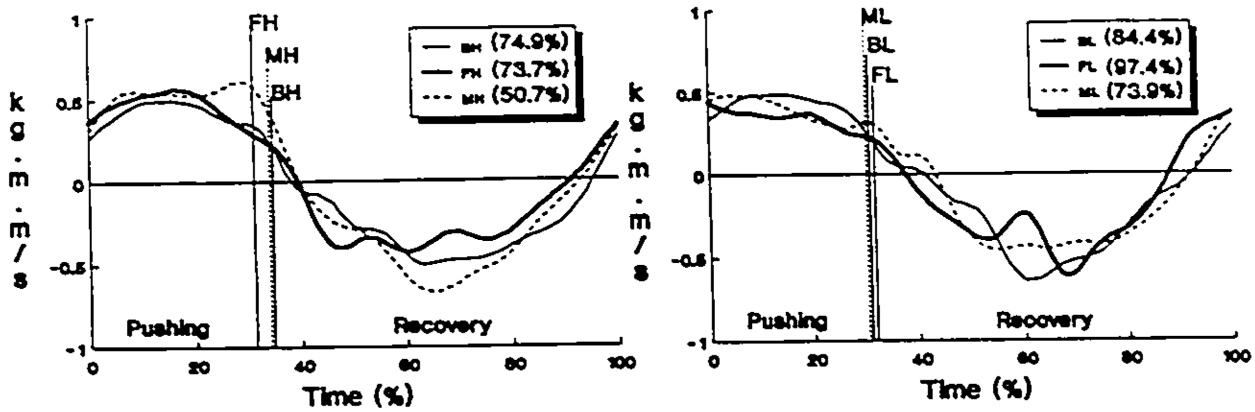

Figure 6: Mean angular momentums of the trunk for all the seat positions: Backward-High (BH), Backward-Low (BL), Middle-High (MH), Middle-Low (ML), Forward-High (FH), and Forward-Low (FL). The coefficient of variation are indicated in parentheses.

positions. Then, 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. Since the height of the seat plays an important role in stabilizing the posture (Brattgard, Lindstrom, Severinson & Wink, 1984), it can be expected that a change in seat position would affect the motion of the trunk. However, from Figure 6 there does not seem to exist a clear relationship between a change in seat position and the trunk angular momentum. The lack of difference between the trunk angular momentum and changes in seat position can be explained in terms of the variability among the subjects (mean variation of approximately 75.8%). The high variation in trunk momentum among the subjects might have resulted from weak hip flexors and abductors associated with class IV lesions (McCann, 1979). Similarly, even though the abdominal muscles are said to be fully functional within class IV, being confined in a wheelchair for an extended period of time might contribute to 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, the subjects were slouching their back in order to overcome the change in position. However, the lack of difference is not to say 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 of the trunk angular momentum with a change in seat position.

A distinct pattern of EMG activity was observed for all the seat positions (see Figures 7). The LE EMG curves of Figure 7 represent an ensemble average of 5 subjects and 9 trials for all the seat positions and muscles. For each muscles the coefficient of variation is indicated in parentheses. 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 were 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 (1989); Ross and Brubaker (1984); and Steadward (1976). The pull motion occurred when the hands initially contacted the pushrims until they reached the top of the pushrims. Then, as the hands crossed the top of the pushrims, the pushing motion started and it 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. During the pushing phase they were both active and a burst of activation was observed in the latter part of the recovery phase. The pectoralis major and the deltoid posterior muscles appeared to function as arm flexors during the pushing phase and at the end of the recovery phase. The pectoral muscle may have also been used as a stabilizer for medial rotation which may have occurred at the end of the pushing phase and during the recovery phase. Finally,

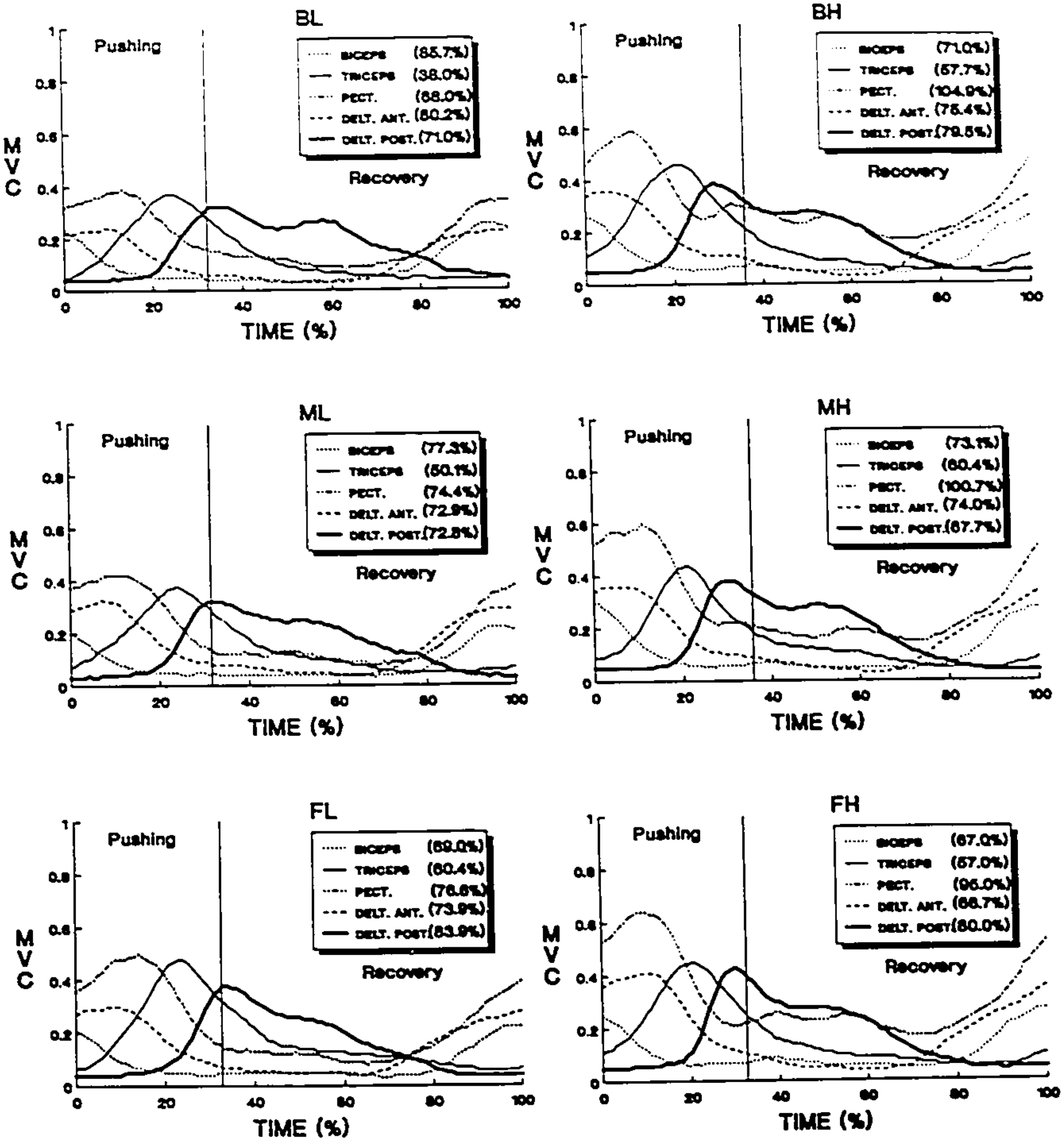

Figure 7: Mean linear envelopes for the biceps brachii, triceps brachii, pectoralis major, deltoid anterior, and deltoid posterior for all the seat positions: Backward-Low (BL), Backward-High (BH), Middle-Low (ML), Middle-High (MH), Forward-Low (FL), Forward-High (FH). The coefficients of variation are indicated in parentheses.

the deltoid posterior activity was observed in the latter part of the pushing phase and 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 (1984).

Although, the patterns of LE EMG activity were similar at all the seat positions evaluated, the amplitude of LE EMG activity varied with a change in seat position as shown by Figure 7. In Figure 8 the mean IEMG for all the trials and subjects at each seating position are illustrated. 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 there are thought to be responsible for the pushing phase. The biceps brachii was also observed to be active during the pushing phase, however since it was minimally recruited (approximately 30% of MVC) it did not significantly contribute to the motion and was therefore not greatly affected by a change in seat position. Harburn and Spaulding (1986) observed that the muscles most active for the 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, the deltoid anterior and posterior were used for propulsion, which differed from Harburn and Spaulding's (1986) results. Since the shoulder muscles complex offers greater range of motion than any other articulations, it offers greater abilities to compensate (Carlin, 1963), and this may explain some of the differences we observed with Harburn and Spaulding's studies.

Since a large amount of variation exists among the subjects it becomes impossible to assess if the differences observed among the seating positions are significant. However, Figure 7 and 8 clearly indicates that less IEMG 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 (1974); Henriksson and

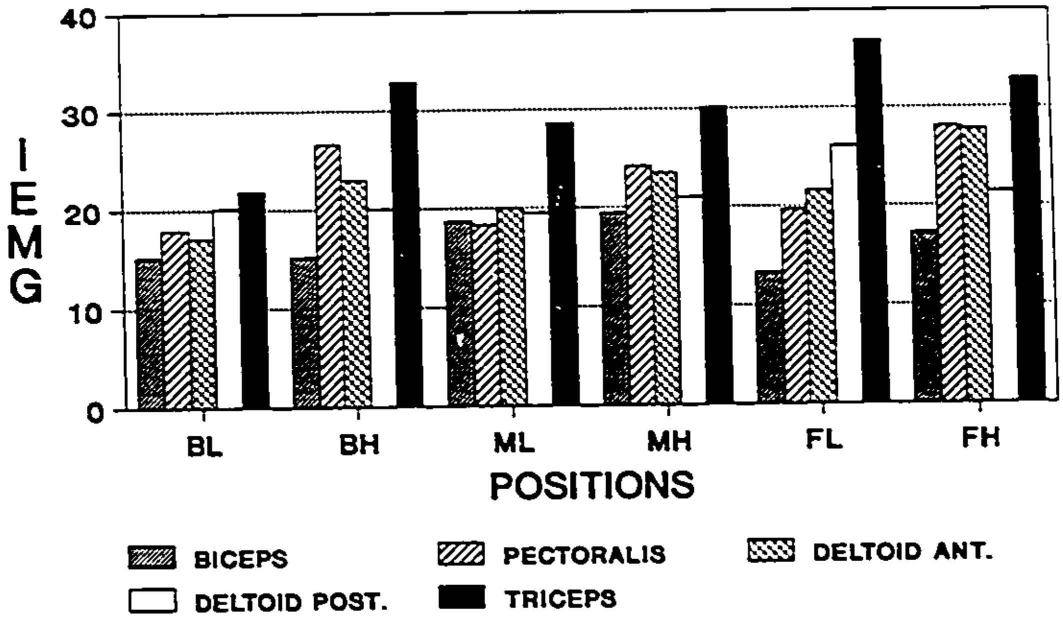

Figure 8: Means IEMG of each muscles for each seating position; Backward-Low (BL), Backward-High (BH), Middle-Low (ML), Middle-High (MH), Forward-Low (FL), and Forward-High (FH).

Bonde-Petersen (1974) 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 were thought to be caused by greater degrees of contact with the pushrim, since it was previously observed that greater degree of contact were found with lower seat positions (see Table 3). Also, a better orientation of the segments on the pushrim during wheelchair propulsion may affect the amplitude of the IEMG recorded. Although, the overall amplitude in IEMG were lower for the Backward-Low position it should be noted that the Middle-Low position would offer more stability than the Backward-Low position, because the longer the distance between the centre of mass and the main axles, the greater the resistive moment arm will be (Peizer et al., 1964). Therefore, if the centre of mass is closer to the main axles the wheelchair stability is less (Loane & Kirby, 1985).

The lower overall IEMG activity observed for the Backward-Low positions would indicate that in moving the centre of mass, the amount of force applied to the pushrim was affected, since moving the seat backward caused a decrease in rolling resistance ratio between the two rollers. Brubaker (1986) found that in moving the centre of mass closer to the axle of the rear wheels the rolling resistance ratio decreased. Therefore, lower IEMG activity should occur at the backward positions since less forces are required to propel. However, the Backward-High position had slightly higher IEMG activity than 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 a poor contact (location and orientation of the hand) made with the pushrims, mainly since the orientation and location with which the hands contact the pushrims differ, this would agree with Sanderson and Sommer (1985).

The kinematic has revealed that for the Backward-Low and the Middle-Low positions the time at which the peak velocities of the upper limbs joints and segments occured was closer to the end of the pushing phase when compared to the high positions. Therefore indicating a better synchronisation in the motion for these positions. Also for the Backward-Low and Middle-Low the change of motion of the upper limb joints and segments were found to occur in a sequence (hand, forearm, and upper arm for the segments; wrist, shoulder, and elbow for the joints) which revealed that the segments and joints motion were more continuous. For the Backward-Low position the elbow and forearm decceleration slopes were less abrupt during the pushing phase which showed a smoother change of motion. The wrist and hand positive and negative acceleration was also lower for the Backward-Low position which revealed that the hand and wrist are subjected to less change of motion in comparison to the High positions.. The Middle-Low position was found to have the lowest pushing frequency while the Backward-Low had the lowest overall IEMG when compared to all the other positions. Thus the kinematic pattern and the EMG activity seems to indicate that the Backward-Low and the Middle-Low positions are slightly better seating position than the other positions evaluated. However, it should be noted that the small number of subjects and the high variation observed among the subjects have made the use of statistics inappropriate, thus the above conclusion are reached from a descriptive comparaison. These conclusions would be in agreement with Burk (1986), Schuman (1979) and Walsh (1986). Walsh (1986) specified that the seat should be slightly behind the hip bone thus aligning the shoulder with the main axle in order to allow the forces to be applied directly downward. A Backward-Low seat position according to Higgs (1983) was also found to be related to greater success for long distance racers. However, some discrepancies exist with the findings of Brubaker et al. (1980) and a Middle-Middle and Middle-Forward seat position for the pushrim propulsion was observed to be correlated with higher efficiencies. The differences observed with our results are difficult to explain since the seat positions tested by Brubaker et al. (1980) did not correspond to our positions and the subjects used for the study were able bodied. Also, Walsh et al. (1986) 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). Also, the hands were subjected to less acceleration at the end of the recovery phase for the lower seat positions, thus assuring a better hand contact at the point of grabbing with the pushrims. Among the positions evaluated the Backward-Low position had the overall lowest IEMG and the elbow and forearm acceleration slopes were less abrupt than the other seat positions thus indicating a smoother motion. While 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 were recorded and greater degrees of contact were observed for the Low positions and that the overall IEMG was lower for the Backward-Low position. Based on these descriptive observation 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 trunk angular momentum. This non-relationship was explained by the variability among the subject's technique of propulsion, weak hip flexors associated with class IV, and the 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 the development of wheelchair design. However, further studies should be conducted in this area to confirm our observation and to provide more information about the ideal seating position.

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Technical Note
Equipment for Testing Various Wheelchair Designs

Running Head: TESTING WHEELCHAIR

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Abstract

A testing wheelchair was developed to permit the evaluation of various wheelchair designs such as; seat height, seat base angle, seat tilt, horizontal position of the wheel, wheel camber, and different sizes of pushrims and wheels. The chair was mounted on high inertial rollers developed to simulate propulsion in a stationary position in order to allow the analysis of wheelchair design. The roller's speed was measured through a tachometer which was calibrated to an accuracy of less than 5%.

Equipment for Testing Various Wheelchair Designs

Sport for the physically disabled was first introduced in the mid 1940's. The wheelchair which was used for sporting activities then did not differ from the one used for everyday activities. Since athletes are always looking for new ways of improving their performance, major modifications occured to the existing wheelchair. This included removing the arm rest and brake, lowering the seat, cambering the wheels and moving them backward, increasing the size of the rear wheels and reducing the size of the pushrims (Walsh, Marchiori & Steadward, 1986). The sport wheelchair of today has evolved into a sleek light-weight racing machine, much like a competitive racing bicycle. The sport wheelchair of today uses tubular tires able to withstand 6.9 mPa, high quality bearings, and a light-weight responsive frame (Rudwick, 1979).

Most of the improvements observed over the years in wheelchair design were done by the athletes themselves in their desire to attain better performance. As of now the athletes are lacking some scientific knowledge to help them adjust their wheelchairs or to design their own. In order to perform studies on wheelchair design it is essential to have a wheelchair which permits one to alter certain variables. Therefore, an adjustable wheelchair must be built for investigations into this area. Also, the study of wheelchair design can be difficult in a laboratory setting without a system which can allow for wheelchair propulsion to be simulated in a stationary position. The use of a treadmill proves to be useless since in most cases the wheelchair is too wide and it is not safe enough for the subject. Therefore the purpose of this paper is to introduce an adjustable wheelchair which can be used to test different wheelchair designs, and to show a system which can be used to simulate wheelchair propulsion in a stationary position as well as monitoring the speed of propulsion.

Testing wheelchair

The designed wheelchair shown in Figure 1 can offer a wide range of adjustments such as; the seat height, the seat base angle, the seat tilt, the horizontal position of the rear wheel and the wheel camber. The rear wheels and the pushrims size can also be varied with this chair.

All the seat adjustments are made by moving part A and B of the chair (see Figure 1). In moving only part A the angle between the seat base and back rest is changed. If part B is raised alone the seat base angle is varied. To vary the seat tilt part A and B need to be moved in different directions (up and down), and moving both parts the same direction caused a variation in seat height.

The wheel is mounted on an axle block (see Figure 1 part C and Figure 2 for a detailed illustration of part C) which can be moved horizontally on the supporting frame. To move the axle block the four thightening screws need to be loosened then the axle block can either be moved forward or backward on the frame (see Figure 2). The axle of the wheel is fixed to a metal support which is then mounted to the axle block as shown in figure 2. This double joint connection was used to rotate the axle in order to vary the wheel camber. The wheel camber adjustment is done by moving the screw which connects the plate camber extension and the axle block extension, both extensions serve to secure the inclination of the wheels.

As shown in Figure 1, racing wheels were mounted on the axle of the chair and pushrims were fixed on the spokes. The mechanism used to attach the pushrims on the spokes of the wheels is illustrated in Figure 3. In using this wheelchair design the size of the wheels and the pushrims can be easily changed and various wheelchair design can easily be simulated.

Wheelchair rollers

To simulate wheelchair propulsion in the laboratory the wheelchair was mounted on two sets of rollers as shown in Figure 1. Due to the hollow characteristic of the rollers the moment of inertia was very small. Therefore, to improve the inertial properties of the rollers the mass of each roller was increased by adding two iron rings on each of the rollers (see Figure 1). Each ring had a mass of 2.72 Kg with a moment of inertia of 0.052 Kg.m.m., the inertia was mathematically calculated by integration. Adding the rings augmented the system's rotational inertia and therefore increased the rollers angular momentum properties.

Tachometer

To monitor the speed of rotation a tachometer was used. The shaft of the tachometer was supported on an arm which was fixed on the side of the rollers (see Figure 1). Between the shaft and the support arm joint rotation was possible. A small wheel was embedded on the shaft and rested on the roller. The wheel was surrounded by rubber to prevent any slippage between the roller and the tachometer. In using this system, any rotation of the wheel was translated into the tachometer through the rollers. For each rotation the tachometer registered the difference of potential. The difference of potential was then recorded on a microcomputer (Compaq 386, 16 MHz) using a data acquisition system (Lamontagne, Bradley & Lemaire, 1989).

Calibration

The tachometer was calibrated with an electronic counter to convert the voltage into speed (m/s). The counter was mounted on a small piece of wood which extended in front of the roller and a piece of reflective paper was taped on the roller. The counter recorded each time the reflective tape passed in front of the counter, which corresponded to one complete revolution. The calibration consisted of spinning the roller at various speeds and to monitor the counter and tachometer signals. The electronic counter and the tachometer signals were stored on microcomputer (Compaq 386, 16 MHz) through a data acquisition system (BIOAD, University of Ottawa) to digitally convert the signal. The data were collected at 50 Hz for a period of 20 seconds for 10 trials. Then, the tachometer data were filtered at a cutoff frequency of 4 Hz (single pass critically dampened digital filter). The conversion factor (voltage into speed (m/s)) was then easy to calculate since we knew the distance and time required to perform one revolution (each time the roller completed one revolution the counter was increased by 1) and the corresponding voltage. Therefore, the following equation was used to calculate the speed of rotation:

$$\text{Speed} = d / (T_{(n+1)} - T_{(n)})$$

where

d= distance for one complete revolution (0.4021 meters)

$T_{(n)}$= time to complete revolution n.

$T_{(n+1)}$= time to complete revolution n+1

The speed and the voltage value were least square fit to calculate the equations of the curves. The least square fitting served to generate an array of speed and volt value for equivalent time. Thus, the slope was a simple ratio of the speed and the voltage:

$$\text{Slope}_{(n)} = \text{speed}_{(n)} / \text{voltage}_{(n)}$$

where

$\text{Slope}_{(n)}$= slope at data point n

$\text{Speed}_{(n)}$= speed at data point n

$\text{Voltage}_{(n)}$= voltage at data point n

For each trial the mean slope and the coefficient of variation was calculated. The within and between trial varibility of scores was expressed as a proportion of their mean by the coefficient of variation (CV):

CV = (SD/mean) x 100%

From Table 1A, it was calculated that the within trial variability was less than 2%. The between trial variability showed that less than 5% of variation was associated with the conversion factor (1.16 (m/s)/volt).

Table 1A
Calibration of the tachometer

Trial	Mean slope (m/s)/volt	Standard deviation	Coefficient of variation
1	1.20	.0197	1.64%
2	1.24	.0154	1.24%
3	1.17	.0203	1.74%
4	1.23	.0128	1.04%
5	1.10	.0170	1.54%
6	1.07	.0191	1.78%
7	1.19	.0012	0.99%
8	1.11	.0189	1.70%
9	1.14	.0154	1.45%
10	1.13	.0153	1.36%
Mean	1.158		
S.D.	0.057		
C.V. (%)	4.92%		

The speed could be made visual to the subject through a voltmeter, as shown in figure 1. To help the subject know their actual speed of propulsion the volt scale on the volmeter can be converted into speed (m/s) using the conversion factor previously calculated.

Conclusion

A simple testing wheelchair was designed to permit the evaluation of various wheelchair design such as: the seat height, the seat base angle, the seat tilt, the horizontal position of the rear wheel, and the wheel camber. Also, wheelchair roller has been developed to simulate wheelchair propulsion in a stationary position. The speed of propulsion was monitored through a tachometer which was calibrated to an accuracy of less than 5%.

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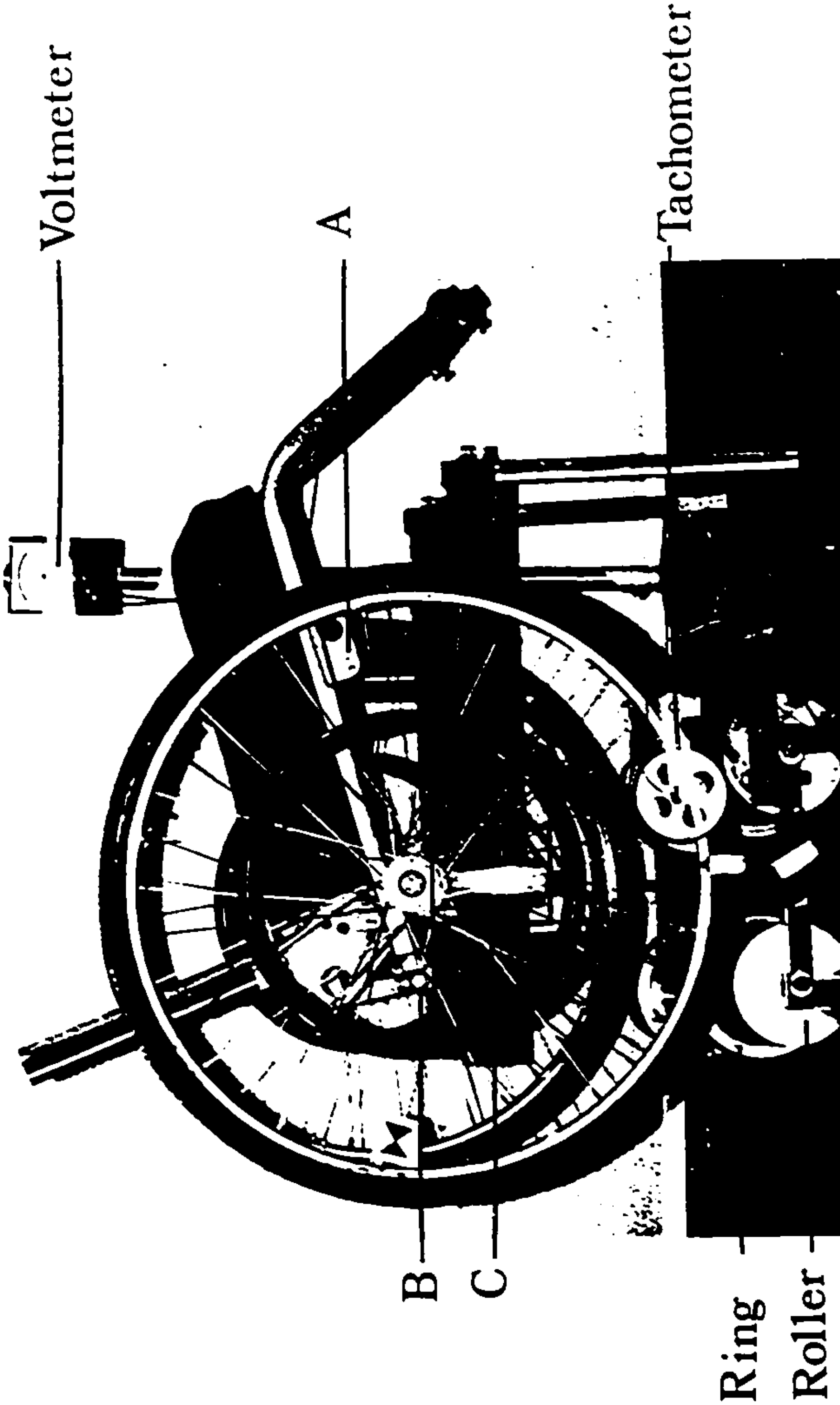

Figure 1: Testing wheelchair and rollers

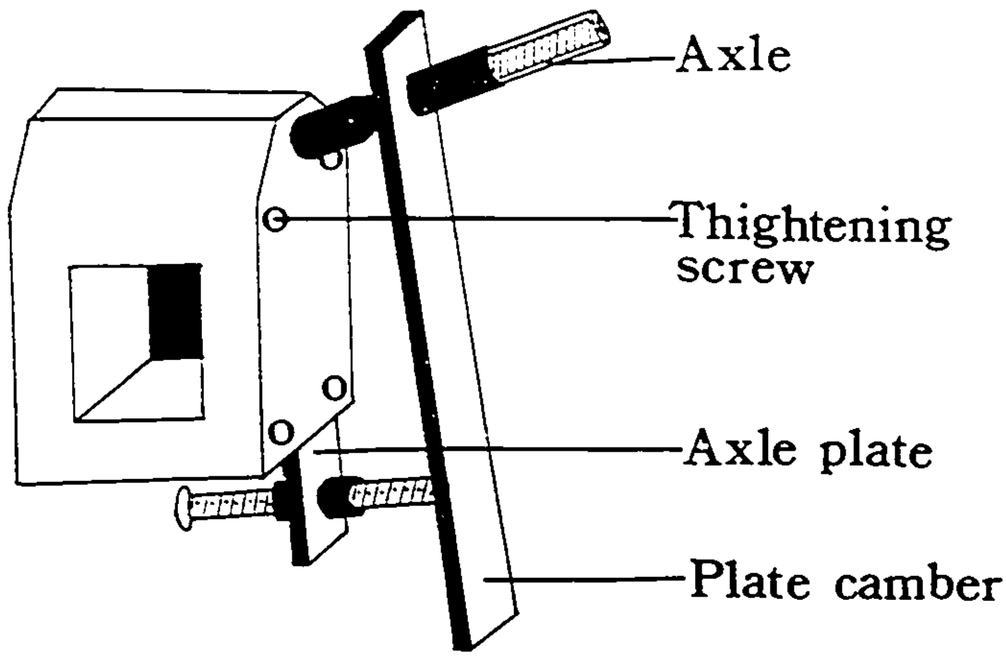

Figure 2: Axle block

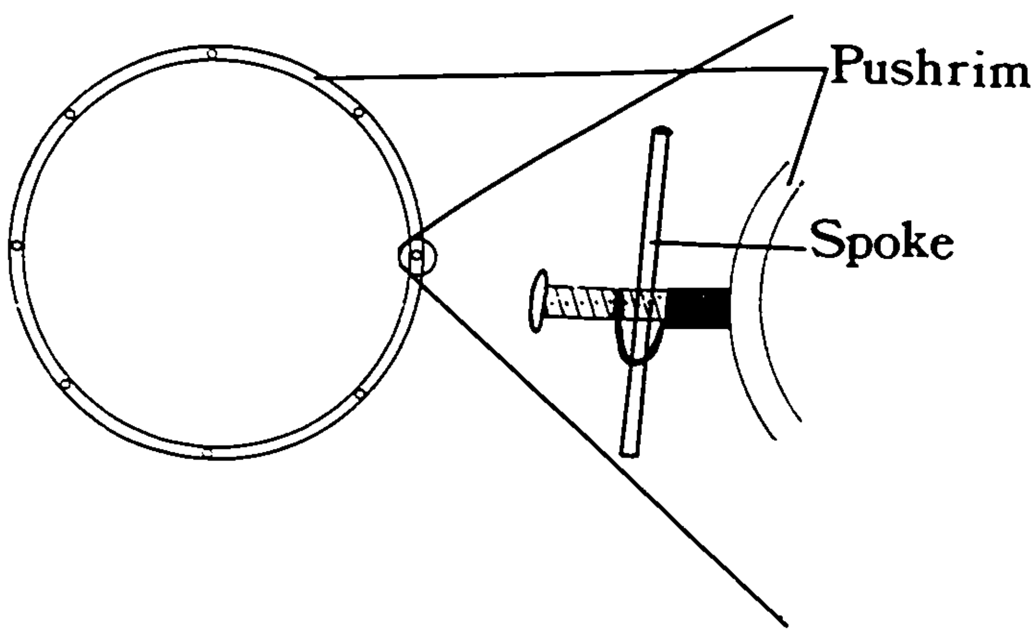

Figure 3: Pushrim attachments

APPENDIX A
ANTHROPOMETRIC MEASUREMENTS

The segments length for this investigation were defined as follow:

Upper arm: Glenohumeral joint \ lateral epicondyle of the humerus

Forearm: Lateral epicondyle of the humerus \ styloid process of the ulna

Hand: Styloid process of the ulna \ proximal phalanges of the second finger

Total arm: Glenohumeral joint \ distal phalange of the second finger

Trunk: Greater trochanter \ glenohumeral joint

Table A
Anthropometric measurements for all the subjects

Subject	Upper arm (cm)	Forearm (cm)	Hand (cm)	Arm (cm)	Trunk (cm)
1	24.6	24.3	9.7	68.1	34.0
2	27.1	25.7	7.9	69.7	34.4
3	28.7	24.6	9.3	72.8	38.1
4	26.7	25.3	9.2	72.1	40.1
5	27.1	24.2	11.3	70.4	45.8
X	26.8	24.8	9.5	70.6	38.5
S.D.	1.31	0.58	1.09	1.68	4.31

APPENDIX B

CONSENT FORM

Faculty of Health Science
School of Human Kinetics
Department of Kinanthropology

Consent to a Kinematic and EMG Analysis of Wheelchair Propulsion for Various Seating Positions

Louise Mâsse and Mario Lamontagne Ph.D.

Name of Volunteer:__________________________Date:____________________

I **UNDERSTAND** the terms of my participation in this study including the procedures to be used and the possible risks involved, and I agree to participate as a volunteer.

I **AM FREE TO WITHDRAW** this consent and to discontinue my participation at any time without penalty or discrimination.

MY PRIVACY will be protected in the following manner:

All research data obtained about me during the course of this study will be kept confidential and accessible only to the principal investigator and assistants on the project.

Should the study be published, my identity will not be released.

INFORMATION ABOUT THE STUDY:

The purpose of the study is to examine how the technique of propulsion is influenced by changes in seat position, using kinematic and electromyographic parameters.

One session of two hours will be required for the experimentation.

The only discomforts or inconveniences expected are: the markers or electrodes placed on the skin and propelling the wheelchair in different positions.

The procedure to be follow is discussed on the next page.

Signature of volunteer and Date: ____________________________________

Signature of Witness ____

Continuation to another page: **YES**

PROCEDURE

Prior to the experiment you will be asked to propel the designed racing wheelchair on the rollers in order to determine your maximal speed of propulsion. This maximal test takes approximately three minutes including the warm up period. The muscular activity of the upper arm will be recorded with surface electrodes. Before the application of the electrodes, the skin will be rubbed with alcohol, shaved and rubbed with electrolyte paste. The lengths of some body segments (forearm, arm, trunk, leg...) will be measured and skin landmarks will be affixed to the upper-body joints. The experiment will consist of propelling the wheelchair at 60% of your maximal speed at six seating positions which consist of a combination of two seat heights (low and high) and three horizontal positions of the back wheels (forward, middle and backward). At each experimental condition the upper arm motion will be filmed and the EMG activity will be recorded. Each condition will take approximately five minutes since there are six conditions you will be asked to moderately exercise for approximately 35 minutes intermittently for the entire experiment. You may be asked to remove some clothing to affix the body markers as well as the surface electrodes. You may refuse to take-off your clothing or to perform the movements that the experimenter requests without penalty or discrimination.

In signing this consent form you acknowledge that you have read and understood the above statements. You enter the biomechanical investigation willingly and you may withdraw **At ANY TIME** without penalty or discrimination.

Signature of Volunteer and Date: ______________________________

Signature of Witness ______________________________

APPENDIX C
THESIS PROPOSAL

Kinematic and Electromyographic Analysis of Wheelchair Propulsion for Various Seating Positions

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INTRODUCTION

In the past decade we have witnessed tremendous improvements in wheelchair racing records. For the mile the record was changed from approximately six minutes to less than four minutes, and for the Marathon from three hours to one hour and forty minutes (Brubaker, 1986). 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 1940's, 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 (Walsh, Marchiori, & Steadward, 1986). The athletes themselves are responsible for most of the improvements observed in wheelchair design. They have slowly modified the conventional wheelchair into a streamlined light-weight racing machine according to their individual needs. The racing wheelchair of today is the result of a laborious process of trial and error work. Despite the changes that have occurred over the past years a large amount of uncertainty still exists as to the design of the ideal racing wheelchair.

Their are several factors which may affect the athlete's performance, including; the strength of upper body muscles, the physical capacity of the individual, the level of neurological lesion, and the interaction between the user and the wheelchair (Brubaker & McLaurin, 1982; van der Woude, Veeger, & Rozendal, 1986). Wheelchair design plays an important role in optimizing the athlete's performance, since different seat positions alter the athlete's pattern of propulsion and consequently affect the performance. One of the most prevalent problems facing the athletes today is to define the ideal seat position needed to achieve an optimal propulsion technique (Walsh et al., 1986). Therefore, research in wheelchair design may enable athletes to improve their performance through a better technique of propulsion.

The relationship between seating position and the pattern of propulsion has not yet been investigated, however, some authors have looked at the effect of varying the seating positions on performance. Brubaker, McLaurin, and Gibson (1980) studied the effect of varying the seat position on the mechanical efficiency and they observed that both a middle-middle and a middle-forward seat position were found to have higher mechanical efficiency. As well, lower pushing frequency and smaller energy expenditure were associated with higher mechanical efficiency. However, their findings do not agree with those of Higgs (1983) and Walsh et al. (1986). Higgs' (1983) static analysis of wheelchair racing used during the 1980 Olympic games for the disabled revealed that a low-backward seat position was found to be highly correlated with success for long distance athletes. Walsh et al. (1986) studied the effect of seat position on the maximal linear velocity of wheelchair sprinting and found that no significant differences existed between the maximal linear velocities with a change in seat position. Since the subjects' levels of neurological lesion were not taken into account by Higgs (1983) and Walsh et al. (1986) experiments and because a small number of subjects were used for these investigations (Brubaker et al., 1980 & Walsh et al., 1986) this might explain the discrepancies observed among these studies. According to Steadward (1979) subjects' having various levels of lesions were observed to use different technique of propulsion and were found to have different electromyographic recruitment patterns. Therefore, the neurological level of lesion and the anthropometric variability of the subjects as well as a change in seat position can alter the technique of propulsion and consequently the performance. At the present time doubts still remains as to the identification of the ideal seating position. Therefore, further research should be conducted in this area to facilitate the process of optimizing wheelchair design.

Statement of the Problem

This investigation was designed to examine the influence of seating positions change on the kinematic and the electromyographic parameters of the upper limbs during wheelchair propulsion. The pushrim propulsion technique was investigated for six seat positions which consisted of a combination of three horizontal rear wheel positions at two sitting heights at 60% of the maximal speed of propulsion.

Rationale

This study will provide an initial understanding of how the technique of propulsion is affected by a change in seat position. Before any improvement in seat position can be made it is necessary that the researchers develop a full understanding of how the technique of propulsion is influenced by a change in seat position.

Researchers in the rehabilitation area will gain valuable information about the technique used for propulsion at different seating positions, which may be useful in wheelchair design. This information can help the researchers in developing a wheelchair which will be compatible with any level of performer. Athletes and coaches may also benefit from this information allowing them to perfect the technique of propulsion through the use of a better seating position.

Limitations

The limitations within which this investigation was conducted can be summarized as follows:

1) The sample of subjects for this investigation was limited to five subjects having a neurological lesion between T11 to L2. The results of this investigation must be considered within the context of this delimitation before generalizations can be made.

2) This study was limited to the subject's ability to adapt to the laboratory chair, the wheelchair rollers, and the various changes in seating position. It was

also limited to the ability of the subject's to maintain a constant speed of propulsion.

3) The cinematographical analysis was restricted to two dimensions along the subject's right sagittal plane. Therefore, the motion of the upper extremities during wheelchair propulsion are assumed to be bilaterally symmetrical and to move mainly in the sagittal plane. This implies that any rotation is insignificant and the right and left side motion are identical. It then, becomes acceptable to digitize only one side of the body and it eliminates the use of a three-dimensional cinematography.

4) This investigation was limited to inherent errors which are introduced in using cinematographical technique (Winter, 1979). In order to decrease some of the perspective and parallax errors, the camera was placed as far as possible (approximately 11 meters away from the subject). To minimize the digitizing errors, the image was projected as wide as possible to increase the accuracy of digitizing. The marker coordinates were then digitized to an accuracy of approximately 1.98 mm using an image projected at 9.3% life-size.

5) The anthropometric model used for the BIOMECH program was based on Dempster's studies (Winter, 1979). Dempster's measurements served as a good approximation for paraplegics in the absence of better data. In using a link segment model, it is assumed that the body segments are rigid, therefore the inertial properties are constant, meaning that the movement of the body fluids or muscles do not affect the inertial properties of the limb considered (Hay, 1973). Also, the anthropometric data from Dempster's assumes that the joints are hinge joints.

Definitions

The following terms have been operationally defined for this study.

Contact angle: the angle formed by the line joining the point at which the hand contacts the pushrims with the main axle and the positive vertical axis.

Cycle time: the amount of time required to perform one cycle. The time from when the hands are in contact with the pushrims and are increasing the speed of the wheels until they contact it again to increase the speed of the wheels.

Degree of contact: the angle formed by the point at which the hand contacts the pushrim, the main axle of the rear wheels, and the point at which the hand releases the pushrim.

Rear wheel: the main wheels (big wheels) of the wheelchair.

Pushing phase: The percentage of time during which the hands are in contact with the pushrims and increasing the velocity of the pushrims.

Pushing frequency: the number of strokes per minute.

Pushrim: the circular rims which are attached to the rear wheels. They are used to propel and steer the wheelchair.

Recovery phase: the percentage of time during which the hands are not generating an increase in speed of the rear wheels.

Release angle: the angle formed by the line joining the point at which the hand releases the pushrims with the main axle and the positive vertical axis.

Seat back angle: the angle formed by the seat base and the back seat of the wheelchair.

Seat base angle: the angle of the seat made with the positive horizontal axis.

Wheel camber: the angle formed by the rear wheels and the positive vertical axis in the frontal plane.

Review of Literature

The following chapter will outline the findings of researchers in the area of wheelchair kinematics, wheelchair design, and electromyographical analysis during wheelchair propulsion. The level of neurological lesion will also be discussed in an attempt to understand how it can affect the technique of wheelchair propulsion.

Level of lesion

The performance or technique of propulsion is dependent upon the level of the neurological lesion of the individual. Therefore, guidelines have been established for the purpose of classifying competitors in an attempt to provide fair competition. The medical classification system is presently used as a guideline for classifying the athletes. It is based on a medical diagnosis which indicates the level of neurological lesion and the amount of muscle available for use by the individual (McCann, 1979). This classification system has helped to regroup athletes with similar disabilities thus avoiding unfair competition. There are seven classes within the classification system; class IA, IB, and IC are for quadriplegics and class II, III, IV and V are for paraplegics (Corcoran et al., 1980). The validity and fairness of such a classification system has been studied by Steadward (1979). Steadward (1979) evaluated the validity of the medical classification system for two tasks: wheelchair dash and the shot put event using cinematographic and electromyographic techniques. The purpose of Steadward's study was to evaluate the techniques used and the amount of trunk musculature involved during performance of each task and to see if any similarities existed within each class. In the shot put event, the paraplegics showed more consistencies than the quadriplegics. The quadriplegics needed more individual compensation which explained the variability among the subjects (Steadward, 1979). For the wheelchair dash event, the kinesiological activity showed no trunk movement for the quadriplegics group. Also, very small differences in the electromyographical data were found between class IB and IC. Therefore, the data did not support the existence of a separate class. The EMG activity of the trunk muscles demonstrated variation between the classes. It was observed that class IV had a definite advantage over class II and class III due to their stronger lower abdominal musculature (Steadward, 1979). Steadward's results supported the classification in it's present form except for class IB and IC, because of the similarities observed within the class and the differences observed between the classes.

For the proposed investigation, subjects in class IV were used. A neurological lesion from T11 to L2 is classified as class IV (Corcoran et al., 1980; McCann, 1979). Subjects within that class have full use of their upper extremities and possess good abdominal musculature as well as spinal extensors with the possibility of some hip flexion and adduction (McCann, 1979; Madorsky, 1979). 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 phases of propulsion

Wheelchair propulsion consists of a cyclic movement which can be divided into phases. Steadward (1976) was the first to describe wheelchair propulsion as a function of two phases; the push phase or drive phase and the recovery phase. The drive phase was defined as the time during which the hands are in contact with the pushrims. While, in the recovery phase the hands are not in contact with the pushrims and are repositioned for a new drive phase.

The definition presented by Steadward (1976) referred to the phases of propulsion in terms of contact only while Spooren (1981), Higgs (1984), and Sanderson and Sommer (1985) introduced the concept of force to define the phases of propulsion.

Spooren (1981), described the push phase as the time during which forces are applied to the pushrims. The dead phase was defined as the time during which no forces are applied to the pushrims.

Higgs (1984) proposed an improved description of the phases of propulsion consisting of four phases: the initial contact phase, the propulsive phase, the disengagement phase and the recovery phase. The initial and disengagement phases were said to represent the time during which the hands are in contact with the pushrims but do not generate any forces, while in the propulsive phase the hands are applying forces to the pushrims in order to increase or maintain the speed of propulsion. The recovery phase was defined as the time during which the hands are not in contact with the pushrims and are repositioned for a new initial contact phase.

Although Higgs definition is best, it was not used in his study. Higgs (1984) explained that a cinematographical analysis did not permit him to differentiate between the initial phase, the propulsive phase, and the disengagement phase. The phases which Higgs utilized were then the same as those proposed by Spooren (1981).

Sanderson and Sommer (1985), described wheelchair propulsion as consisting of two phases; the propulsive phase and the recovery phase. The propulsive phase was said to represent the time during which the hands are in contact with the pushrims. It is during this phase that forces are applied in an attempt to either increase or maintain the speed of propulsion. The recovery phase was said to represent the time during which the hands are not in contact with the pushrims and are repositioned for a new propulsive phase. As explained by Sanderson and Sommer (1985) these definitions are more operational because there exists a time during which forces are not applied to the pushrims but the hands are still in contact. There may also be a time when the hands are contacting the pushrims, yet moving at a slower speed, and resulting in the reduction of velocity.

Kinematics

Only a small number of researchers have investigated the kinematics of propulsion. All have essentially attempted to understand the technique which is used by wheelchair sportsmen.

The technique of propulsion

Steadward (1976), described the technique which should be used by wheelchair racers. According to Steadward (1976), before the propulsive phase, the arms and hands should be well back as the hands contact the pushrim. Upon contact the shoulders should forcefully flex thus pulling the hands forward. Once the hands are at the top of the pushrims the elbows should extend thus pushing the hands down and at this point the trunk should lean forward as much as possible. The athlete should remain in contact with the pushrims until the elbows are fully extended and the hands are pushed as far as possible. As the athlete releases the pushrims his trunk should remain forward and he should move his arms in a circular motion consisting of elbow flexion and shoulder extension. Steadward (1980) claimed that it is important to minimize the trunk motion in order to maximize the streamlining effect. The use of a pump arm action during the recovery phase was also not recommended, however Steadward (1980) did not provide any evidence to support this view. He also claimed that the recovery phase should be as short as possible, since it does not contribute to the motion. Although Steadward (1976) has provided a good basic description of the technique, we lack the kinematic analysis to fully understand the technique of propulsion.

In an attempt to understand the kinematic of wheelchair propulsion Sanderson and Sommer (1985) studied the technique of propulsion for three male paraplegics having a neurological lesion at T12. The subjects were filmed at 60-65% of their maximal oxygen consumption on a treadmill. The kinematic analysis revealed that each subject exhibited a different movement when propelling their wheelchair, but a regular pattern was observed within the subject for a period of 80 minutes. Therefore, the amount of time spent in the recovery or the propulsive phase was found to be constant within the individual. Different pushing styles were observed among the subjects, these being either a circular or a pump arm action. Sanderson and Sommer (1985) seem to favor the circular pushing style rather than the cyclic pump arm motion, this is in agreement with Steadward (1979). Sanderson and Sommer (1985) claimed that the pump arm action would necessitate a more abrupt change in the hand direction. Therefore, the shoulder and elbow joints would necessitate more deceleration and acceleration at the end of the propulsive phase or the recovery phase. Effectively, the peak angular velocities of flexion and extension for the elbow or shoulder joints were found to be higher for the pump arm motion when compared to the circular pushing style (Sanderson and Sommer, 1985). However, the peak angular velocities for the upper limb joints did not provide the information on the changes in angular velocities for both phases of propulsion. Although it seems to make some intuitive sense that the upper limb would be subjected to more acceleration and deceleration with the pump arm motion. This implies that more muscular activity would be needed to decelerate and accelerate the limb (Sanderson and Sommer, 1985). Therefore, greater energy would be expended as a result of higher acceleration and deceleration of the limb as observed by van der Woude, Veeger, Rozendal, van Ingen Schenau, Rooth, and Nierop (1986). Sanderson and Sommer (1985) claimed that the more forward lean the individual had, the better his ability to transfer power to the pushrim and more time could be spent in the propulsive phase. The increase in power is thought to occur as a result of more effective force application (Sanderson& Sommer, 1985).

The effect of different pushrim diameters on subjects physiological parameters as well as the technique of propulsion was investigated by van der Woude et al. (1986). In their study they found that the pushrim size had no effect on the pushing frequency, push time, recovery time, and the push angle. However, the subjects' oxygen consumption were found to systematically increase with pushrim diameter. Greater pushrim diameter caused a significant increase in the linear hand velocity, as result the upper arm was thought to be subjected to greater acceleration and deceleration. These factors combined with greater angular trajectories may served to explain the increase in oxygen consumption. Therefore, an increase in oxygen consumption may have resulted from greater acceleration and deceleration of the upper limb.

The athlete Jan-Owe Mattson demonstrated that the technique of propulsion has a great effect on performance (Alexander, 1987). A film analysis of Mattson's propulsion technique revealed that it was superior to the technique employed by quadriplegics of any class. The technique which he used made the most efficient use of the muscles which he had at his disposal. Mattson used a backhanded pronated to supinated hand position with high friction gloves. This permitted him to recruit the following muscle groups; the pectoralis major, the deltoids, the serratus anterior, the levator scapulae and the rhomboids (Alexander, 1987). Since these muscles are not greatly affected by a C6-C7 neurological lesion one can see the advantages of using this technique. The film analysis of Mattson technique also revealed that 55% of the cycle was spent in the propulsive phase and 0.32 seconds was required for one stroke (Alexander, 1987). This is in contrast with other quadriplegics of the same class which spent 33% of their time in the propulsive phase and required 0.92 seconds for one stroke (Alexander, 1987).

The effect of speed on the technique of propulsion

A kinematics comparison between wheelchair sprinters and long distance racers was made by Higgs (1984). Higgs (1984), filmed sixteen world class athletes (class IV and V) during the Special Olympics competition held in Holland in 1980. It was found that the movement of the hands during wheelchair propulsion differed between the two classes of racer. The best long distance racers had a circular hand motion while the best sprinters had a back and forth hand motion. Higgs (1984) reported that the sprinters point of contact and release for the propulsive phase was different than the long distance racers. The long distance racers were found to start the propulsive phase sooner and completed it later than did the sprinters. This resulted in a greater push angle and propulsive phase for the long distance racers, indicating that the hands for the sprinters were in contact with the pushrims only when maximal forces can be generated. The sprinters were also found to have greater pushing frequency (numbers of stroke per minute) and shorter recovery time than did the long distance racers. Many of the athletes observed by Higgs (1984) were found to have wasteful movements such as the presence of an anterior or posterior loop in their pattern of propulsion. The presence of a loop was explained to be nonproductive since it only has the effect of lengthening the cycle time. The technique of propulsion for the best sprinters and long distance racers did not have an anterior or a posterior loop (Higgs, 1984).

According to Walsh et al. (1986), a negative correlation existed between the time-on the pushrim and the linear velocity as well as the pushing frequency and the linear velocity. According to Walsh et al. (1986) an increase in pushing frequency is thought to be more critical in producing an increase in speed than the amount of time spent in the contact phase. This was however, in opposition to Spooren (1981) who suggested that to maximize the speed of propulsion the push time should be as long as possible. However, Spooren (1981) did not provide any data to support his view.

Lamontagne (1988), compared the kinematics of wheelchair propulsion for a male paraplegic at two leg positions and for various speeds. The amount of time spent in the propulsive phase was approximately 38% of the cycle at the two leg positions and at all speeds tested, indicating that an increase in the speed of propulsion was independent of the amount of time spent in the propulsive phase. This is in disagreement with Walsh et al. (1986), who observed that an increase in the speed of propulsion was generated by a decrease in the propulsive phase combined with a greater cycle time. Lamontagne (1988) stated that a better transfer of momentum was observed at the trunk for the leg down position than with the leg up position for both speeds of propulsion.

Van der Woude, Hendrich, Veeger, van Ingen Schenau, Rozendal, De Groot, and Hollander (1988) studied the relationship between the power output on the oxygen uptake and the technique of propulsion. Two workload strategies were used in this investigation. In strategy one the speed was varied but the slope was kept constant while in strategy two the speed was constant but the slope was varied. The measured cardio-respiratory parameters and the technique for propulsion were found to vary with the power output, slope and speed. At an equal power output, a higher resistance with a lower speed caused a lower cardio-respiratory response as well as a distinctive difference in the technique of propulsion. The push time was found to decrease with a greater speed of propulsion but increase with a change in slope (van der Woude et al., 1988). This last result is in agreement with Walsh et al., 1985 who stated that an increase in speed was negatively related to the time-on the pushrim. Greater power output was found to be mainly generated by an increase in pushing frequency, regardless of the strategy. Therefore, an increase in speed was mostly generated by greater pushing frequency. The push angle defined as the angular trajectory of the hand during the propulsive phase with respect to the main axle, was not found to vary for either strategy (van der Woude et al 1988). Van der Woude et al. (1988) claimed that a constant push angle may be related to the seat configuration which restrained the body motion.

Wheelchair design

At any velocity of propulsion less energy is expended for the sport wheelchair when compared with the conventional wheelchair (Hilbers & White, 1987). The differences in energy expenditure were attributed to the difference in design. Some studies (van der Woude et al., 1986) have compared the mechanical efficiency of the sport and the conventional wheelchairs and found that the sport wheelchair was more efficient. This was in part attributed to a better seating position and a better wheelchair design. Therefore, improved performance can be achieved with the racing wheelchair because it requires less energy. However, the performance itself may be affected by several factors. Some of the factors which affect the performance include: the forces exerted on the pushrim, the physical capacity of the subject, the disability level and the interface between the users and the wheelchair (Brubaker & McLaurin, 1982; van der Woude, Veeger & Rozendal, 1986). The seating position plays a major role in the application of forces to the pushrims. Therefore, a proper orientation of the forces applied to the pushrims will result in a greater performance and higher efficiency (Brubaker, 1987).

Most studies related to wheelchair design have attempted to relate a good performance and/or a high mechanical efficiency with a specific seating position or design. The major concern of these studies was to evaluate if different wheelchair designs or seating positions would affect the mechanism of force application.

Seating position and the location of the centre of mass

The propulsive forces which are applied to the pushrims are affected by the position of the rear wheels. In moving the rear wheels horizontally it has the same effect as moving the seat backward or forward. As stated by Sanderson and Sommer (1985), the location of the rear wheels will affect the magnitude, the duration, and the location of the forces applied to the pushrims, since the amount of forward lean is directly related with the position of the rear wheels. Moving the rear wheels has a tremendous effect on the position of the centre of mass (C.M.) (Burk, 1986). If the rear wheels are moved forward this will facilitate the turning speed at the cost of the stability because it lightens the front end by moving the C.M. backward (Sanderson & Sommer, 1985; Walsh, 1985).

In a study performed by Peizer, Wright and Freiberger (1964), it was found that wheelchair stability was affected by the position of one's C.M. The front wheels lift off the ground when the C.M. mass falls outside the base of support. The longer the distance between the C.M. and the axle of the rear wheels, the greater the resistive moment arm will be (Pelzer et al., 1964). This means that backward rotation about the axle will happen more easily if the moment arm is shorter, therefore a smaller torque will be needed to cause a backward rotation. So the placement of the seat in relation to the rear wheels will play a major role in the stability of the chair.

Loane and Kirby (1985) examined the rear static stability of the conventional and light-weight wheelchair for four axle positions. The four positions consisted of two vertical and two antero-posterior seat adjustments. The stability was calculated by mounting the wheelchair on a platform. The angle at which the chair was tipping was used to evaluate the rear stability. The light-weight chair when compared to the conventional one had substantially less rear seat stability, as was previously observed by Pelzer et al. (1964). Therefore, the C.M. of the light-weight chair was closer to the main axle, which indicated that wheelchair design has a great effect on the position of the C.M. At all the positions tested the low posterior seat position was found to have greater stability (Loane & Kirby, 1985). Reports in the literature have suggested that the seat position for marathon and tracks events should be a posterior and low position (Brubaker, 1987; Burk, 1986; Schuman, 1979). Walsh (1985) added that the axle of the rear wheels should be slightly behind the hip bone, thus aligning the shoulder with the main axle which will allow forces to be applied directly downward.

Seating position and the rolling resistance

The rolling resistances are affected by the position of the seat in relation to the rear wheels. If the C.M. is closer to the rear wheels (farther from the casters) the rolling resistance will be less and will therefore require less energy to propel (Brubaker, 1986). Brubaker (1986) calculated that moving the C.M. 2.4 inches backward toward the main axle, caused a reduction in the ratio of caster to wheel rolling resistance by 6%. Hildebrandt, Voigt, Bahn, Berendes, and Kroger (1970) indirectly found that the location of the rear wheels will change the rolling resistance. In their study the location of the rear wheels was varied. Two different types of wheelchairs were used, one having the large rear wheels fixed at the rear axle and another one having the rear wheels placed at the front. Their results showed a decrease in oxygen consumption and heart rate for the wheelchair having the large wheels placed at the rear axle. The decrease observed in the physiological parameters tested was related to a higher mechanical efficiency (Hidebrandt et al., 1970). Their results may be explained by a decrease in the rolling resistance ratio with the rear wheels placed at the rear axle.

A study performed by Anderson, Brattgard, and Severinson (1978) demonstrated that less rolling resistance was found with the large wheels placed at the rear axle. In Anderson et al.'s (1978) study, the pushrims were substituted with a lever fixed to the centre of the large wheels. The propulsive forces were measured with strain gauges placed on the lever. Anderson et al. (1978) showed that less propulsive force was required with the large wheels placed at the rear which again indicated that there is less rolling resistance at this position.

Seating position and vertical displacement of the seat

As previously mentioned, the amount of force exerted on the pushrims plays a major role in performance. However, to obtain maximal force application the height of the seat is critical. In optimizing the seat height the relationship between muscle length and tension is very important. Because muscles generate greater tension at a greater muscle length, it becomes important to position the seat at a height which will assure maximal tension. The seat height is also important in stabilizing the posture but mainly to assure good contact with the pushrims (Brattgard, Lindstrom, Seberinson & Wink, 1984). The height of the seat depends on several variables but it is mainly a function of the arms and trunk lengths (Brubaker, McClay & McLaurin, 1983; Walsh, 1985). The activity in which the individual is engaged will also dictate the position of the seat. For wheelchair racing. ideally the individual should be able to contact the pushrim from top to bottom. without contacting the underarm area (Walsh. 1985).

Seating position and photographic analysis

A static analysis of racing wheelchairs was performed by Higgs (1983) during the 1980 Olympic Games for the Disabled. A photographic analysis was done on 49 wheelchairs used during the 1980 Olympics. In his evaluation Higgs assumed that the most successful athletes possessed the best racing wheelchair design. With this assumption. wheelchair design was related to success and the distances raced. Greater success was observed with wheelchairs having lower seat position and a higher seat base angle. The seat was located 9.22 cm up from the main axle with a seat base angle of 13.0 degrees from the horizontal (Higgs. 1983). Rudwick (1979) and Spooren (1981) stated that athletes preferred a seat base angle of 7 to 10 degrees which is a little less than the observation made by Higgs. A difference was observed in the type of events considered. If the individual was involved in a long distance event. the seat had a lower and a more posterior position than in a sprinting event (Higgs. 1983). The seat height of the long distance racers and the sprinter wheelchair were 3.01 cm and 6.84 cm up from the main axle respectively. while the rear edge of the seat was located at 11.29 and 7.49 cm posterior to the main axle respectively.

A similar study was performed by York and Kimura (1987). a photographic analysis was done on 24 wheelchairs which were used in the 1984 International Games for the Disabled. The procedures employed were essentially the same as those used by Higgs (1983). The wheelchair design was related to the distance raced and the disability level. A comparison between CP4 (cerebral palsy class IV) and AMP (amputee) was performed. York and Kimura (1987) found that the AMP group used a lower and posterior seat position than did the CP4 group. regardless of the distance raced. Thus. the C.M. of the AMP group was lower resulting in an increase in stability (York & Kimura. 1987). They also found that no significant differences in wheelchair design existed between the long distance racers or the sprinters, this is in contradiction with Higgs (1983). York and Kimura (1987) investigation leads one to think that there might exist a link between the design of the wheelchair with regards to the level of disability.

Seating position and the performance

The relationship between seat position and performance was studied by Brubaker, McLaurin, and Gibson (1980). The level of efficiency and the torque produced during wheelchair propulsion were examined at nine combinations of horizontal and vertical seat positions, using four able bodied males. The test was conducted on a motor driven wheelchair dynamometer at a constant power output for each subject. The oxygen consumption was monitored throughout the experiment. The torque was measured by means of strain-gauges bounded to a tube connected to the pushrims. Brubaker et al. (1980) found that the middle-middle and the middle-forward seat positions were observed to have higher efficiency levels and the peak torque values were lower for those positions. A greater propulsive phase was observed for the middle-middle position but this advantage seemed to be offset by the application of force along a downward trajectory in the middle-forward position (Brubaker et al., 1980). Brubaker et al. (1980) found that the product of stroke frequency and mean torque decreased with efficiencies and that the duration of the propulsive phase increased with efficiencies. Brubaker et al. (1980) also observed an increase in push angle as a result of a lower seat position. The longest push angle occurred at a middle height, 17.78 cm from the main axle.

In another study, Brubaker, Ross, and McLaurin (1982) looked at the generation of forces in relation to the vertical and horizontal displacement of the seat. In their study, eight disabled males propelled their wheelchair on a motor driven dynamometer. Nine combinations of seat positions were tested at a constant power output. The amount of forces applied to the pushrim were found to vary considerably with the different seat positions. More forces were generated with a combination of a posterior and low or middle seat position. The posterior position referred to the rear edge of the seat being 15.5 cm behind the main axle. The low and middle seat height referred to the upward distance from the main axle, 10 cm and 15 cm respectively.

Brubaker, McClay, and McLaurin (1984) determined the quantitative relationship between the seat position and mechanical efficiencies based on the subjects characteristics. Nine combinations of horizontal and vertical seat positions were tested for two modes of propulsion: the pushrim and the lever type of wheelchair. Each of the nine subjects propelled their wheelchair at a constant power output on a motor driven wheelchair dynamometer. The amount of torque applied to the wheels was recorded and the oxygen consumption was measured. The efficiency of propulsion was then correlated with the anthropometric and performance factors which included weight, sitting height, upper extremity segment length, grip strength and pushing strength. A regression analysis revealed that the prediction of efficiency for the pushrim type of wheelchair, was found to be related to the pushing strength, the upper extremity length and the horizontal distance from the shoulder to the wheel hub. The lever type of wheelchair efficiency could be predicted from the strength, the shoulder width and the segment length. Since the lever mode of propulsion was not sensitive to a change in seat position it was thought that the subject could adapt from a wider range of seat positions without affecting their efficiency (Brubaker et al., 1984).

Walsh et al. (1986), studied the effect of seat position on the maximal linear velocity in wheelchair sprinting. Nine male subjects were filmed at nine different seat positions which were commonly used by wheelchair racers. Walsh et al. (1986) examined the seating positions in relation to the maximal linear velocity, time-off and time-on the pushrim and the pushing frequency. For all the subjects tested the linear velocity did not significantly vary for the seating positions examined. The investigators claimed that the range of seat positions tested might not have been large enough to produce any differences. Also Walsh et al. (1986) stated that the subjects were able to compensate the effect of poor seating positions by changing the degree of upper body lean. This investigation showed that the linear velocity was not affected by the seat position, but that the energy expenditure might have been affected due to the variation of time-on the pushrim. Since the oxygen consumption was not taken the researchers were unable to relate the energy expenditure with the time spent in the contact phase. Also, the subjects investigated by Walsh et al. (1986) ranged from quadriplegics to low paraplegics which might explain the non significant differences observed due to high variability among the subjects.

Through varying the seat position, the investigators reviewed hoped to place the subject in a position which would allow the force to be applied to the pushrim in the most efficient manner. The major drawback of these investigations was that the anthropometric variability and the neurological level of lesions of the subjects was not taken into account when the seat position was varied and most importantly they used a small numbers of subjects. Therefore, there is a need to conduct research in this area in order to determine the optimal seat position.

Wheel camber

The wheel camber represents the angle made with the vertical axis when the wheels are inclined toward the frame. This angle has three beneficial effects. It increases the stability of the chair and makes turning easier and quicker (Burk, 1986; Walsh 1985; Bair, 1982). Also, cambering the wheels eliminates the elbow spread, therefore the subject can assume a more comfortable position which is less tiring and can therefore, reduce local fatigue by minimizing wasteful movement (Hilbers & White, 1987; Spooren, 1981). It has been suggested that cambering the wheels improves shoulder alignment which results in a better transmission of forces that serve to improve propulsion (van der Woude, Veeger, and Rozendal, 1987). However, cambering can be undesirable if the angle of inclination is too great for it will impede the motion by increasing the rolling resistance (Rudwick, 1979). If a higher efficiency is desired it is important to find the optimal angle at which the wheels should be placed. Higgs (1983) observed that success in wheelchair racing was related with an angle of

7.16 degrees. However, York and Kimura (1987) observed greater wheel inclination for the two disability groups evaluated. The AMP group was found to use a smaller angle than the CP4 group, 9.33 and 14.45 degrees respectively (York & Kimura, 1987). Since Higgs (1983), York and Kimura (1987) performed their evaluation in a static way, doubt remains as to what are the variables that affect the wheels angle.

A review of literature related to wheelchair design clearly indicates that more research needs to be conducted in this area. None of the research to date has been able to provide guidelines on how to ameliorate wheelchair design to ultimately improve the technique of propulsion and the performance.

Electromyography and wheelchair propulsion mode

Wheelchair performance will most likely be influenced by the muscle function. The muscle function is affected by the position of the user in relation to the pushrim, the resistive forces, and the level of disability (Ross & Brubaker, 1984). The studies performed into Electromyography (EMG) have attempted to identify the major muscles which are responsible for the propulsion.

Electromyography for lever propulsion

Cerquiglini, Figura, Marchetti, and Ricci (1981), investigated the muscles used for wheelchair propulsion. In their study they substituted the pushrim with a brace fixed at the centre of the rear wheel. Strain-gauges were mounted on the brace to measure the tangential and radial forces applied to the brace. An electrical goniometer was fixed on the brace to measure the rotation. The subjects were filmed and electromyographic measurements were recorded for the upper limb. This setting provided information on the muscle responsible for the generation of forces. It was found that during the propulsive phase the deltoid anterior, pectoralis major and biceps brachii were mostly involved. The triceps were found to be recruited only at the end of the push phase.

Brubaker, McLaurin, and McClay (1985) measured the EMG activity with a lever arm for a combination of nine horizontal and vertical seating positions. The subject propelled the wheelchair at a constant velocity at a power output of 15 Watts on a motor driven wheelchair dynamometer. The torque was monitored by a torque hub, the oxygen consumption was measured, and the limb segment positions were determined with stereometric position sensors. The investigators related the EMG activity with the efficiency and the seat position. Brubaker et al. (1985) observed that a middle-middle and middle-posterior seat position had an overall lower EMG activity, which was reflected in higher efficiency. The investigators also observed that a greater mechanical efficiency was found when the deltoid anterior, the pectoralis major and the lateral head of the triceps were mainly recruited. A lower mechanical efficiency was associated with higher muscular activities in the recovery phase. It was also noted that excessive angle in arm abduction, internal rotation and arm flexion was consistent with a decrease in efficiency (Brubaker et al., 1985). However, the lever or brace propulsion does not truly represent the real pattern of propulsion, therefore care must be taken when extrapolating these results (Brubaker et al., 1985; Cerquiglini et al., 1981) to pushrim propulsion.

Electromyography for pushrim propulsion

Harburn and Spaulding (1986) studied the muscles activity and the pattern of muscle recruitment during pushrim propulsion. A conventional (Everest and Jenning Premier model) wheelchair was employed to test the muscular activity of quadriplegics, paraplegics, and non disabled subjects. The subjects propelled the wheelchair on a level surface at a speed of one cycle per second which was monitored by a metronome. The movement was videotaped and synchronized with the EMG pattern. The linear envelope was ensemble and normalized by time and by the isometric maximal voluntary contraction to allow intersubject comparison. The results showed a wide variability of muscle recruitment within the group, which indicated that the technique used for propulsion varied considerably among the subjects. The quadriplegics were found to recruit a greater percentage of muscle compared with paraplegics and non disabled subjects. However, the paraplegics were also less efficient in terms of muscle activity when compared with the non disabled group (Harburn & Spaulding, 1986). It was observed that muscles most involved during wheelchair activity were the middle deltoid, posterior deltoid and the triceps brachii. The triceps were mostly involved at the end o. .he propulsive phase and the recruitment of the triceps was not consistent among the subjects. Harburn and Spaulding (1986), claimed that wheelchair propulsion required high muscular demands and that shoulder abduction appears to be the cause of the high muscle requirement.

In a study performed by Ross and Brubaker (1984), the EMG activity was recorded to determine the muscles responsible for the development of torque. Wheelchair propulsion was simulated on a motor compensated wheelchair dynamometer for paraplegics and non disabled subjects. The pushrim torque was monitored by means of strain-gauges bounded on a tube connected to the pushrim. The muscles corresponding to the propulsive forces and the development of torque were then determined. The shape of the torque curve was found to be bimodal. Therefore, the movement which was used to propel was a push-pull type of motion (Ross & Brubaker, 1984). The brachioradialis and the pectoralis major were active throughout the torque phase. The biceps brachii (long head), deltoid anterior, serratus anterior and upper fiber of the trapezius were responsible for generating the torque at the beginning. At the end of the torque phase the biceps brachii (long head), triceps brachii (lateral head), deltoid posterior and upper fiber of the trapezius were active. During the recovery phase the biceps brachii (long head), triceps brachii (lateral head), deltoid posterior and upper fiber of the trapezius were activated. At the beginning of the recovery phase the serratus anterior was recruited and at the end the deltoid anterior and the brachioradialis were recruited. Considerable individual variations in EMG pattern were observed mostly for the serratus anterior and the upper fiber of the trapezius (Ross & Brubaker, 1984). No apparent differences in the EMG pattern were observed for the paraplegics and the non disabled subjects. However, the paraplegics were found to exhibit a smoother and more consistent stroke pattern, which was thought to be related to practice (Ross & Brubaker, 1984).

The above investigations used a small number of subjects, various types of wheelchairs (conventional, lever and racing wheelchair) and various levels of neurological lesion. Therefore, a generalization of the above result can only be tentative. A review of the literature related to EMG and wheelchair propulsion clearly indicates a lack of knowledge in this area. It then becomes important to conduct such research in this area but with subjects having similar lesion levels to be able to draw any conclusions.

METHODOLOGY

Subjects

Five male paraplegics with neurological lesions ranging from T11 to L2 (class IV, McCann, 1979) and aged between 19 and 37 (years) were asked to participate in this investigation. The subjects were selected from the Ottawa region on a voluntary basis. Furthermore, the subjects exercise lifestyle was used as a criterion for the selection (active individuals were selected first). Any subjects with known pathological disorders of the upper extremities were not used in this experiment.

Wheelchair roller

The simulation of wheelchair propulsion was performed on stationary rollers. Due to the hollow characteristic of the rollers the moment of inertia was very small. Therefore, to improve the inertial properties of the rollers and to better simulate actual locomotion, the mass of each roller was increased by adding two iron rings on each of the rollers. Each ring had a mass of 2.72 kg and a moment of inertia of 0.052 kg.m^2. The ring's inertia was mathematically calculated. By adding the rings on the rollers the system's rotational inertia was augmented and therefore the roller's angular momentum properties were increased.

To monitor the speed of rotation a tachometer was used. The shaft of the tachometer was supported on an arm which was fixed on the side of the rollers (see Figure 1). Between the shaft and the support arm joint rotation was possible. A small wheel was embedded on the shaft and rested on the roller. The wheel was surrounded by rubber to prevent any slippage between the roller and the tachometer. In using this system, any rotation of the wheel was translated into the tachometer through the rollers. For each rotation the tachometer registered the difference of potential. The difference of potential was then

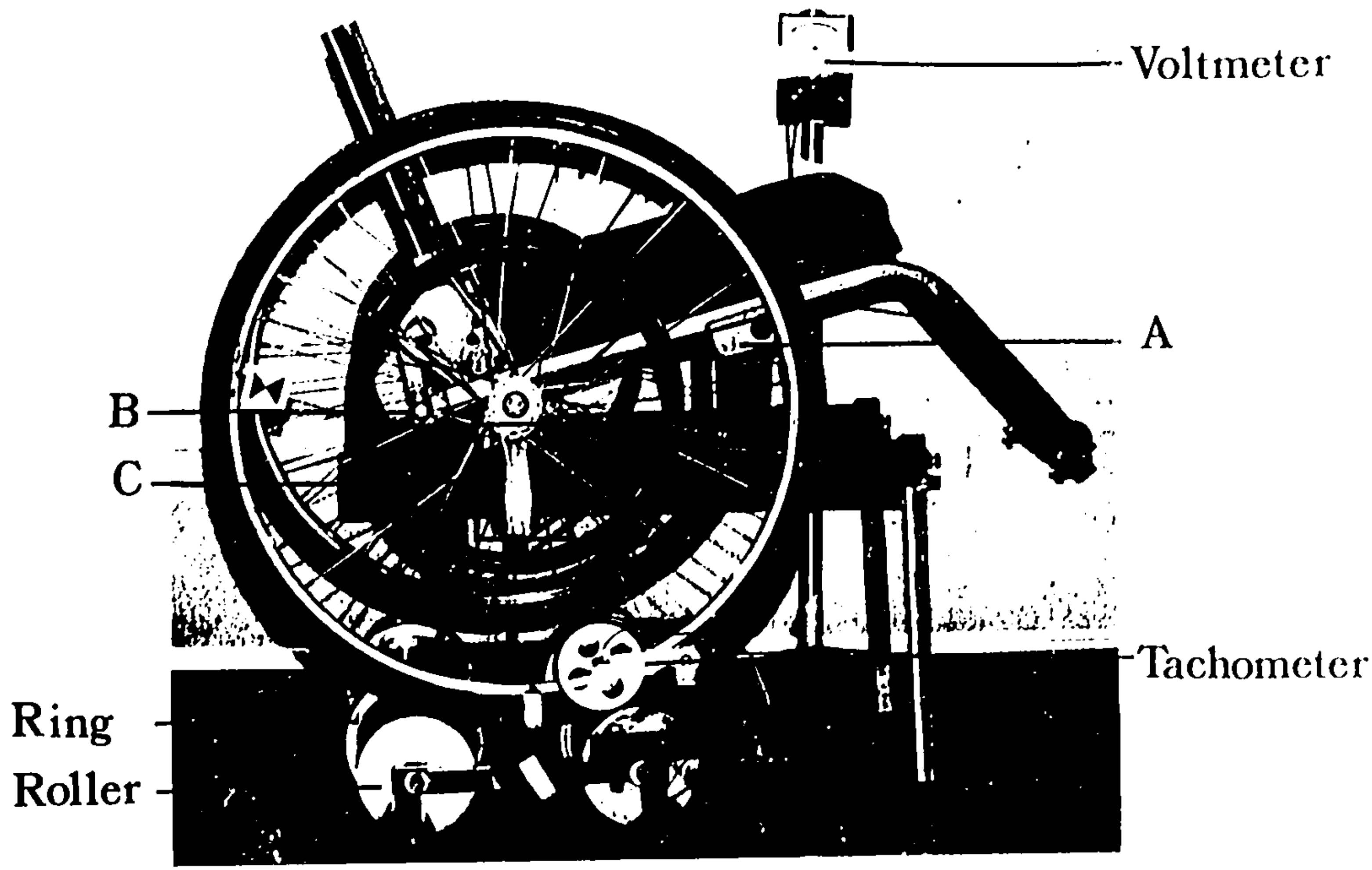

Figure 1: Testing wheelchair and rollers

recorded on a microcomputer (Compaq 386, 16 MHZ) using a data acquisition system (Lamontagne, M. Bradley, D. & Lemaire, E.D., 1989).

The tachometer was calibrated with an electronic counter to convert the voltage into speed (m/s). The counter was mounted on a small piece of wood which extended in front of the roller and a piece of reflective paper was taped on the roller. The counter recorded each time the reflective tape passed in front ofthe counter, which corresponded to one complete revolution. The calibration consisted of spinning the roller at various speeds and to monitor the counter and tachometer signals. The electronic counter and the tachometer signals were stored on microcomputer (Compaq 386, 16 MHz) through a data acquisition system (Lamontagne et al., 1989) to digitally convert the signal. The data were collected at 50 Hz for a period of 20 seconds for 10 trials. Then, the tachometer data were filtered at a cutoff frequency of 4 Hz (single pass critically dampened digital filter). The conversion factor (voltage into speed (m/s)) was then easy to calculate since we knew the distance and time required to perform one revolution (each time the roller completed one revolution the counter was increased by 1) and the corresponding voltage. Therefore, the following equation was used to calculate the speed of rotation:

$$\text{speed} = d \,/\, (T_{(n+1)} - T_{(n)})$$

where

d= distance for one complete revolution (0.4021 meters)
$T_{(n)}$= time to complete revolution n.
$T_{(n+1)}$= time to complete revolution n + 1.

The speed and the voltage value were least square fit to calculate the equations of the curves. The least square fitting served to generate an array of speed and volt value for equivalent time. Thus, the slope was a simple ratio of the speed and the voltage:

$$Slope_{(n+1)} = speed / voltage$$

where

$$Slope_{(n+1)} = \text{slope at each data point n.}$$

Then, for each trial the mean slope and the coefficient of variation was calculated. The within and between trial varibility of scores was expressed as a proportion of their mean by the coefficient of variation (CV):

$$CV = (SD/mean) \times 100\%$$

From Table 1A, it was calculated that the within trial variability was less than 2%. The between trial variability showed that less than 5% of variation was associated with the conversion factor (1.16 (m/s)/volt).

Table 1B

Calibration of the tachometer

Trial	Mean slopes (m/s)/volt	Standard deviation	Coefficient of variation
1	1.20	.0197	1.64%
2	1.24	.0154	1.24%
3	1.17	.0203	1.74%
4	1.23	.0128	1.04%
5	1.10	.0170	1.54%
6	1.07	.0191	1.78%
7	1.19	.0012	0.99%
8	1.11	.0189	1.70%
9	1.14	.0154	1.45%
10	1.13	.0153	1.36%
Mean		1.158	
S.D.		0.057	
C.V. (%)		4.92%	

The speed could be made visual to the subject through a voltmeter, as shown in figure 1. To help the subject know their actual speed of propulsion the volt scale on the volmeter was converted into speed (m/s) using the conversion factor previously calculated.

Wheelchair

A special wheelchair was designed by Advance Mobility System Corporation (AMS) for this experiment, which allowed different adjustments. The following adjustments were possible on the testing wheelchair; the seat height and the horizontal position of the rear wheels; the wheel camber; and the seat base and back rest inclination as shown in Figure 1.

All the seat adjustments are made by moving part A and B of the chair (see Figure 1). In moving only part A the angle between the seat base and back rest is changed. If part B is raised alone the seat base angle is varied. To vary the seat tilt part A and B need to be moved in different directions (up and down), and moving both parts the same direction caused a variation in seat height.

The wheel is mounted on an axle block (see Figure 1 part C and Figure 2 for a detailed illustration of part C) which can be moved horizontally on the supporting frame. To move the axle block the four thightening screws need to be loosened then the axle block can either be moved forward or backward on the frame (see Figure 2). The axle of the wheel is fixed to a metal support which is then mounted to the axle block as shown in figure 2. This double joint connection was used to rotate the axle in order to vary the wheel camber. The wheel camber adjustment is done by moving the screw which connects the plate camber extension and the axle block extension, both extensions serve to secure the inclination of the wheels.

As shown in Figure 1, racing wheels were mounted on the axle of the wheelchair and pushrims were fixed on the spokes. The mechanism used to attach the pushrims on the spokes of the wheels is illustrated in Figure 3. In using this

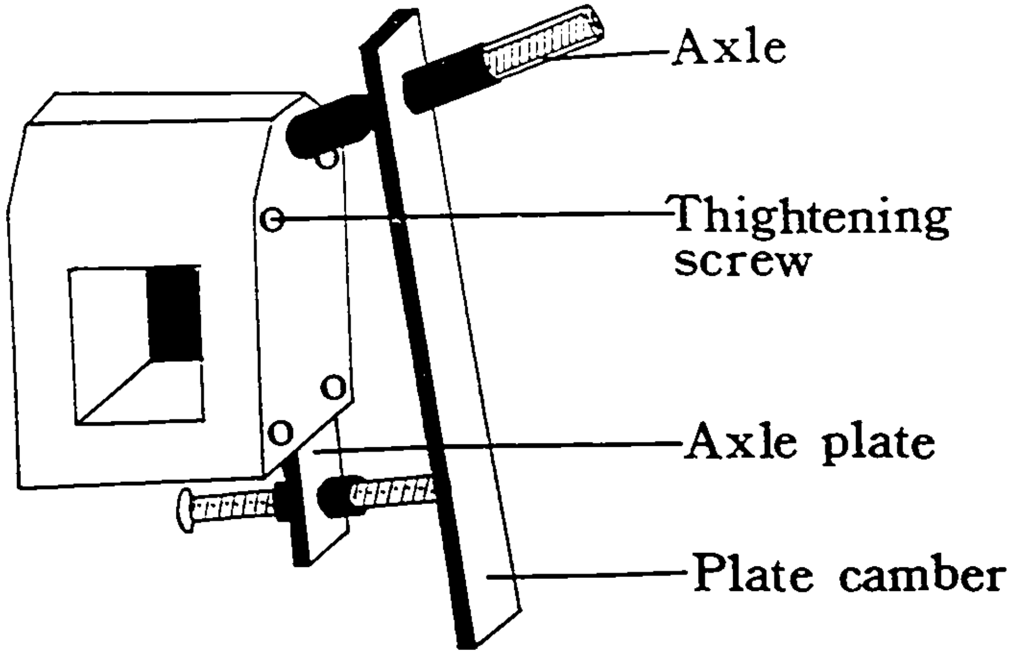

Figure 2: Axle block

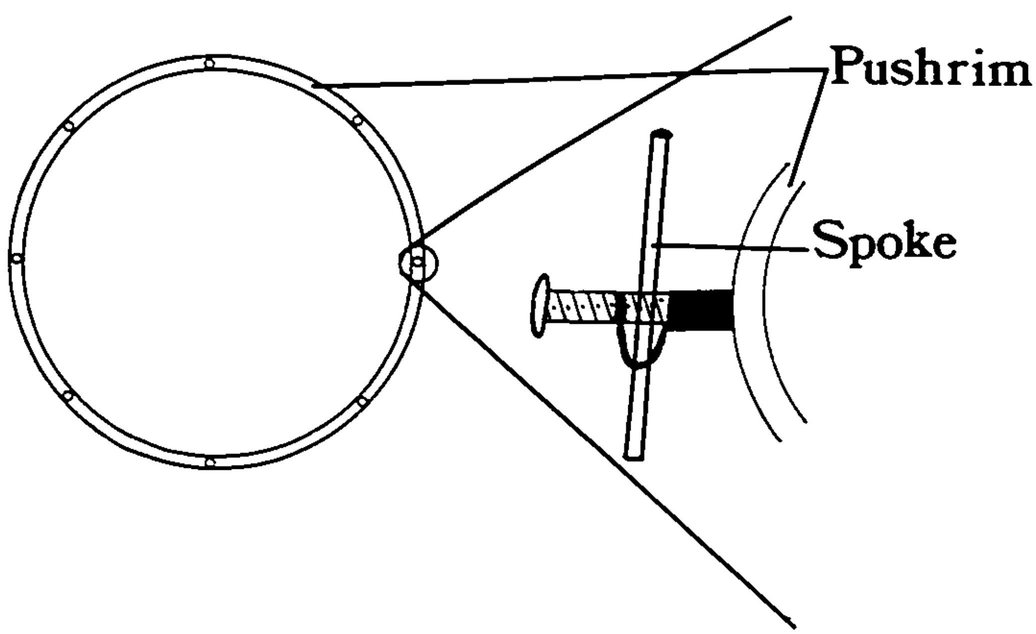

Figure 3: Pushrim attachments

wheelchair design the size of the wheels and the pushrims can be easily changed and various wheelchair design can easily be simulated.

Subject's position in the wheelchair

Six experimental conditions were evaluated for this experiment: three horizontal positions of the rear wheels (Forward, Middle, Backward) at two sitting heights (High and Low). The position of the rear wheels was determined according to the centre of mass (C.M.) of the wheelchair. The position of the C.M. was obtained using a modified reaction board technique (Lemaire, Lamontagne & Barclay, 1989). The C.M. positions were located in front of the main axles. For the Forward, Middle and Backward position of the rear wheels, the C.M. was located at 10 cm, 6 cm and 3 cm in front of the main axles respectively. This also corresponded to a seat base and back rest intersection located at 7.6 cm, 4.4 cm, and 2.2 cm behind the main axles respectively. Since the position of the C.M. was not reported in the literature reviewed, the selected position represented a range between the conventional and racing wheelchair which was reported by Peizer et al. (1964).

The High and Low positions were established by moving the seat in the vertical direction. Both positions depended on the subject's arms and trunk lengths. 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 case the lowest position depended on the design of the wheelchair, but for all the subjects it was found to closely correspond to the desired position (distal phalanges of the second finger is aligned with the lowest portion of the pushrims). The High position was set at 10% of the subject's arm length above the Low position. The Low and High positions were selected because they were believed to represent a range of positions which were often used by wheelchair athletes (Higgs, 1983; York & Kimura, 1987). For each experimental condition the wheel camber, the seat base and, the back rest angle remained constant. These positions were determined according to Higgs (1983); York and Kimura (1987) as being those positions most often used by long distance racers and also were found to correlate with a high level of success. The wheel camber was set at 8 degrees from the vertical; the seat base was set at 12.5 degrees from the horizontal; and the back rest was set at 90 degrees from the seat base.

Cinematographic technique

The kinematics of the upper body was recorded by filming the subjects at 50 frames per second using a 16 mm cinecamera (Locam II).

Table 2B
Marker's landmarks

Marker	Marker Position
Head	Centre of the ear
Shoulder	Greater tubercle of the humerus
Elbow	Lateral epicondyle of the humerus
Wrist	Styloid process of ulna
Finger	Third distal metacarpal bone
Rib	Midline of the rib cage halfway between the iliac crest and the shoulder.
Hip	Greater trochanter of the femur
Wheel	On the inside portion of the tire

Markers were placed over selected anatomical landmarks on the right side of the body as shown in Table 1B. A marker was also placed on the wheel to determine the velocity of propulsion. The camera was mounted perpendicular to the subject's right sagittal plane at a distance of approximately 11 meters away.

Electromyographic technique

The muscular activity during wheelchair propulsion was recorded for 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 because they are mainly recruited during wheelchair propulsion (Harburn & Spaulding, 1986; Ross & Brubaker, 1984). The motor point of each muscle was determined according to Delagi, Perotto, Iazetti and Morrison (1975). Disposable surface silver/silver chloride electrodes (Medi-trace) were placed in the main direction of the muscle fiber with approximately 2.5 cm apart both centres of the electrodes. The muscular activity was recorded simultaneously with the cinefilm for each trial. The raw Electromyographic (EMG) signals were recorded at 1000 Hz for three consecutive cycles and each trial for a period of 5 seconds. The signals were fed to a bioamplifier (University of Ottawa, input impedance of 10 megohms, 10-700 Hz bandpass), converted in digital value by a data acquisition system (BIOAD system, Lamontagne, Bradley, Lemaire, 1989), and then stored in the memory of a microcomputer (Compaq 386, 16 MHz).

Procedures

Prior to the experiment each subject was asked to sign an informed consent (see Appendix B). Markers were placed over the anatomical landmarks as described in Table 1B. Then, the subject's anthropometric measurements of the upper body were recorded according to Jette's procedure (1983). Then, the Low sitting position was measured with the shoulder aligned with the main axles. The subject was asked to keep his back against the back rest and to keep his head straight. The arms were lying against the pushrim. The seat was then moved until the subject's distal phalanges of the second fingers were aligned with the lowest portion of the pushrims.

EMG electrodes were fixed on the subject's skin. The skin preparation consisted of rubbing with alcohol, shaving the skin until it turned slightly pink, and rubbing electrolyte paste over the appropriate area in order to reduce the skin impedance. The surface electrodes (Medi-trace) were fixed over the motor point and the skin impedance was measured with a ohmmeter. if the impedance was higher than 2 kilohms the skin was cleaned again and new electrodes were placed. The electrode wires were taped to the skin and the leads were fixed to the skin or clothing in order to reduce movement artifacts and to allow freedom of movement.

Each subject was then trained with the testing wheelchair on the rollers until the subject felt comfortable with the equipment. Then, the subject's maximal speed of propulsion was obtained by incrementing the speed over a five minute period with the maximal speed being recorded as the highest speed attained by the subject. The subject was asked to hold his maximal speed for approximately three seconds. Each subject repeated this procedure three times. For each trial the average speed of propulsion was calculated with the computer. The three trials were then averaged and 60% of this speed was used in the remainder of the experiment. The maximum voluntary isometric contraction for each muscle was then recorded. For each muscle the subject was asked to perform a maximum voluntary contraction against a resistance. The subject was asked to hold his contraction for a period of about 5 s, the signals were recorded with a data acquisition system (Lamontagne et al., 1989) and were stored on a microcomputer (Compaq 386, 16 MHz).

Finally, the subject was asked to propel the designed wheelchair at each experimental condition FH (Forward and High), FL (Forward and Low), MH (Middle and High), ML (Middle and Low), BH (Backward and High) and BL (Backward and Low) at 60% of their maximal speed of propulsion. The speed of propulsion was monitored with a tachometer and was recorded on the computer and was visually available to the subject by a voltmeter. The subject was randomly assigned to each experimental condition. 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 s. For each trial, the subject's technique of propulsion was filmed and EMG synchronized. In between trials the subjects had approximately 2 min of rest.

Data reduction and analysis

The pushing phase and recovery phase were determined from the speed of the rear wheel as measured by a tachometer. The pushing phase represented the time where forces were applied to the pushrims, which was reflected in an increase in speed of the rear wheel. The recovery phase represented the time where no forces were applied to the pushrim and was characterized by a decrease in speed of the rear wheel.

All the EMG data were processed on a microcomputer (Compaq 386, 16 MHz) with the BIOPROC program (Lamontagne, Bradley & Lemaire, 1989). The bias of the raw EMG signal was removed by mean, full-wave rectified, and filtered with a 6 Hz single pass critically dampened digital filter to obtain the linear envelope (LE EMG) of the signals. For each cycle, the LE EMG signal for each muscle was normalized over time (%) and by amplitude (MVC). The LE EMG ensemble average of each condition was calculated from 45 cycles (5 subjects, 3 trials and 3 cycles).

Each film trial was digitized using a Hewlett-Packard 9874A digitizing tablet and the raw data was then transferred to a main frame computer (Amdahl IBM VM/SP, University of Ottawa) for complete data processing. The filmed data were projected at 9.3% of the life-size image and was digitized to an accuracy of 1.98 mm. Finally, the data were digitally filtered (second-order Butterworth low-pass filter) with a 6 Hz cut off frequency, to randomly attenuate noise (Winter, 1979). The filtered absolute data were processed by the BIOMECH package (University of Ottawa) to calculate the kinematic parameters such as; linear and angular velocities, accelerations and momentums. The contact angle, release angle, and degrees of contact were also computed from the kinematic data. Simple trigonometry was used to calculate the contact and release angle, which correspond to the formula used by the BIOMECH program to calculate the absolute angle. Then, for each condition and subject the data were normalized over time 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 averages were in turn averaged across all subjects (n=5) to yield a grand ensemble normalized average (n=15, 3 trials per subject times 5 subjects) for each condition. The segment mass, centre of mass and, the radius of gyration were calculated from Dempsters (1955) table.

Both descriptive analyses and descriptive statistics were conducted on the EMG and kinematic data. The descriptive analysis was performed to show the differences in the amplitudes of the kinematics and the normalized LE EMG among the six seating positions. The descriptive statistics were used to complement the descriptive analysis. The descriptive statistics used for the EMG consisted of integrating the within-subject (n=9) normalized average LE EMG using trapezoidal integration. Then, these integrated EMG (IEMG were ensemble averaged across subjects (n=5) to give a grand ensemble normalized IEMG (n=45, 9 trials per subject times 5 subjects) for each condition. For the kinematic data the descriptive statistics comprised the comparison of the cycle time, the push phase, the recovery phase, the pushing frequency, and the degrees of contact. A grand ensemble averaged (n=15, 3 trials per subjects times 5 subjects) was also computed on each of the above parameters for each condition. Also, an observation of the averaged linear and angular velocities, accelerations, and momentums, and LE EMG curves were used to further contrast the seating positions. The LE EMG curves were used to compare the muscle activity with respect to temporal percentage of cycle, cycle phase and percentage of activity among the seating position.

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