Effect of choice of recovery patterns on handrim kinetics in manual wheelchair users with paraplegia and tetraplegia

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Background: Impact forces experienced by the upper limb at the beginning of each wheelchair propulsion (WCP) cycle are among the highest forces experienced by wheelchair users.

Objective: To determine whether the magnitude of hand/forearm velocity prior to impact and effectiveness of rim impact force are dependent on the type of hand trajectory pattern chosen by the user during WCP. Avoiding patterns that inherently cause higher impact force and have lower effectiveness can be another step towards preserving upper limb function in wheelchair users.

Methods: Kinematic (50 Hz) and kinetic (2500 Hz) data were collected on 34 wheelchair users (16 with paraplegia and 18 with tetraplegia); all participants had motor complete spinal cord injuries ASIA A or B. The four-hand trajectory patterns were analyzed based on velocity prior to contact, peak impact force and the effectiveness of force at impact.

Results: A high correlation was found between the impact force and the relative velocity of the hand with respect to the wheel ($P < 0.05$). The wheelchair users with paraplegia were found to have higher effectiveness of force at impact as compared to the users with tetraplegia ($P < 0.05$). No significant differences in the impact force magnitudes were found between the four observed hand trajectory patterns.

Conclusion: The overall force effectiveness tended to be associated with the injury level of the user and was found to be independent of the hand trajectory patterns.

Keywords: Spinal cord injuries, Kinetics, Paraplegia, Tetraplegia, Wheelchair, Propulsion, Biomechanics

Introduction

Wheelchair propulsion (WCP) is an essential component of activities of daily living (ADL) for individuals dependent on the use of a manual wheelchair for mobility. Depending on the age of the wheelchair user and the level of spinal cord injury, WCP has been found to involve anywhere from 1000 to 2000 repetitive hand contacts with the push rim of the wheel during the course of a day.¹⁻³ The impact forces between the hand and rim range from 30 to 40 N during normal propulsion and 120 to 150 N during graded propulsion.⁴ The generation of reaction forces at the hand-rim interface during WCP involves a significant work load for the upper limbs.⁵ The repetitive mechanical loading imposed on the upper limbs causes muscle fatigue and

can lead to reduction in scapular humeral and scapular thoracic muscular control.⁶ Over time, repetitive loading under diminished muscular control conditions can lead to injury that impairs an individual's ability to maintain ADL.⁷ The greater susceptibility to pain or injury of the upper limb⁸ in manual wheelchair users with cervical level spinal cord injury (SCI) is often attributed to the reduced number of innervated muscles available for controlling the trunk and upper limbs.

The largest and potentially most injurious reaction force during the propulsion cycle generally occurs immediately after the hand initiates contact with the pushrim. This initial hand contact marks the end of the recovery phase and beginning of the push phase. The recovery phase of WCP has been studied and four types of hand movement patterns have been identified, based on the kinematics of the hand or the wrist relative to the rim during the propulsion cycle^{9,10} (Fig. 1). These

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Figure 1 Classification of wheelchair propulsion techniques based on hand movement trajectories. The thicker portion of the hand trajectory represents the pushrim (push cycle) and the thinner line represents the recovery path of the wrist joint. Pattern names are based on number of loops and trajectory shape: SLOP: single-loop-over propulsion, SC: semicircular, DLOP: double-loop-over propulsion, ARC: arcing.

movement patterns include semi-circular (SC), double-loop-over propulsion (DLOP), single-loop-over propulsion (SLOP) and arcing (ARC) kinematic trajectories of the hands.

Hand movement patterns have been used to identify an individual's pushing technique and were compared relative to the range of the upper limb joint motions, acceleration of segments, percentage of time spent in the push phase, propulsion efficiency and effectiveness of the rim forces during the push phase.^{9–13} Shimada *et al.*¹⁰ in their study of experienced wheelchair users found the SC style to be optimal in terms of lowest joint accelerations during propulsion. Boninger *et al.*¹¹ agreed with the above findings and also favored the SC pattern because, relative to other patterns, it involves fewer repetitions for the same distance traveled. The ARC pattern was determined to be the most preferred while propelling uphill.¹³ The ARC pattern was also found to have the lowest metabolic cost.¹²

The existing studies on hand movement patterns have been limited to non-wheelchair users and individuals with paraplegia. The wheelchair users with tetraplegia experience a loss of strength and control due to partial or complete loss of innervation of the muscles involved in WCP and hence are expected to display different propulsion characteristics than the users with paraplegia. How these patterns are related to the magnitude of the reaction forces applied to the hand immediately after pushrim contact has not been investigated in relation to pushing capabilities, WCP performance or risk of upper limb injury.

The purpose of this study was to determine (a) whether the hand movement patterns used by manual wheelchair users affect the magnitude of hand/forearm velocity prior to contact with the pushrim, (b) whether one hand movement pattern results in a higher effectiveness of propulsion than other patterns and (c) whether during propulsion, participants with tetraplegia exhibit patterns that differ from individuals with paraplegia. Differences in the hand movement patterns were expected to result in differences in the velocity of the hand immediately before contact with the pushrim. We hypothesized that the velocity of the hand at contact with the pushrim would be directly related to the magnitude of the impact force experienced by the wheelchair user. We also hypothesized that the wheelchair users with tetraplegia will show relatively higher impact forces than the wheelchair users with paraplegia. These hypotheses were tested by comparing the magnitude and effectiveness of the impact force between the four-hand movement patterns observed during WCP on an ergometer by the participants with paraplegia and tetraplegia.

Methods

Individuals with SCI from the outpatient clinics throughout the Rancho Los Amigos National Rehabilitation Center (RLANRC) were asked to participate. All subjects were free of shoulder pain and rotator cuff pathology was ruled out with Jobe's Empty Can test,^{14,15} Codman's Drop Arm test^{16,17} and resisted external rotation.¹⁷ Individuals were excluded from participation if they reported a history of shoulder pain that altered the performance of daily function or required medical treatment. Thirty-four (31 males, 3 females) experienced wheelchair users with complete motor paraplegia or tetraplegia who met the inclusion criteria agreed to participate (Table 1). Sixteen participants had thoracic and lumbar level (T6–L1) SCI and 18 participants had cervical (C6–C7) level SCI, all had motor complete spinal injury ASIA A or B. The participants had an average weight of 74.5(12) kg and average height of 1.75(0.06) m. The participants were asked to read and sign a statement of informed consent approved by the Institutional Review Board at RLANRC.

Table 1 The number of participants in each group using each hand movement pattern

	ARC	DLOP	SC	SLOP
Paraplegia	2	4	1	9
Tetraplegia	7	0	7	4

The data collection was performed in the participants own wheelchair. The wheelchair-participant combination was positioned on a stationary ergometer which consisted of a supporting frame and split rollers so that the participant was required to propel each wheel separately. The rollers were coupled, by means of a differential to an alternator to provide resistance. The resistance was controlled by a modified Velodyne (Schwinn Bicycle Company, Chicago, IL, USA) bicycle-ergometer computer. Removable flywheels proportional to the weight of both the participant and the wheelchair were used to simulate the translational inertia of 'over ground' propulsion. Hydraulic jacks were placed under the frame of the wheelchair to support some of the weight of the participant and the wheelchair, thereby helping to adjust the friction between the tires and the rollers. To determine the friction force between the tire and the ergometer roller, a coast down test was performed with the participant sitting in the test wheelchair on top of the ergometer. The coast-down test involved the following: the two independent rollers were locked (by locking the differential) and spun by an electric drill motor until the wheels were spinning at 182 m/minute as indicated by the ergometer controlling computer. The drill motor was then removed and the wheels were allowed to coast down to 35 m/minute. The ergometer controlling computer used the friction force, recorded during the coast-down test, to determine the resistance applied to the rollers in subsequent data acquisition trials. The data were collected while each participant performed WCP using their preferred propulsion technique at their customary speed with friction level set similar to over ground (tile surface) propulsion. The participants propelled the wheelchair for at least 30 seconds immediately before beginning the data collection to achieve near steady-state propulsion. Ten-second trials (6–9 propulsion cycles) of WCP were recorded.

The three-dimensional motion of the trunk and the right upper limb was tracked using a Vicon^c Motion Analysis system. A six-camera system was used to track 16 reflective markers placed on landmarks on the body. Markers were placed on the trunk at the manubrium, the xiphoid process, the spinous processes of T3 and T10 vertebrae, greater tubercle of the humerus, lateral epicondyle, medial epicondyle, deltoid tuberosity, mid forearm, radial styloid, ulnar styloid, head of the third metacarpal and head of the fifth metacarpal. Three reflective markers were also placed on the wheel. The collection frequency for the kinematic data was 50 Hz. The kinematic data were filtered in Visual3d (C-Motion, Inc.) using a lowpass sixth-order

Butterworth filter with a cutoff frequency of 8 Hz, consistent with previous literature.¹⁸ The pushrim force applied by the hand to the pushrim within the plane of the right wheel¹⁹ was measured using three strain gauge force transducers at 2500 Hz (Smart^{Wheel}) and recorded using a PDP-11/73 computer (Digital Equipment Corporation, Cambridge, MA, USA). The force data were later sampled down using a spline to 50 Hz to match the collection frequency of the kinematic data. This was done after observing that the down sampling did not lead to a significant loss of peaks in the force data. For further details and description of the collection equipment, marker set and techniques employed refer to Mulroy *et al.*²⁰

The data were processed in Visual3D and then exported for further calculations using custom code in Matlab (Mathworks Inc., Natick, MA, USA). The segment inertia parameters were taken from de Leva, 1996.²¹ The length of the segments was calculated using the endpoints of the segments exported from the Visual3D model. Contact with the hand-rim was defined as the point when the resultant force on the wheel exceeded 5 N and release from the hand-rim was the point when the force went below 5 N. The peak resultant force during impact was divided into its radial and effective components using the following equations²² where F_{Eff} is the effective force applied on the wheel and F_{Rad} is the radial force applied on the wheel.

$$F_{\text{Eff}} = F_X \cos \alpha + F_Y \sin \alpha \sin \beta + F_Z \sin \alpha \cos \beta$$

$$F_{\text{Rad}} = F_X \sin \alpha + F_Y \sin \alpha \sin \beta + F_Z \cos \alpha \cos \beta$$

F_X , F_Y and F_Z are, respectively, the anterior/posterior, medial/lateral and inferior/superior forces applied on the pushrim. Alpha(α) is the angle between the line joining the hand and the wheel axle and the vertical axis through the wheel axle. Beta (β) is the camber angle of the wheel (set to 5°). The percentage of effective force is the square of the ratio of F_{Eff} to the resultant force and the percentage radial force is the square of the ratio of F_{Rad} to the resultant force. The percentage of effective force during the impact phase of the push cycle was calculated and compared between hand movement patterns. The impact force was defined as the first maxima observed in the resultant reaction force immediately after hand contact with the push rim (Fig. 2). The recovery patterns of the participants were visually examined by one researcher and classified into the four previously mentioned subcategories namely ARC, SC, SLOP, and DLOP.

Figure 2 Shows the resultant force profile along with the effective force and the radial force for an exemplar participant with tetraplegia.

The primary hand areas initiating rim contact for wheelchair users with paraplegia and tetraplegia were the thenar and hypothenar prominences of the hand, respectively. The wrist velocity at the end of the recovery phase just prior to hand contact was found to be more representative of the velocity of thenar and hypothenar prominence than the velocity of the center of mass of the hand. The recovery phase was defined as the interval between hand release from the pushrim and hand contact with the pushrim. The push phase was defined as the period of hand contact. The impact phase was defined as the interval between hand contact and the local minimum in the resultant reaction force immediately following the impact force.

The statistical analysis was performed in R statistical software. Descriptive statistics (means and standard deviations) were used to characterize group means for the impact force, effective force and radial force. A modified bootstrap method was used to compare the correlation coefficients between wrist velocity and magnitude of impact force. The Yuen test for independent groups was used to compare the tetraplegia group with the paraplegia group. Welch's test was used to compare the magnitude and effectiveness of the impact force of the four recovery patterns. All statistical analysis was performed in R statistical software. A value of $P < 0.05$ was considered significant.

Results

The velocity of the wrist prior to contact was found to be significantly correlated ($r = 0.74$, $P < 0.05$) with the magnitude of impact force for all 34 participants with tetraplegia ($0.81(0.24)$ m/second and $0.062(0.02)$ N/kg) and paraplegia ($0.95(0.37)$ m/second and

$0.061(0.03)$ N/kg). The correlation between wrist velocity prior to contact and magnitude of impact force normalized to body weight was found to be much stronger for wheelchair users with paraplegia ($r = 0.92$) as compared to tetraplegia ($r = 0.45$) as shown in Fig. 3.

The Yuen test for independent groups showed no significant differences in the magnitude of impact force ($P > 0.05$) between participants with tetraplegia and paraplegia. The participants with tetraplegia, however, had significantly higher radial component of impact force than the participants with paraplegia. The mean percent radial component of impact force was significantly higher ($P = 0.02$) for participants with tetraplegia (9.2%) than paraplegia (4%). On average, the participants with tetraplegia applied more radial impact force (16 N higher) than those with paraplegia.

The percent of impact force applied in the tangential direction (effective force) was significantly higher ($P = 0.005$) for participants with paraplegia (94%) than tetraplegia (88%). On average, the participants with paraplegia applied more effective force (19.8 N higher) than those with tetraplegia. These results indicate that the participants with tetraplegia have a lower effectiveness of force application at impact than participants with paraplegia, regardless of the hand trajectory used during the recovery phase.

ARC, SC, and SLOP propulsive patterns were preferred by both groups of participants (Table 1). However, the DLOP pattern was not used by the participants with tetraplegia and the SC hand movement pattern was observed in only one participant with paraplegia. The most common hand movement pattern in the combined sample population was SLOP pattern, observed in 13 of 36 participants.

Figure 3 Correlation of wrist velocity (m/second) prior to contact and peak rim force during impact normalized to arm weight (N). Symbols reflect propulsion pattern (SC (C), ARC (A), DLOP (D), SLOP (S) and spinal cord injury level (TP (Δ), PP (O)).

The impact force was not found to be significantly different ($P > 0.05$) between the hand movement patterns. The box plots of the impact force for the four hand movement patterns are shown in Fig. 4. The impact force is normalized with respect to arm weight to account for the between-subject differences in body mass.

Force effectiveness was not found to be significantly different between hand movement patterns. The range of percent effective force at contact for participants with tetraplegia was between 0% and 25% and for participants with paraplegia was 25% and 95% for all hand movement patterns. The same hand movement pattern, as seen in different participants showed different percentages of effective force at contact. For example, the ARC hand movement pattern for

participants with paraplegia showed effectiveness in the 90% range, whereas the participants with tetraplegia had less than 30% force effectiveness (Fig. 5)

Discussion

The reaction force generated at the hand–pushrim interface used to propel the wheelchair imposes significant repetitive mechanical loading on the upper limb. The highest reaction force during a push cycle is often experienced immediately after the hand initiates contact with the pushrim. In this study, our aim was to identify the hand movement patterns used during manual WCP by the individuals with tetraplegia and paraplegia, and determine whether some movement patterns result in greater impact forces or have a more effective impact force during WCP. The results indicate that the velocity of the wrist prior to contact is highly correlated with the impact force and the correlation is stronger in participants with paraplegia compared to those with tetraplegia. In this study, individual preferences for a hand pattern were not uniquely defined by the injury level of the participants. The magnitudes of impact force during WCP were related to wrist velocities prior to contact but were not inherently higher for a particular hand movement pattern. The effectiveness of impact force was found to be significantly different between the participant groups with different levels of SCI but not between the types of hand movement pattern. The wheelchair users with paraplegia demonstrated significantly greater impact force effectiveness than those

Figure 4 Normalized impact force (normalized by arm weight) for the four hand movement patterns. The line inside each box is the median, the whiskers denote the largest and the smallest values not declared as outliers and the dots are the outliers.

Figure 5 Percent effective impact force for the two user populations (P (Paraplegia) and T (Tetraplegia) in front of the technique name stands for the participant group). ARCP and SCP have only two and one data point, respectively.

with tetraplegia – consistent with between group differences in functional capacities. Kulig *et al.*²³ calculated the joint kinetics during the push phase of WCP in person with paraplegia and tetraplegia and determined that the superior push force in subjects with tetraplegia was significantly higher than in those with high paraplegia, after co-varying with velocity. The pattern of increased superior push forces and high rotator cuff muscle activity, coupled with weakness of thoracohumeral depressors seen in those with tetraplegia, has the potential to increase the susceptibility of the subacromial structures to compression and increase the risk of muscle damage, such as rotator cuff tears.

One objective of WCP during the impact phase is to create a firm contact with the push rim in order to generate sufficient force to propel the wheelchair. A high radial force during impact contributes to increasing friction between the hand and rim from the beginning till the end of the push phase. The participants with tetraplegia demonstrated significantly higher radial, but lower tangential, force during impact than did those with paraplegia. The wheelchair users with C6-level tetraplegia typically have weak grip strength that affects their ability to sustained contact between the hand and the push rim. The use of a higher radial force may serve to increase the friction needed to establish hand

contact with the rim. The lower tangential forces observed in those with tetraplegia compared to those with paraplegia are consistent with rim forces reported by Dallmeijer *et al.*²² during the entire push cycle. They found that the participants with paraplegia exhibited rim forces that were 20% more effective than those with tetraplegia. The participants with paraplegia typically have stronger grip strength than those with tetraplegia. This low FEF is mainly the consequence of a larger lateromedially directed force component, which is most likely a result of reduced muscle function in sternal pectoralis major and triceps brachii and the loss of hand grip function which necessitates extra hand rim friction to propel the wheelchair. The participants with paraplegia have the capacity to approach the wheel more tangentially relative to the wheel than those with tetraplegia, resulting in a relatively higher percentage of effective impact force.

The hand movement patterns were not related to the effectiveness of the impact force. The effectiveness of the impact force for each hand movement pattern was dependent on the level of SCI of the participants. The participants with tetraplegia exhibited lower impact force effectiveness than the participants with paraplegia, regardless of the hand movement trajectory pattern used. In addition, the same hand pattern had higher

effectiveness of impact force when observed in a participant with paraplegia than when observed in participants with tetraplegia. This was found to be true for all recovery patterns used by participants in both groups. These results suggest that the distribution of the impact force between the radial and tangential components may be dependent on the functional capacity of the individual, especially the capability to firmly grasp the handrim. The participants with tetraplegia require more radial components to overcome grip limitations, whereas the participants with paraplegia can direct the impact force more tangentially to maximize propulsion performance.

The participants in this study exhibited all four hand movement patterns which have been previously characterized in the literature. We observed that the primary choices of hand movement pattern by participants with tetraplegia were the ARC and the SC patterns, whereas those with paraplegia exhibited the SLOP and DLOP patterns. The statistical comparisons made between hand movement patterns by the participants in this study are based on a sample size and reflect the uneven distribution of the observed patterns across participants. As such, this limits the conclusions drawn from the participants in this study to the broader population and cautiousness is indicated with regard to the applications of the clinical practice guidelines proposed, especially among the individuals with tetraplegia.

The hand movement patterns preferred by an individual may depend on a number of factors including those associated with their position in the chair during propulsion as well as an individual's strength and motor control. Likewise, the use of a stationary wheelchair ergometer eliminates some of the dynamics and wheelchair control and hence may require less trunk control. The SLOP and DLOP patterns require raising the hand above the push rim for a part of or the whole recovery period of the propulsion cycle, whereas the ARC and SC patterns showed hand movement that follow the push rim or drop below the push rim during the recovery phase of the propulsion cycle. In contrast to the SLOP and DLOP, the ARC and SC pattern favored by the participants with tetraplegia do not require lifting the arm above the push rim and hence may require less involvement of the shoulder abductor. The ARC pattern also keeps the hands close to the rim and avoids multiple changes in direction, a potential advantage identified by Richter *et al.*¹³ The DLOP pattern incorporates high braking and acceleration of the upper limb, as noted by Shimada *et al.*¹⁰ which is likely to impose a relatively high level of muscular control. Additional research with larger sample sizes of the participants with tetraplegia can explain the

lack of DLOP and SLOP hand movement patterns in manual wheelchair users with tetraplegia.

Wrist (hand) velocity prior to contact was found to be significantly correlated with the magnitude of impact force and impact force magnitude was not found to be specific to hand movement patterns. Together, these results indicate that if the impact force magnitudes have a negative effect on the wheelchair users, then the magnitude of the impact force may be reduced by modifying the control of hand velocity prior to contact. If higher impact forces are needed (e.g. in the case when more friction is needed), then increasing the hand velocity is a mechanism for increasing the impact force magnitude. Since the relationship between wrist velocity and impact force is not specific to hand movement patterns, the regulation of wrist velocity can be addressed without requiring extreme modification to preferences in a particular hand movement pattern. If the goal is to improve the effectiveness of impact force, we found that the ARC pattern for participants with paraplegia, although the least preferred pattern in our study had the highest effectiveness. For the participants with tetraplegia, all recovery patterns display low effectiveness of impact force and suggest that factors other than WCP technique are highly weighted by the limitations in the participants with tetraplegia (e.g. the need to overcome grip limitations). Finally, the wheelchair users with tetraplegia who are interested in increasing the effectiveness of the impact force or reducing the magnitude of the impact force may benefit from alternative means of increasing the frictional forces between hand and push rim such as high friction gloves or alternative propulsion mechanisms (e.g. power-assisted pushrim wheelchairs).

The conclusions drawn from this study are limited by the small sample size. A small sample size contributes to a reduction in statistical power to differentiate between experimental conditions. The number of wheelchair users with cervical level injuries who use manual wheelchairs, are pain free, and can maintain desired velocities is relatively small. A multicenter study that can draw from a greater pool of subjects may be a solution for increasing the subject sample size.

Conclusions

The purpose of this study was to identify the hand movement patterns used by manual wheelchair users with tetraplegia and paraplegia and to examine the role of the impact force in relation to the propulsion technique. Relating the impact forces to the WCP technique used by an individual can potentially facilitate the development of guidelines and intervention for preserving

functional mobility for the aging manual wheelchair users. The wheelchair users with tetraplegia do not exhibit some of the propulsion patterns seen in the wheelchair users with paraplegia. The magnitude of the impact force was found to be highly correlated with the wrist velocity prior to contact and was more effective for WCP for users with PP than tetraplegia. As there was no significant difference between the hand movement patterns with regard to the magnitude or percent effectiveness of the impact force, the preference for one technique over another seems to be determined by other factors such as metabolic cost, preferred mechanical loading, as well as the individual's capacity and ability to control the trunk and upper limb during both the push cycle and recovery phases of WCP. Future studies will continue to explore the pros and cons of effective WCP techniques for individual users.

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