8-17-2009

Effects of Stroke Patterns on Shoulder Joint Kinematics and Electromyography in Wheelchair Propulsion

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ACCEPTANCE

This dissertation, EFFECTS OF STROKE PATTERNS ON SHOULDER JOINT KINEMATICS AND ELECTROMYOGRAPHY IN WHEELCHAIR PROPULSION, by LI-SHAN CHANG, was prepared under the direction of the candidate's Dissertation Advisory Committee. It is accepted by the committee members in partial fulfillment of the requirements for the degree Doctor of Philosophy in the College of Education, Georgia State University.

The Dissertation Advisory Committee and the student's Department Chair, as representatives of the faculty, certify that this dissertation has met all standards of excellence and scholarship as determined by the faculty. The Dean of the College of Education concurs.

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ABSTRACT

EFFECTS OF STROKE PATTERNS ON SHOULDER JOINT KINEMATICS AND ELECTROMYOGRAPHY IN WHEELCHAIR PROPULSION

by

Li-Shan Chang

The purpose of this dissertation was to analyze shoulder joint kinematics and electromyographic activities of wheelchair propulsion between two stroke patterns. Twenty physical therapy students (14 females and 6 males, age 27.4 ± 5.9 years, body mass 64.41 ± 9.37 Kg and body height 169.32 ± 9.12 cm) participated. Eleven reflective markers were placed on thorax and right scapula, humerus, third metacarpophalangeal joint and wheelchair axle. Surface electrodes were placed on right pectoralis major, anterior and posterior deltoids, infraspinatus, middle trapezius, biceps brachialis long head and triceps brachialis. Participants propelled a standard wheelchair on a stationary roller system at 0.9 m/s and 1.8 m/s with semicircular (SC) and single loop (SL) stroke patterns for 20 seconds. Three-dimensional body movement and muscle activities were recorded at 100 and 1000 Hz, respectively. All data were compared for differences between two patterns and two speeds using 2-way repeated measures ANOVA ($\alpha < .05$).

Results showed longer drive phase and shorter recovery phase in SC when compared to SL, with no difference found on cycle time. Smaller release angles in SC caused longer angle ranges of hand contact on the pushrim while initial contact angles did not change. During drive phase, smaller scapular protraction range of motion (ROM) was found in SC. Shoulder abduction in drive phase was larger in terms of the maximal angle

and ROM. In the recovery phase, minimal scapular tilting, protraction, and shoulder abduction and internal rotation were larger in SC when compared to SL pattern. Shoulder linear velocities and accelerations were higher in both phases for abduction/adduction and flexion/extension in SC. For SC pattern, pectorals major and middle trapezius showed lower activities during drive phase while posterior deltoid and triceps showed higher activities during both phases when compared to SL.

Although posterior deltoid and triceps muscles work harder in SC pattern, longer drive phase and lower muscle activities in pectorals major and middle trapezius during the drive phase may make SC the better stroke pattern in wheelchair propulsion when compared to SL.

EFFECTS OF STROKE PATTERNS ON SHOULDER JOINT KINEMATICS AND ELECTROMYOGRAPHY IN WHEELCHAIR PROPULSION

by

Li-Shan Chang

A Dissertation

Presented in Partial Fulfillment of Requirements for the
Degree of
Doctor of Philosophy
in
Sport Science
in
the Department of Kinesiology and Health
in
the College of Education
Georgia State University

Atlanta, Georgia
2009

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CHAPTER 1

EFFECTS OF STROKE PATTERNS ON SHOULDER JOINT KINEMATICS AND ELECTROMYOGRAPHY IN WHEELCHAIR PROPULSION

INTRODUCTION

1.1 Shoulder Pain in Manual Wheelchair Users

According to the 1994-1999 National Health Interview Survey on Disability (NHIS-D) conducted by the Centers for Disease Control and Prevention (CDC), there are over 2.3 million people using wheelchairs as major assistive devices for activities of daily living and mobility (National Health Interview Survey on Disability, 1999). This indicates a transfer from leg to arm work for ambulation and all other functional activities. Advances in medicine and health care in the past decades have increased life expectancies for this population, especially for individuals with spinal cord injury who represent the typical users of wheelchairs (Cooper, Boninger, Spaeth, Ding, Guo, Knnotz, et al., 2006). Though the population of Americans grew 13% from 1980 to 1990 according to the US Census, the population of those who used wheelchairs almost doubled during that same period of time (LaPlante, Hendershot & Moss, 1992).

Among those who use manual wheelchairs as a means of locomotion, repetitive stress associated with long-term use may cause overuse injuries and degenerative soft tissue changes in shoulder, elbow or wrist joints (van der Woude, de Groot & Janssen, 2006). Compared to the lower extremities, work of upper extremities is less efficient and more straining, and leads to a lower physical capacity. Those musculoskeletal injuries, such as carpal tunnel syndrome, impingement of subacromial structures and rotator cuff tears, may significantly impair the independence of manual wheelchair users and impede the effects of rehabilitation (Rogers, Tummarakota & Lieh, 1998). These secondary injuries would also pose a major burden on the society and economy due to the increases of health care expenses.

The shoulder joint has been reported to be the most common site of musculoskeletal injuries in manual wheelchair users and counts for 31% to 73% of injury prevalence in this population (Helm & Veeger, 1996; Mulroy, Gronley, Newsam & Perry, 1996; Boninger, Cooper, Robertson & Shimada, 1997; Kulig, Rao, Mulroy, Newsam, Gronley, Bontrager & Perry, 1998; Cooper, Boninger, Shimada & Lawrence, 1999). In daily life, unrestricted shoulder function is one of the major factors influencing the degree of functional independence of wheelchair users (Boninger et al., 1997; Kulig et al., 1998). The shoulder is vital not only for mobility but also for locomotion. To alleviate shoulder pain and prevent shoulder joint degeneration, research identifying the biomechanical characteristics during wheelchair propulsion is crucial to provide a better understanding of injury-related factors.

1.2 Mechanical Loads in Shoulder Joints during Wheelchair Propulsion

The mechanism of shoulder injuries related to wheelchair propulsion can be explained by relatively high load and high frequency of this load on the shoulder during wheelchair propulsion (Bayley, Cochran & Sledge, 1987; van Drongelen, van de Woude, Jassen, Angenot, Chadwick & Veeger, 2005). In the author’s thesis study, shoulder joint reaction forces and net moments were compared in twenty manual wheelchair users during two speeds (1.3 m/s and 2.2 m/s) of wheelchair propulsion. The mean peak shoulder joint forces were found to be greatest in the anterior direction (88.98 N at 1.3 m/s and 184.69 N at 2.2m/s), followed by the superior direction (41.42 N at 1.3 m/s and 59.44 N at 2.2 m/s), and then the lateral direction (20.99 N at 1.3 m/s and 33.30 N at 2.2 m/s). The greatest net shoulder joint moments were found to be in the flexion direction (21.93 Nm and 35.79 Nm at 1.3 m/s and 2.2 m/s, respectively), followed by the abduction direction (19.31 Nm and 30.34 Nm at 1.3 m/s and 2.2 m/s, respectively).

It has been suggested that the combination of the superior and anterior forces may be the most harmful to shoulder structures because they force the humerus up and forward inside the joint (Koontz, Cooper, Boninger, Souza & Fay, 2002). Therefore, more depressive muscular forces from the sternal portion of the pectoralis major and rotator cuff muscles are required to keep the humeral head from displacing into the acromio-humeral space and impinging the supraspinatus tendon against the overlying acromio-clavicular arch (Kulig, Newsam, Mulryo, Rao, Gronley, Bontrager & Perry, 2001; Koontz et al., 2002). In addition, Hall (1995) emphasized the importance of the medial shoulder joint force because this force drives the humeral head into glenoid cavity and could help to stabilize the shoulder joint in the drive phase.

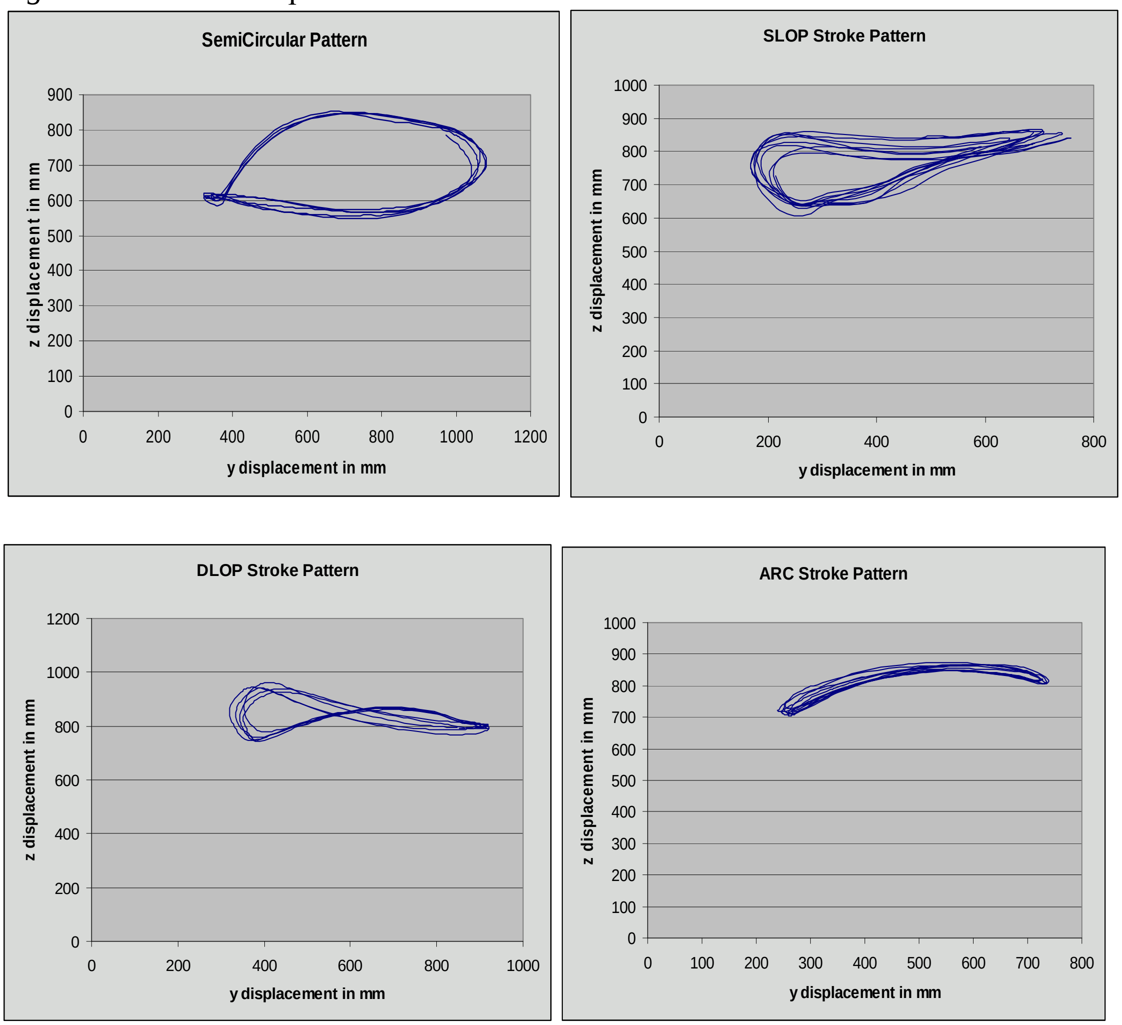
During fast or inclined propulsion, the increased superior force in the shoulder joint may contribute to compression of subacromial structures, especially when the humerus is positioned in abduction and internal rotation (Newsam, Rao, Mulroy, Gronley, Bontrager & Perry, 1999). Raising the arm above 30° has been shown to increase intramuscular pressure in the supraspinatus muscle to an extent that normal blood perfusion may be impaired (Newsam el al., 1999). By combining shoulder flexion with abduction, the supraspinatus muscle insertion passes under the coracoacromial ligament or the anterior process of the acromion, placing it in the most vulnerable position for impingement (Hall, 1995). Therefore, shoulder abduction combined with the repetitive anterior forces during wheelchair propulsion may predispose manual wheelchair users to develop rotator cuff impingement syndrome (Koontz et al., 2002).

1.3 Propulsive Technique

Robertson, Boninger, Cooper and Shimada (1996) investigated two-dimensional pushrim forces and joint kinetics during wheelchair propulsion between inexperienced and experienced wheelchair users and suggested that experienced wheelchair users tended to push longer, generated forces with lower peaks, and took a longer time to reach peak values. These characteristics described a more efficient pattern of propulsion which may be developed to reduce the chance of shoulder injuries by minimizing the loadings at shoulder joints.

Although the risk of repetitive shoulder injuries during manual wheelchair propulsion is well known (due to the mechanical load when the humeral head is positioned in a compromised position), little information is provided to manual wheelchair users about the appropriate way to push their wheelchairs. Boninger, Souza, Cooper, Fitzgerald, Koontz and Fay (2002) classified stroke patterns during manual wheelchair propulsion as: (1) semicircular pattern, recognized by the hands falling below the pushrim during the recovery phase; (2) SLOP (single loop over propulsion) pattern, identified by the hands rising above the pushrim during the recovery phase; (3) DLOP (double loop over propulsion) pattern, identified by the hands rising above the pushrim, then crossing over and dropping under the pushrim during the recovery phase; and (4) arcing pattern, recognized when the third metacarpophalangeal (MP) joint follows an arc along the path of the pushrim during the recovery phase of the stroke (Figure 1).

Figure 1. Four stroke patterns.



Among 38 manual wheelchair users, Boninger and colleagues (2002) reported that during the normal daily propulsion (0.9 m/s), the SLOP pattern was found to be the most used pattern by manual wheelchair users (40%), followed by the DLOP pattern (26%), then the semicircular pattern (21%) and the arcing pattern (13%). This finding was similar with the preference of pattern use found in the author's thesis study, which showed that nine of 20 participants (45%) used the DLOP pattern at 1.3 m/s. In addition, Boninger et al. (2002) stated that at the fast propulsion speed of 1.8 m/s, the SLOP pattern was also found to be the most popular pattern (53%), followed by the DLOP pattern (21%), then the arcing pattern (16%) and the semicircular pattern (10%). In the author's thesis study, five of 20 participants (25%) changed their stroke patterns when the speed increased from 1.3 m/s to 2.2 m/s, three from the SLOP pattern to the DLOP pattern, one from the semicircular pattern to the SLOP pattern, and one from the arcing pattern to the SLOP pattern. Shimada, Robertson, Boninger and Cooper (1998) also reported changes in propulsion patterns when pushing speed increased.

Boninger et al. (2002) suggested that the semicircular pattern is the most advantageous propulsion strategy because this pattern follows an elliptical path. An elliptical path avoids abrupt changes in hand direction and minimizes the need for extra hand movement. This pattern is similar to that employed by wheelchair racers, who follow a semicircular pattern to keep their hands on the pushrim and minimize abrupt changes in direction. Boninger et al. (2002) concluded that use of this stroke pattern might reduce the risk of trauma to the upper extremities.

The considerations in the prevention of shoulder injuries related to the wheelchair stroke patterns have fallen into (1) the pushing path, (2) the cadence of propulsion, and (3) the ratio of push time to recovery time. These spatial and temporal characteristics in different stroke patterns have been discussed in the literature (Robertson et al., 1996; Shimada et al., 1998; Boninger et al., 2002). The semicircular pattern with the hand below the pushrim during the recovery phase of propulsion is associated with an elliptical path, a lower cadence and more time spent in the drive phase relative to the recovery phase (Boninger et al., 2002). The elliptical path without extra hand movements is related to less effort to redirect hand movement and may result in lower muscle activities. The low cadence decreases the repetition of the shoulder movement during wheelchair propulsion. More relative time spent in the drive phase results in fewer loads per unit of time during the drive phase at the same propulsion velocity. Therefore, it may be wise to suggest manual wheelchair users to employ the semicircular pattern of propulsion (Boninger et al., 2002).

Although spatial and temporal characteristics in different stroke patterns have been discussed in the literature, and pushrim forces and mechanical efficiency have been compared in different patterns (Veeger, van der Woude & Rozendal, 1989; Veeger, Woude & Rozendal, 1991; Robertson et al., 1996; Boninger et al., 2002), there is a need to investigate the shoulder movements and muscle activities in different stroke patterns in order to gain a broader and deeper view related to the mechanism of shoulder injuries during manual wheelchair propulsion. With a better understanding of the electromyographic patterns along with scapular and glenohumeral kinematics in different wheelchair stroke patterns, therapists and clinicians may instruct manual wheelchair users with the optimal stroke pattern not only in the prevention of shoulder injuries but also in the improvement of performance during wheelchair propulsion training.

1.4 Purpose of the Study

The purpose of this study was to investigate three-dimensional shoulder kinematics and determine electromyographic patterns of the shoulder muscles in two different stroke patterns during two speeds of manual wheelchair propulsion. It was also to examine the effects of stroke patterns on shoulder kinematic and electromyographic parameters. The findings of this study would enhance the understandings of the shoulder kinematic and electromyographic patterns of two aforementioned stroke patterns, as well as the advantages and disadvantages of each stroke pattern during manual wheelchair propulsion. These understandings could be used for wheelchair propulsion training.

1.5 Hypothesis

H1: the semicircular stroke pattern produces lower shoulder muscle activity (lower electromyographic amplitude) compared to the other stroke pattern under the same condition of manual wheelchair propulsion.

Rationale: Research has demonstrated that the semicircular stroke pattern is related to the elliptical pushing path and the higher ratio of push time to recovery time (Robertson et al., 1996; Shimada et al., 1998; Boninger et al., 2002). The elliptical path without extra hand movements requires less effort (lower electromyographic amplitude) to redirect the hand movement during the recovery phase and may result in lower muscle activities. Less muscle activities required for segment deceleration during follow-through may be related to balanced muscle activities and less co-contraction in the recovery phase. Longer time spent in the drive phase may result in less muscle activity during the propulsion phase in order to keep the same propulsion speed.

CHAPTER 2

REVIEW of LITERATURE

Most manual wheelchairs users, especially individuals with spinal cord injury, perform activities of daily living without the functional use of their lower extremities. Tasks such as transfer and locomotion are shifted from lower extremities to upper extremities. This functional shift results in shoulder joint degeneration due to the cumulative effects of repetitious joint loading forces. Impaired joint function can significantly affect the quality of life of individuals with spinal cord injury and diminish their independence (Bayley, Cochran & Sledge, 1987; Lal, 1998). Lal (1998) assessed the shoulders of 53 subjects who had spinal cord injury for more than 15 years and found that 72% of them had degenerative changes identified and confirmed by radiology. It was also reported that older subjects with greater wheelchair dependence and females were more likely to develop degenerative changes in the shoulders (Lal, 1998).

2.1 Epidemiology of Shoulder Pain in Manual Wheelchair Users

Manual wheelchair propulsion has been reported as one of the major causes contributing to musculoskeletal injuries and degenerative changes of shoulder joints (Bayley, Cochran & Sledge, 1987; Silfverskiold & Waters, 1991; Sabick, Zhao & An, 2001). Shoulder pain is experienced by 31% to 73% of individuals who use manual wheelchairs according to several studies (Bayley, Cochran & Sledge, 1987; Sie et al., 1992; Dalyan, Cardenas & Gerard, 1999; Cooper et al., 2006). Shoulder pain in individuals with spinal cord injury is thought to be a consequence of overuse of the upper limbs. The structures of the upper extremities are designed primarily for prehensile activities. Because the upper extremities of the individuals with spinal cord injury are also needed for daily functions such as mobility, they are used more frequently and strenuously and subject to increased stress compared to those of able-bodied individuals. Based on the studies of Veeger, Rozendal and van der Helm (2002) and van Drongelen et al. (2005 & 2006), it can be suggested that one hour manual wheelchair activities would lead to some 1800 bi-manual pushes. Each push generates a reaction or compression force in the shoulder joint of about 40 kg. As a result, it is suggested that shoulder pain is mostly associated with wheelchair mobility, ambulation and transfer, since those activities require loading and repetitive movements on the shoulders (Dalyan, Cardenas & Gerard, 1999).

2.1.1 Age and Wheelchair Use

Advances in the health care of the individuals with spinal cord injury have resulted in an increased life expectancy in this population. As an individual with spinal cord injury ages, an increase in the prevalence of shoulder pain might be expected. Nichols and colleagues determined that 51.4% of 517 respondents with spinal cord injury reported either current or prior shoulder pain (Nichols, Norman & Ennis, 1979). While overall shoulder pain prevalence in their sample was 31%, Gellman, Sie and Waters (1988) found that prevalence increased with duration of wheelchair use. Similarly, Sie, Waters, Adkins, and Gellman (1992) documented that the overall prevalence of shoulder pain in their 103 participants with paraplegia was 36%, with the prevalence increasing over time (i.e., approaching 72% in those 20 years or more after injury). To distinguish the effects of age and duration of spinal cord injury on shoulder pain, Pentland and Twomey (1994) compared 52 men with paraplegia to 52 able-bodied men, matched on age and activity level. They found shoulder pain to be associated with paraplegia and duration of injury, exclusive of chronological age (Pentland and Twomey (1994).

Although shoulder pain prevalence increases with age and years of wheelchair use, it is suggested that the age of the individual with spinal cord injury has the greater influence, particularly among extremely young or older subjects (Nyland, Quigley, Huang, Lloyd, Harrow & Nelson, 2000).

2.1.2 Gender

Gender may also be a risk factor for shoulder pain in manual wheelchair users. Comparing 11 women with paraplegia to their counterparts matched on age, gender, and activity level, Pentland and Twomey (1991) found that 73% of the women with paraplegia had shoulder pain, compared to only 27% in the control group. Since the observed prevalence (73%) was higher than the prevalence in the male samples of earlier studies (Nichols, Norman & Ennis, 1979; Bayley, Cochran & Sledge, 1987; Gellman, Sie and Waters, 1988), the investigators suggested that gender might have effects on the development of shoulder pain in wheelchair users.

2.2 Mechanism of Shoulder Pain in Manual Wheelchair Users

It has been reported that possible causes of shoulder injuries include the repetitive nature of wheelchair propulsion (Burnham et al., 1993; Rodgers et al., 1994), the high-strength requirements placed by wheelchair propulsion on the shoulder muscles (Bayley, Cochran & Sledge, 1987), loading of the joints at extremes of motion (Bayley, Cochran & Sledge, 1987, Gellman & Sie, 1988; Veeger, Meershoek, van der Woude &

Langenhoff, 1998), and muscular weakness or imbalance (Burnham et al., 1993; Miyahara, Sleivert & Gerrard, 1997). Others have concluded that, in conjunction with high internal joint forces, the abnormal stresses applied to the subacromial area during wheelchair propulsion and transfers contribute to the high rate of shoulder problems in the paraplegic patients (Bayley, Cochran & Sledge, 1987, Pentland & Twomey, 1991).

2.2.1 High Intra-articular Pressure Related to Degenerative Injuries

It is believed that the repetitive high intra-articular pressure in the shoulder joint during wheelchair transfer and propulsion contributes to the high rate of shoulder pain in the paraplegic population (Bayley et al., 1987; Veeger & Rozendal, 1992). Particularly, the effects of repetition can be magnified when combined with awkward posture or loading of the shoulder girdle such as occurs in wheelchair propulsion (Andersen, Kaergaard, Frost, Thomsen, Bonde, Fallentin, Borg & Mikkelsen, 2002; Frost, Bonde, Mikkelsen, Andersen, Fallentin, Kaergaard & Thomsen, 2002). During wheelchair propulsion, the requirement of active stability in the glenohumeral joint relies on more muscle activities resulting in higher joint compression forces (van Drongelen, van der Woude, Janssen, Angenot, Chadwick & Veeger, 2005). The compression force on the joint surface may cause damage to the joint surfaces, while the muscle forces can be high in order to stabilize the joint and therefore may lead to soft-tissue damage. Although van Drongelen and colleagues (2005) found the glenohumeral contact forces to be low during lever wheelchair propulsion, it was suggested that when the external load is increased by external resistance, increased velocity, or a slope, the contact forces will increase as well.

2.2.2 Muscle Imbalance

Several studies have identified a number of influences, which are known to predispose the shoulder joint to pathology (Perry, Gronley, Newsam, Reyes & Mulroy, 1996; Cooper et al., 1999). These factors include muscle strength imbalance and changes of pushing patterns in wheelchair propulsion (Miyahara, Sleivert & Gerrard, 1997). Muscular imbalance occurs when the opposing muscles are unevenly developed. As a result of this imbalance, the integrity of the joint is compromised and the risk of injury increases. Such injuries often involve stretch weakness, which is cause by the prolonged elongation of the inadequately developed antagonist muscles. To investigate the role of shoulder muscle strength imbalance as a factor for the development of rotator cuff impingement, the isokinetic examinations for shoulder abduction/adduction and internal/external rotation with peak torque values were measured and the strength ratio of abduction:adduction and internal rotation:external rotation were determined in assessing the shoulder function of individuals with paraplegia (Burnham, 1993). The results demonstrated that subjects who had subacromial impingement also had rotator cuff and glenohumeral joint adductor muscle weakness. Paraplegics' shoulders with rotator cuff impingement had higher abduction:adduction and abduction:internal rotation strength ratios than the shoulders of paraplegics without impingement syndrome. It is suggested that prolonged wheelchair use creates imbalance in propulsion agonists and antagonists. Training of the antagonists may correct these imbalances and reduce the risk for associated shoulder pain (Rogers, Keyser, Rasch, Gorman & Russell, 2001).

2.2.3 Improper Scapular Positioning

Scapular positioning is controlled by muscle attachments and thoracic spine posture. When sitting, lumbar spine posture changes from its normal lordosis to a more straight or flat alignment, increasing thoracic spine kyphosis, prompting downward rotation and protraction of the scapula (Nyland et al., 2000). Prolonged sitting and wheelchair propulsion further promotes this scapular orientation. Downward rotation and protracted scapular position contributes to abnormal forces impinging the subacromial tissues such as glenohumeral joint capsule, subacromial bursa and rotator cuff tendons. Therefore, individuals with compromised neuromuscular function are at particular risk for developing subacromial impingement when performing overhead tasks in a sitting position (Nyland et al., 2000). The direct relationship between lumbar and thoracic spine postures and their influence on scapular orientation suggests that maintaining a neutral lumbar spine posture may help alleviate shoulder pain related to impingement and overhead reaching tasks.

Dalyan, Cardenas and Gerard (1999) surveyed 130 individuals with spinal cord injury one year after the onset of injury and reported that 28 percent of individuals with shoulder pain had limited functional abilities and 26 percent needed additional help to perform functional activities due to shoulder pain. They suggested that the severity of shoulder pain contributed significantly to the loss of independence. Indeed, there is a need for the implementation for shoulder pain prevention and management programs for manual wheelchair users, especially for individuals with spinal cord injury both in the early phase of the rehabilitation program and during ongoing care even for decades. These programs should include patient education about the basic biomechanical principles on avoiding impingement and overuse, optimizing the proper pattern and technique, and managing the early signs of strains and overuse. In addition, education and training in balanced strengthening of muscles acting around the shoulder and optimizing posture to achieve a normal alignment of shoulder, head, and the spine are critical for the avoidance of shoulder injuries (Dalyan, Cardenas & Gerard (1999).

2.3 Shoulder Joint Kinematic Studies in Wheelchair Propulsion

To relate biomechanical parameters to shoulder injuries, a thorough notion of the joint kinematics during wheelchair propulsion under controlled external conditions is needed. The insight into joint kinematics can only be achieved and made meaningful when forces, moments and motion are placed into a local coordinate system (Boninger et al, 1997). By using a local coordinate system it is possible to describe joint biomechanics in anatomical terms. However, some studies published to date that address wheelchair propulsion biomechanics refer to a global coordinate system (Veeger, van der Woude & Rozendal, 1989; Veeger & Rozendal, 1992; Newsam et al., 1999).

Veeger & Rozendal (1992) described the global shoulder kinematics of five able-bodied participants during the drive phase as starting with a flexion of the shoulder from an extended position combined with abduction during the first part of the push, which changes into flexion and adduction during the last part of the drive phase. These findings were confirmed by Boninger, Cooper & Shimada (1998) and Newsam et al. (1999). When overcoming a high resistance such as negotiating a steep slope or starting from standstill, trunk flexion at initial hand contact could induce an internal rotation position at the shoulder joint accompanied by eccentric work of the external rotators. A repetitive condition of an internal rotation position, combined with high external rotator and abductor load, would be a serious risk factor for supraspinatus tendon impingement (Boninger, Cooper & Shimada, 1998).

Shimada, Robertson, Boninger and Cooper (1998) investigated shoulder motion during two speeds (1.3 m/s and 2.2 m/s) of wheelchair propulsion. This study emphasized the differences of angular position, range of motion, and peak accelerations of the shoulder between two selected propulsion speeds. These results demonstrated that in the drive phase, the minimum shoulder abduction angle and range of motion in shoulder flexion/extension and abduction/adduction were significantly different between the speeds of 1.3 m/s and 2.2 m/s. It was suggested that those changes in pushing patterns with lower biomechanical efficiency revealed possible causes of shoulder injuries (Shimada et al., 1998).

Wang, Deutsch, Hedrich, Martin and Millikan (1995) concluded that the range of motion of the upper extremities increased when propulsion speed increased. The results of their study also showed an increase in trunk flexion at the faster speed, resulting in a larger contact angle and more elbow flexion during initial contact. The time during drive phase and recovery phase both decreased when the speed increased. Also, the percentage of drive phase time by cycle time decreased with increasing speed.

In a study to investigate the effects of handrim velocity on mechanical efficiency in wheelchair propulsion (Veeger, Woude & Rozendal, 1991), the findings revealed that when handrim velocity increased, the percentage of recovery phase time in a cycle increased, and the cycle time and drive time decreased, while recovery time remained constant. However, the increased stroke arc with increasing propulsion speed reported by Veeger, Woude and Rozendal (1991) was different from the result found in the study conducted by Wang which showed that the range of stroke arc was not a monotonic increase (Wang et al., 1995). This difference in findings may result from different analytical techniques, different wheelchair types, and different stroke patterns used by the participants.

Three-dimensional shoulder kinematics in wheelchair propulsion have been reported by Rao, Bontrager, Gronley, Newsam & Perry (1996), Newsam et al. (1999), and Finley, Rasch, Keyser & Rodgers (2004). It was found that all three directions of movements of the humerus attained maximal and minimal positions during the recovery phase. Mean humeral elevation and internal rotation reached relative minimum at 22.5º and 11.6º, respectively, while the peak humeral abduction was 23.2º during the early part of the recovery phase at 40-42% of the cycle. During the later part of recovery at 93-95% of the cycle, humeral elevation and internal rotation reached relative maximum at 56.6º and 86.2º while the plane angle decreased to a minimum value (-57.3º). The mean ranges of motion for the humeral plane, rotation and elevation were 81.6 º, 76.1 º, and 35.4º, respectively. The relatively small range of motion for trunk lean (5.5º) coupled with a small inter-cycle variability (1.3º) suggested that the trunk excursion during manual wheelchair propulsion was minimal and consistent for individual subjects. However, an inter-subject variability value of 10.0º, nearly twice the average range of motion, indicated that there were trunk postural offsets between individuals (Rao et al., 1996). Boninger, Cooper and Shimada (1998) reported shoulder motion at low speed (1.3 m/s) to be within the range of 64° of flexion to 11° of extension in the sagittal plane, 21 to 47° for abduction and 54 to 91° for internal rotation. Shoulder kinematics was speed-dependent since increasing speed resulted in a marked increase in angular acceleration for flexion/extension and abduction/adduction movements (Boninger, Cooper & Shimada, 1998).

Rodgers and colleagues (2001) compared the trunk and upper extremity kinematics during wheelchair propulsion in 19 manual wheelchair users before and after 6 weeks of therapeutic training. Stretching exercises were performed for the anterior and internal rotation shoulder muscles (anterior deltoid, subscapularis, pectorals, latissimus dorsi, and teres major), triceps, and wrist flexors. Strengthening activities were concentrated on the following muscle groups: the posterior deltoid, infraspinatus, teres minor, rhomboids, middle trapezius, erector spinae, biceps, and wrist extensors. Trunk angles, joint angles (shoulder, elbow and wrist flexion/extension, shoulder abduction/adduction, and wrist radial/ulnar deviations), velocities and accelerations were calculated. Three kinematic measures were found to have significantly increased with training. These included shoulder flexion/extension range of motion, maximum elbow extension, and trunk flexion (Rodgers et al., 2001).

Muscular imbalance and inflexibility at shoulder joints in manual wheelchair users have been addressed in the literature (Burnham, 1993). Olenik advocated exercises for strengthening and stretching upper-limb muscles to address these imbalances, since they may lead to injury (Olenik, Laskin, Bumham, Wheeler & Steadward, 1995). Muscular imbalance occurs when the opposing muscles are unevenly developed. As a result of this imbalance, the integrity of the joint is compromised and the likelihood of injury increases. Training may produce a kinematic increase in motion at the trunk, shoulder, and elbow (Rodgers et al., 2001). This movement pattern allows the manual wheelchair users to rely on trunk and shoulder excursion to generate translational forces necessary for wheelchair propulsion. The trunk movement pattern may have been compensatory for peripheral muscle fatigue, since it is more pronounced in the fatigued state (Wang et al., 1995).

Whether this adaptation is beneficial is unclear. It is suggested that further investigation of propulsion patho-mechanics may identify additional factors contributing to injury (Rodgers et al., 2001).

Mechanisms of impingement have been linked to the reductions in the subacromial space (the space available for the clearance of the rotator cuff structures and long head of the biceps tendon beneath the coracoacromial arch). Factors contributing to impingement include anatomic abnormalities such as changes in acromial shape and slope (Zuckerman, Kummer, Cuomo, Simon, Posenblum & Katz, 1992), poor rotator cuff muscle function and/or muscle fatigue (Chen, Simonian, Wickiewicz, Otis & Warren, 1999), or abnormal scapular kinematics (Lukasiewicz, McClure, Michener, Pratt & Sennett, 1999; Ludewig & Cook, 2000). These factors may lead to a reduction in the subacromial space and insufficient clearance for the rotator cuff tendons beneath the coracoacromial arch (Flatow, Soslowsky & Ticker, 1994). Zuckerman et al. (1992) showed that a 5° change in the slope of the acromion and an average 20% reduction in the available subacromial space were significantly related to the incidence of rotator cuff tears.

In addition, altered kinematics of the shoulder complex are believed to exacerbate the pain and pathology associated with impingement syndrome (Lukasiewicz et al., 1999; Ludewig & Cook, 2000). Recent attention has been directed to three-dimensional analysis of scapulothoracic kinematics and the relationship to impingement problems in manual wheelchair users who are exposed to repetitive upper-extremity activities (Nawoczenski, Clobes, Gore, Neu, Olsen, Borstad & Ludewig, 2003). The most consistent evidence of abnormal scapulothoracic kinematics has been shown by a reduction in posterior tipping of the scapula during elevation of the arm. Lukasiewicz et al. (1999) compared subjects clinically diagnosed with shoulder impingement to an asymptomatic control group and found significant reductions in posterior tipping of the scapula (averaging 8°- 9°) during statically held positions with the arm elevated. Ludewig and Cook (2000) measured dynamic three-dimensional glenohumeral and scapulothoracic kinematics in subjects with impingement symptoms and compared their findings with asymptomatic controls. Significant reductions in posterior tipping (6°) were also identified in this investigation, as well as significantly decreased scapular upward rotation (4°) when the humerus was elevated to specific angles in the scapular plane. Ludewig and Cook (2000) noted that these kinematic deviations occurred in directions that may lead to a functional or transient reduction in the available subacromial space (Nawoczenski et al., 2003).

Certain patterns of scapular and humeral motion may functionally decrease the subacromial space and place the tendons of the rotator cuff at greater risk for injury or progression of impingement. It has been suggested that in non-weight-bearing upper-extremity elevation activities, these patterns include a reduction in posterior scapular tipping (or increased anterior tipping), an increase in internal rotation of the scapula, and a decrease in upward rotation (Lukasiewicz et al., 1999; Ludewig & Cook, 2000; Nawoczenski et al., 2003). A 20° decrease in humeral external rotation from the neutral position has also been suggested to contribute to greater deformation of the rotator cuff tendons beneath the anterior acromion (Flatow, Soslowsky & Ticker, 1994). It was found that during both the weight-relief raise and transfer activities, scapular and humeral positions and motion patterns are in these detrimental directions and of sufficient magnitude to pose increased risk for injury or progression of shoulder pain (Nawoczenski et al., 2003). Particularly, manual wheelchair users with impingement tend to perform transfers with reduced thoracic flexion, increased scapular internal rotation, and increased humeral internal rotation when compared to those without impingement (Finley, McQuade & Rodgers, 2005).

Several studies have reported that with increased thoracic kyphosis, the pattern of scapular superior translation and internal rotation increased while posterior tipping and upward rotation decreased, during humeral elevation (Finley and Lee, 2003; Yang, Koontz, Triolo, Mercer & Boninger, 2006). The findings from Finley, McQuade & Rodgers (2005) indicated that manual wheelchair users with impingement do perform transfers with increased scapular internal rotation; however, they also have reduced thoracic flexion (decreased kyphosis) and increased humeral internal rotation.

Most wheelchair propulsion related studies apply kinematic analyses in order to obtain spatial or temporal characteristics, such as frequencies, cadence, drive phase time, recovery phase time, cycle time, push angle, recovery angle, range of cycle and phase ratio, or to investigate joint mechanics, such as joint reaction forces and net moments with inverse dynamics. Few researchers have looked into the shoulder kinematics in terms of joint angles and ranges of motion in different phases. To date, no one has assessed scapular kinematics during manual wheelchair propulsion with comparisons between different stroke techniques/patterns. Further studies investigating three-dimensional glenohumeral and scapulothoracic kinematics associated with the mechanisms of shoulder pain are suggested.

2.4 Analysis of Muscle Activity in Manual Wheelchair Propulsion

The intensity and duration of electromyographic (EMG) activities of shoulder muscles during wheelchair propulsion have been reported in several studies, in combination with three-dimensional kinematic and kinetic parameters to describe the time-dependent role of different muscles and to determine their role in power production (Veeger & Rozendal, 1992; Veeger et al, 1998; Finley et al, 2004; van Drongelen et al, 2005).

Trunk instability due to the absence or impairment of abdominal and back muscle control or the long period of sitting usually leads to an increased kyphotic posture with flattened lumbar spine among individuals with spinal cord injury (Hobson & Tooms, 1992, Yang et al., 2006). This functional sitting posture allows individuals with spinal cord injury to shift the trunk center of gravity back and secure it within their base of support without losing balance in a wheelchair. However, this passive kyphotic sitting posture is associated with back pain and rotator cuff injury (Curtis, Drysdale, Lanza, Kolber, Vitolo & West, 1999; Sinnott, Milburn & McNaughton, 2000; Samuelsson, Tropp & Gerdle, 2004). In addition, lack of trunk stability, which results in less erect posture and poor support of the shoulder girdle complex, may limit production of maximal shoulder strength (Powers, Newsam, Gronley, Fontaine & Perry, 1994). Newsam et al. (1999) assessed upper extremity motion during wheelchair propulsion among persons with different levels of spinal cord injury (C6 tetraplegia, C7 tetraplegia, high paraplegia, and low paraplegia). They reported that participants with high cervical lesions yielded greater range of trunk motion during propulsion. They suggested that stabilizing the trunk might help participants who lose voluntary control of trunk musculature to maintain consistent propulsive stroke patterns.

Poor trunk control also limits the ability of manual wheelchair users to overcome fatigue during wheelchair propulsion. Rodgers et al. (1994) investigated the influence of fatigue on trunk movement during wheelchair propulsion. They reported a significant increase of trunk forward lean with fatigue. This increase in forward lean may aid the application of force to the pushrim and enable the transfer of propulsive power from the trunk and upper extremity to the pushrim (Sanderson & Sommer, 1985). Rodgers, Keyser, Gardner, Russell and Gorman (2000) also found that subjects with increased trunk flexion during propulsion, which was accentuated with fatigue, had greater shoulder flexion and elbow extension when compared to subjects with a more erect posture. The trunk flexion pattern appeared to be a compensatory strategy to generate a propulsion moment during muscle fatigue.

Mulroy et al. (1996) identified two synergies of shoulder muscle function during wheelchair propulsion. The push phase synergy was dominated by muscles with shoulder flexion (anterior deltoid, pectoralis major), external rotation (supraspinatus, infraspinatus) and scapular protraction (serratus anterior) functions. Pectoralis major and supraspinatus had the highest peak (58% and 67% MAX) and average (35% and 27% MAX) EMG intensities in this group. The dominant functions of the recovery synergy were extension (posterior deltoid), abduction (medial deltoid, supraspinatus), internal rotation (subscapularis) and scapular retraction (middle trapezius). This group of muscles had EMG onsets in late push (17% to 26% cycle) with moderate average intensities (21% to 32% MAX). The push phase muscles (anterior deltoid, pectoralis major) were also activated in the late recovery phase to decelerate the back swing of the arm and to prepare the hand (increasing hand speed) for the contact on the pushrim (Mulroy et al., 1996).

The same phenomenon was described for the recovery phase muscles, activated already at the end of the push phase to restrain shoulder flexion (posterior deltoid), adduction (medial deltoid) and external rotation (subscapularis). During both the push and recovery phases, significant activity was found in one or more rotator cuff muscles. Most of the paraplegics in the study conducted by Mulroy et al. (1996) demonstrated a high peak of supraspinatus activity during the push phase, while the duration of the recovery activity was lengthy (59% of the cycle). Four of the 17 participants used this muscle during both phases.

It was revealed by Mulroy et al. (1996) that with reduced activity of the rotator cuff muscles due to fatigue, contraction of the deltoid would result in upward gliding of the humeral head and possible impingement of the supraspinatus tendon against the subacromial arch. During the drive phase, fatigue of pectoralis major would increase the risk of impingement due to the intra-articular force within the glenohumeral joint. During the recovery phase, the greater tubercle moved directly underneath the acromion due to increased shoulder internal rotation and this position would worsen the impingement syndrome (Mulroy, Farrokhi, Newsam & Perry, 2004).

At the elbow joint, biceps brachialis was activated in the late recovery phase and continued its action over a period where an elbow flexion torque would contribute to the propulsion. Likewise, triceps brachialis only became active when elbow extension would contribute to a propulsive force on the hand rim (Rodgers et al., 1994; Mulroy et al., 1996; Schantz, Björkman, Sandberg & Andersson, 1999; Guo, 2003). It is believed that biceps brachialis and triceps brachialis are necessary for an effective force direction (Guo, 2003). After the hand has made contact with the pushrim, the pull phase starts with an initial elbow flexion, accompanied by activity of the biceps brachialis (Veeger, van der Woude & Rozendal, 1989). Anterior deltoid has a high activity at the beginning of hand contact, whereas pectoralis major has a more constant activity of longer duration. These two muscles are considered to be the prime movers in wheelchair propulsion (Veeger, van der Woude & Rozendal, 1989; Rodgers et al., 1994; Mulroy et al., 1996; Guo, Zhao, Su & An, 2003). Vanlandewijck, Spaepen & Lysens (1994) demonstrated that latissimus dorsi shows a similar activity pattern, stressing its importance during the push phase. Although there is an increasing abduction of the arm, no activity of medial deltoid is recorded, suggesting that this shoulder abduction is not an active movement (Veeger, van der Woude & Rozendal, 1989). It is assumed to be caused by the high activity of the prime movers at the start of the push phase, causing an anterior-flexing and endo-rotating torque in a closed chain (Veeger & Rozendal, 1992).

It was concluded by Schantz et al. (1999) that the brachial biceps and triceps, anterior deltoid and pectoralis major muscles could be anticipated to propel the wheelchair forward, whereas the posterior deltoid and trapezius muscles could be expected to play a role, especially during the recovery phase. However, individual differences exist. The posterior deltoid and trapezius muscles showed an unexpected and distinct activity during the push phase, possibly resulting from the function of stabilizing the shoulders. Regardless of the type of recovery movement (semicircular, loop or pumping), the posterior deltoid and trapezius muscles in particular were active during part of, or the whole, recovery phase (Schantz et al., 1999). The general order of activation of, first, the brachial biceps, thereafter the pectoralis major and anterior deltoid, and then the triceps brachial muscle at the normal velocity during the push phase is constant with findings in other studies (Veeger, van der Woude & Rozendal, 1989; Mulroy, 1996).

2.5 Stroke Patterns in Wheelchair Propulsion

It has been noted that experienced wheelchair users tend to adapt their stroking techniques representing mechanics that are less likely to cause injuries than inexperienced wheelchair users (Robertson, Boninger, Cooper & Shimada, 1996; Boninger, Cooper, Robertson & Rudy, 1997). It follows that by adjusting the wheelchair setup to the individual and by training the individual in appropriate stroking technique, shoulder injuries may be prevented and relieved (Schantz et al., 1999).

To study wheelchair propulsion technique at different speeds, five well-trained subjects propelled a wheelchair on a treadmill at four belt speeds of 0.56-1.39 m/s and against slopes of 2 and 3 degrees (Veeger, van de Woude & Rozendal, 1989). The results showed considerable inter-individual differences in stroke pattern, and general changes relative to speed occurred in the group. Although Veeger, van der Woude and Rozendal (1989) reported that the semicircular pattern was significantly more efficient, which is consistent with the results found by Shimada et al. (1998) based on seven experienced wheelchair users, Boninger et al. (2002) did not find differences in kinetic parameters, such as pushrim forces or mechanical efficiency based on different stroke patterns. However, the superiority was based on five participants, and only one of them used the semicircular technique (Veeger, van der Woude & Rozendal, 1989).

The kinematics of wheelchair propulsion was studied by Saccavini, Bizzarini, Magrin, Odelli, Malisan and Pascolo (2003) in order to identify the most effective stroke pattern with regard to speed and acceleration produced.

Fourteen spinal cord injured athletes were analyzed on a wheelchair ergometer and the results revealed that of the three stroke patterns, semicircular, single loop and double loop, the double loop pattern was the most effective for propulsion at submaximal speeds, because it exploits the inertial force of the push in the anterior dead point phase, avoiding the phenomenon of kinematic stop (Saccavini et al., 2003).

In addition to joint accelerations and efficiency, three wheelchair propulsion stroke patterns were characterized by investigating joint ranges of motion and wheelchair propulsion phases (Shimada et al, 1998). Abrupt changes in joint angles can be detected through the analysis of joint accelerations, while extreme range of motions can be identified through large joint excursions. The analysis of propulsion phases may be used to identify inadequate time for force application, while stroke efficiency can be used to detect the extraneous forces that do not contribute to the forward motion of the wheelchair. The findings revealed that subjects with the semicircular pattern had smaller flexion/extension and shoulder abduction/adduction acceleration values, when compared to the subjects with the double loop and single loop stroke patterns, respectively. The decreased acceleration seen in the individuals with a semicircular stroke pattern may lessen the risk of acceleration-related injuries (Shimada et al, 1998). The subjects with the semicircular stroke pattern had significantly larger elbow flexion/extension (during both speeds of propulsion) and shoulder abduction/adduction (during the fast speed) ranges of motion when compared to the other two stroke patterns. The mean maximum elbow flexion/extension and shoulder abduction/adduction angles for these subjects were approximately 120 and 75 degrees, respectively, during both speeds, and the minimum angles were approximately 80 and 35 degrees, respectively. Since these maximum and minimum joint angles associated with the semicircular stroke are well within normal range, the larger ranges of motion found in subjects with a semicircular stroke pattern are not likely to contribute to injury (Shimada et al, 1998).

Richter, Rodriguez, Woods and Axelson (2007) classified stroke patterns for level and uphill propulsion according to 1 of 4 common classifications: arcing, semi-semicircular, single loop (SLOP), and double loop (DLOP). Among twenty-six manual wheelchair users with spinal cord injury in their study, only 3 of the 4 stroke patterns were observed. None of the subjects used the semi-semicircular pattern. For level propulsion, the stroke patterns were fairly balanced between arcing (42%), SLOP (31%), and DLOP (27%). It was found subjects tended to change their stroke pattern for pushing uphill, with 73% of the subjects choosing the arcing pattern by the 6° grade. Chou, Su, An and Lu (1991) observed the stroke patterns used by three experienced wheelchair users and three non-wheelchair users while pushing on level ground. The experienced users were found to use the semicircular pattern while the non-users implemented the arcing pattern. It was noted by Schantz et al. (1999) that the semicircular recovery movement was used by all three subjects under normal conditions. This is understandable, as it involves a swinging phase and thereby appears to be a more relaxed form of recovery, whereas both the pumping movement and the loop movement, used at maximal velocity and maximal start, are movements distinctly aimed at rapid initiation of a new propulsive action (Schantz et al., 1999).

The stroke patterns differ from one another during the recovery phase, but not during the drive phase. During the drive phase, wheelchair users are required to follow the path of the pushrim and are part of a closed kinetic chain. Therefore, the forces generated at the pushrim do not vary by stroke patterns. This consistency of pushrim forces infers no significant differences of shoulder joint forces when comparing different stroke patterns in manual wheelchair propulsion (Veeger, Woude & Rozendal, 1991; Robertson et al., 1996; Shimada et al., 1998; Boninger et al., 2002; Richter, Rodriguez, Woods & Axelson, 2007).

However, Boninger et al. (2002) did find differences between patterns in cadence and in the ratio of push time to recovery time. The semicircular pattern showed the lowest cadence and the highest ratio of push time to recovery time (Sanderson & Sommer, 1985; Shimada et al., 1998). These differences in cadence based on propulsion pattern were consistent with Shimada's results. Wheelchair users who employed the semicircular propulsion pattern hit the pushrim less frequently to reach the same speed, and, for every complete stroke, spent a greater percentage of time pushing than recovering. The stroke pattern that minimizes frequency or cadence may also reduce the risk of injuries (Boninger et al., 2002).

Boninger et al. emphasized that the recovery phase is an important, modifiable parameter that can impact injurious biomechanics. A propulsion stroke during which the arm drops below the pushrim resulted in a greater push angle and lower frequency, both of which likely protect against upper extremity injury. It is also suggested that accelerating and decelerating the arms during the recovery phase requires an amount of energy, although this does not contribute directly to propelling the wheelchair (de Groot, Veeger, Hollander & van de Woude, 2005). If the recovery phase indeed costs energy, then a high cycle frequency leads to more acceleration and deceleration of the arms and subsequently a higher energy loss. Therefore, during the recovery phase the stroke pattern with lower frequency may save manual wheelchair users energy that must come from muscle activities (Boninger et al., 2002).

2.6 Conclusion

Although the recent focus of non-surgical rehabilitation for manual wheelchair users with shoulder impingement has been directed to the correction of faulty scapular and humeral motion patterns (Ginn, Herbert, Khouw & Lee, 1999; Bang & Deyle, 2000), scapulothoracic and glenohumeral kinematics have not been assessed during functional tasks, such as wheelchair propulsion in manual wheelchair users. The prevention of shoulder pain in wheelchair propulsion characterized by its high-repetitive, high-loading motion combined with awkward posture of the shoulder girdle can be approached by the coordinated muscle activities aiding in depression and stabilization of the humeral head, providing protection for the glenohumeral joint. The consideration of the wheelchair stroke patterns in terms of the pushing path, cadence and phase ratio also plays an important role in the prevention of shoulder injury. The semicircular pattern with the hand below the pushrim during the recovery phase of propulsion is associated with an elliptical path, a lower cadence and more time spent in the drive phase relative to the recovery phase (Boninger et al., 2002). The elliptical path without extra hand movements is related to a more efficient pushing force leading directly to wheelchair propulsion. The low cadence decreases the repetition of the shoulder movement during wheelchair propulsion. More relative time spent in the drive phase results in less loading during the drive phase. Therefore, it may be wise to train wheelchair users in the semicircular pattern of propulsion (Boninger et al., 2002).

CHAPTER 3

METHODOLOGY

3.1 Participants

Twenty graduate students from the Division of Physical Therapy (PT) at Georgia State University who have been instructed in wheelchair propulsion and are familiar with both stroke patterns were recruited as participants in this study. The instruction and training of wheelchair use was part of the first year of PT curriculum in Georgia State University and it lasted for two to three weeks. Participants may have different experience in wheelchair propulsion depending on how long they have been in the PT program and where they have completed their clinical training. To ensure the qualification of each participant to be included in this study, a competency test was performed before data collection. The competency test included collecting a trial of kinematic data for each stroke pattern at 1.8 m/s. Participants who were unable to maintain the required speed for 20 seconds, lost the consistency of their pushing rhythm (determined by the hand movement in the sagittal view) during any time of the 20-second period, or could not keep the same stroke pattern (for example, changed from the semicircular pattern to arcing or the double loop pattern, or from single loop to double loop patterns) were excluded from this study. Individuals who currently suffered from shoulder pain or musculoskeletal injuries at the time of data collection, or had overuse/chronic shoulder pain during the past six months before testing were also excluded from this study.

A questionnaire of age, gender, body height, body mass, dominant hand and years of study in Physical Therapy program, was completed by each participant and collected before the experiment procedures. All subjects participated voluntarily and provided informed consent in accordance with guidelines approved by the Institutional Review Board (IRB) at Georgia State University.

3.2 Instrumentation

In this study, a standard rigid frame Quickie II wheelchair which fit each participant was used for all propulsion trials. The wheelchair was secured and propelled on a stationary roller system. The roller system consisted of a metal roller (diameter = 17.8 cm) mounted on the rear end of a metal base. The footplate of the wheelchair was fixed to the front end of the base, and the rear wheels were supported by the roller. The axle of the rear wheels was aligned vertically with the axis of rotation of the roller. A Cat-Eye cycling speedometerwith a scale of 0.1-km/h increments was used to monitor the speeds (0.9 m/s and 1.8 m/s) of wheelchair propulsion.

3.2.1 Kinematic Instrumentation

A two-camera Video Motion Analysis System (Qualisys System, Inc., Sweden) was used to record three-dimensional thoracic, scapular and humeral kinematics in wheelchair propulsion at the selected speed at the sampling rate of 100 Hz. Two high-speed digital infrared cameras (Proreflex MCU240, Sweden) were calibrated using a three-dimensional calibration frame (1.5 x 1.0m) and a calibration wand (1.1m) with 4 and 2 reflective makers, respectively. The total calibrated area was 1.0 x 1.5 x 2.0m. The axes of the calibration were defined as x in anterior (+) - posterior (-), y in medial (+) - lateral (-), and z in vertical (up as positive and down as negative), to the orientation of the wheelchair. The standard error for the calibration was below 0.001m (0.09%). During calibration and data collection, two cameras were oriented oblique (approximately 60 degrees apart) in the sagittal view. Both cameras were located on the right side of the wheelchair.

Automatic capturing and tracking of the markers were completed by using QTrac 2.77 Software (Qualisys Inc., Sweden). The images of thoracic, scapular and humeral movements during data collections were automatically digitized. The three-dimensional data were generated from the filtered data based on the direct linear transformation (DLT) method built in the QTrac software. All digitized data were filtered using a fourth-order Butterworth technique (with a cutoff frequency of 6 Hz suggested by Cooper, DiGiovine, Boninger, Shimada, Koontz & Baldwin, 2002) built in Matlab 6.0 (The MathWorks, Inc., Natick, MA). The Euler angles with $Z-Y-X$ rotation sequence were used to transform the global reference system into segmental local coordinate systems.

3.2.2 EMG Instrumentation

The surface electromyographic measurements were conducted by using the Biopac MP100 (BIOPAC Systems, Inc., Goleta, CA) along with an electromyography 100C module. Semicircular Ag-AgCl electrodes with a 10-mm diameter were used. The skin was prepared by cleansing with alcohol and applying conductive gel. Electrodes were spaced on the right side of pectoralis major (in the middle between sternoclavicular joint and coracoid process, 2 cm below the clavicle), anterior deltoid (2 cm below the anterior rim of the acromion), infraspinatus (in the middle between the spine of the scapula and inferior angle, 2 cm of the medial border of the scapula), posterior deltoid (2 cm below the posterior rim of the acromion), middle trapezius (1 cm medial and 4 cm cranial to the midpoint of triangular surface on medial border of the scapula in line with the scapular spine), biceps brachialis long head (in the middle of the muscle belly), and triceps brachialis (2 cm below the edge of posterior deltoid muscle). A ground electrode was located at the bony part of right wrist (van der Helm & Veeger, 1996).

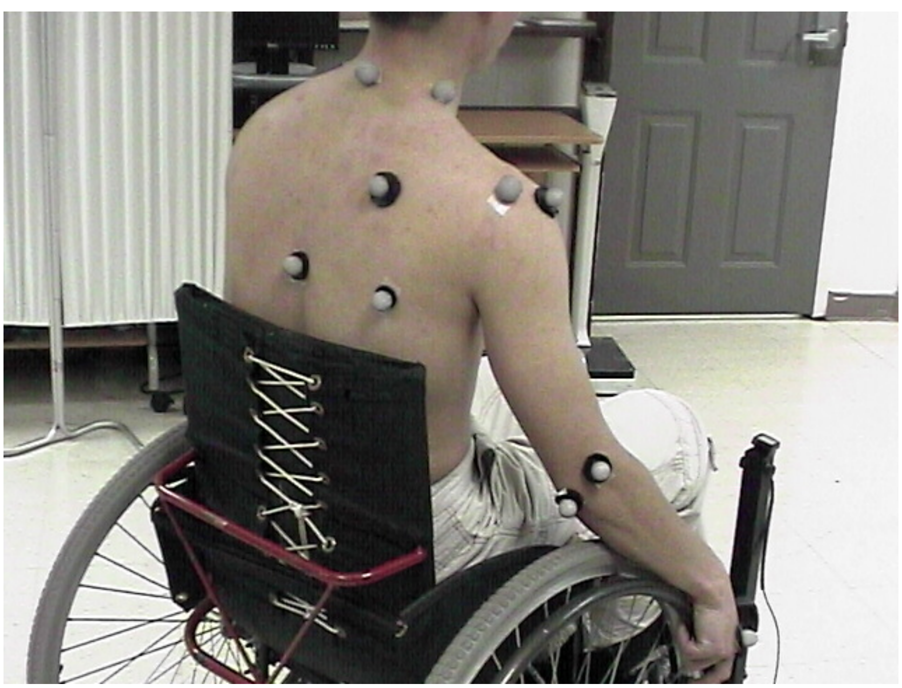
The signals were processed with a differential amplifier (bandwidth, 10-500Hz; input impedance, 1Gμ; common mode rejection ratio, 110dB at 60Hz; gain, 1000). An external trigger was used to ensure that the electromyographic data collection was synchronized with the kinematic data collection. The sampling rate was set to 1000Hz. Linear envelopes of electromyographic signals was constructed by rectifying and low-pass filtering, using a fourth-order recursive Butterworth filter with a 10-Hz cutoff frequency (Richter, Rodriguez, Woods, Karpinski & Axelson, 2006). The conditioned signals were then re-sampled at a frequency of 200Hz, corresponding to two times the sampling rate of the kinematic data collection. Electromyographic data were normalized by each subject's maximum voluntary contraction (MVC) and multiplied by 100, resulting in the percentage of MVC.

3.3 Procedures

Before beginning the experiment, each participant signed a consent form approved by the Georgia State University Institutional Research Board (IRB) for human subject testing. Each participant also completed the questionnaire about age, gender, body height, body mass, dominant hand, years studying in physical therapy program. Finally, the body mass of each participant was measured by a standard dial platform scale.

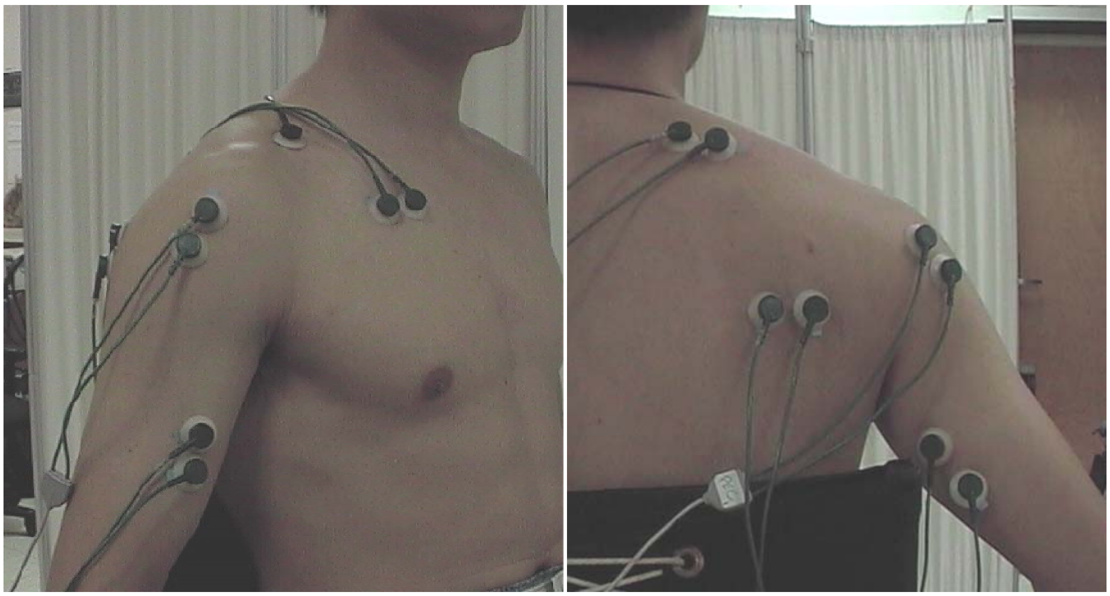
After these preparations, the actual data collection of wheelchair propulsion was initiated. Nine reflective markers were placed on the bony landmarks of each participant on thorax and the right side of scapula and humerus with adhesive tape (Figure 2). The three landmarks on the thorax segment were (1) C7, cervical spinous process of C7; (2) T8, thoracic spinous process of T8; and (3) lateral side of neck. Three landmarks on the scapula were (1) SS, spine of the scapula, midpoint of triangular surface on medial border of the scapula in line with the scapular spine; (2) IA, inferior angle, most caudal point of scapula; and (3) AA, acromial angle, most latero-dorsal point of scapula. Three landmarks on the humerus were (1) GH, glenohumeral rotation center, estimated by motion recordings; (2) EL, the most caudal point on lateral epicondyle; and (3) olecranon process. Furthermore, as reference points, two additional markers were fixed on the right wheel axle and third metacarpophalangeal (MP) joint to identify the point of hand contact on the pushrim (modified from Veeger et al., 1991; Rodgers et al., 1994; Boninger et al., 1997; Dallmeijer et al., 1998; Rodgers et al., 1998).

Figure 2. Reflective marker positions for kinematic data collection.



After the skin being shaved if needed, and cleaned with alcohol and applied conductive gel, surface electrodes were spaced on the muscles that were tested (Figure 3). After an upper extremity warm-up which consisted of stretching of neck, both shoulders, elbows and hands, each participant sat in the instrumented wheelchair which was secured on the stationary roller system. The same warm-up routine was employed by each participant for three minutes. Before the wheelchair propulsion tests, five seconds of electromyographic data were collected with participants seated at rest to determine background noise level inherent to the acquisition system. Then the electromyographic activity elicited during manual muscle tests at maximal effort (maximum voluntary contraction) was recorded. Muscle testing was modified to allow all muscles to be assessed with the subject seated in the wheelchair in the following positions: anterior deltoid in 45° of shoulder flexion with downward force applied to the elbow; posterior deltoid in 90° of shoulder abduction with forward force applied to the elbow; sternal pectoralis major in 90° of shoulder abduction with instruction to pull toward the contralateral knee resisted with upward and outward force applied to the elbow; middle trapezius in 90° of shoulder abduction and full external rotation with forward force applied to the elbow; infraspinatus in a neutral shoulder position and 90° of elbow flexion with inward force applied to the wrist; biceps long head in a neutral shoulder position, 90° of elbow flexion and full supination with downward force applied to the wrist; and triceps long head in 90° of shoulder abduction, full internal rotation, and 45° of elbow flexion with downward force applied to the wrist and stabilization proximal to the elbow (Mulroy et al., 2004). During testing each participant's trunk and wheelchair were stabilized by the investigator. The maximum one second of mean EMG recorded during the manual muscle tests served as the normalized value of 100%.

Figure 3. EMG electrodes positions for muscle activity measurements.



The participants then were asked to propel the wheelchair at 0.9 m/s and 1.8 m/s respectively, based on the normal and fast speeds reported by Boninger, Baldwin, Cooper, Koontz and Chan (2000) and Rodgers et al. (2001) in each of two stroke patterns (single loop and semicircular in a random order). The order of testing conditions (2 speeds x 2 patterns) was randomly assigned. The propulsion speeds were monitored by the Cat Eye speedometer and participants were given feedback by looking at the meter which was placed in front of them.

Once the participant reached the target speed and maintain for at least three seconds, the imaging and EMG systems started synchronically to record twenty seconds of wheelchair propulsion, which included approximately 22-25 complete cycles. Three valid trials of EMG and kinematic data for each selected speed and stroke pattern were collected from each participant. A 60-second rest was given between each trial. Three trials recorded for each condition were sampled at 100 Hz for the imaging system and 200 Hz for the final EMG data. The entire testing time was approximately one hour for each participant.

3.3.1 Determination of Glenohumeral Joint Center

The glenohumeral joint center (GH) was determined based on kinematic recordings. The movement of the humerus relative to the scapula was obtained from measurement in which both the scapula and the humerus were recorded. For the scapula, this was done by placing three markers on the acromion. From the relative rotations and replacements of the humerus to the scapula, a rotation center was estimated by the calculation of the intersection of instantaneous helical axes (Veeger, van der Helm, Chadwick & Magermans, 2003).

3.3.2 Definition and Transformation of Segmental Coordinate Systems

Each coordinate system for the thorax, scapula and humerus was defined by three anatomical landmarks and an origin (Veeger, van der Helm, Chadwick & Magermans, 2003). The global and local (segmental) coordinate systems were presented in Figure 4.

Thorax coordinate system: *X*t*Y*t*Z*t

*O*t: the origin coincident with the C7

*Z*t: the line connecting T8 and C7, pointing upward

*Y*t: the line perpendicular to the plane formed by the C7, T8, and neck marker, pointing to the left

*X*t: the common line perpendicular to the *Z*t- and the *Y*t-axes, pointing forward

Scapula coordinate system: *X*s*YsZ*s

*O*s: the origin coincident with the acromial angle (AA)

*Y*s: the line connecting AA and spine of the scapula (SS), pointing to SS

*X*s: the line perpendicular to the plane formed by inferior angle (IA), AA and *Y*s, pointing forward

*Z*s: the common line perpendicular to the *Y*s- and the *X*s-axes, pointing upward

Humerus coordinate system: *X*h*YhZh*

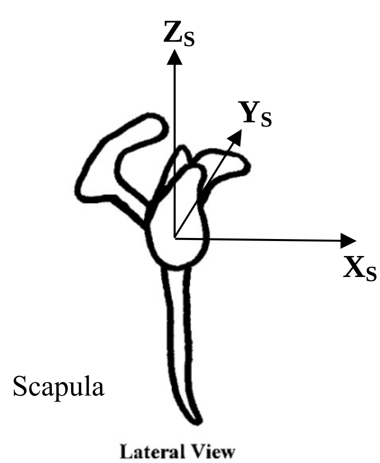
*O*h: the origin coincident with the glenohumeral rotation center (GH)

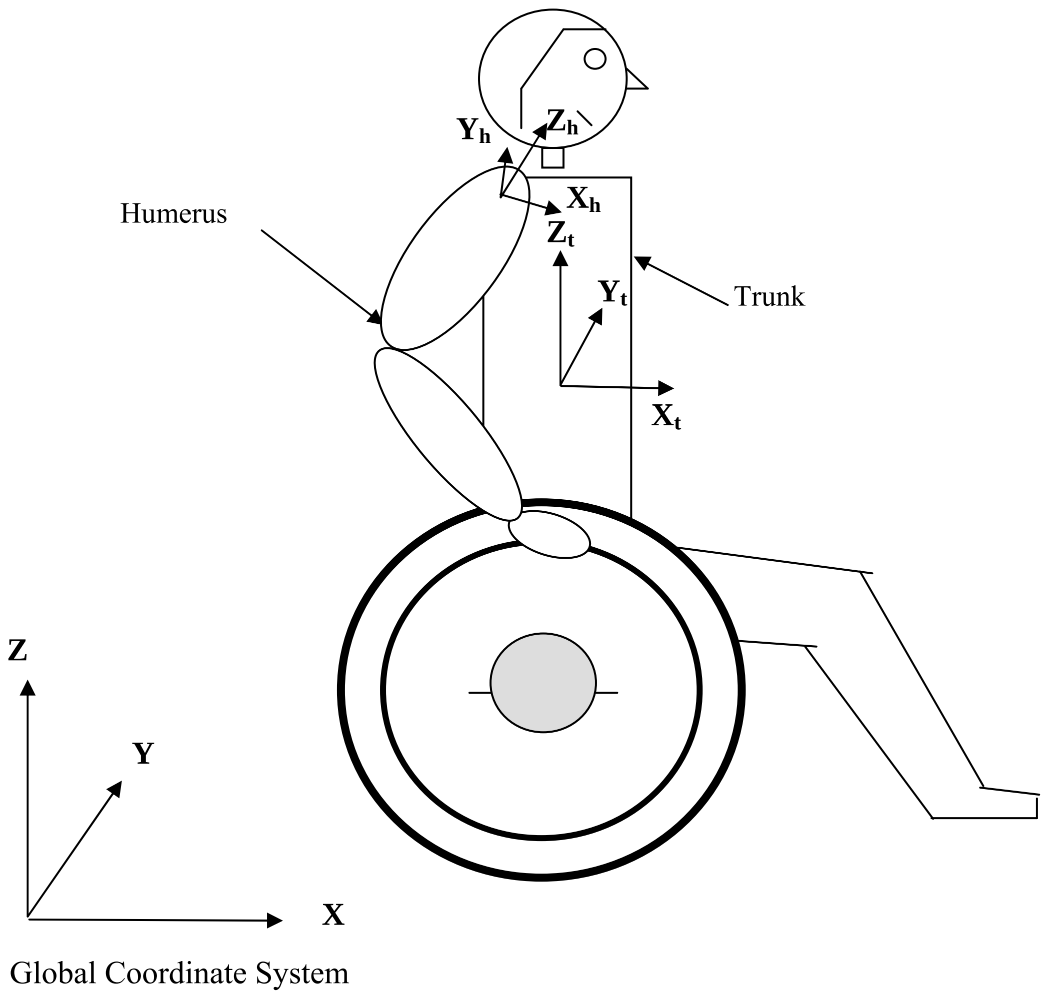
*Z*h: the line connecting the GH and lateral epicondyle (EL), pointing to GH

*X*h: the line perpendicular to the plane formed by the EL, olecranon and *Z*h, pointing forward

*Y*h: the common line perpendicular to the *Zh*- and the *Xh*-axes

Figure 4. Global and local coordinate systems





Euler angles were used to define three-dimensional kinematics of the scapula with respect to the thorax (scapulothoracic motion), and the humerus with respect to the scapula (glenohumeral motion) (Finley, McQuade & Rodgers, 2005). The Euler angles with $Z-Y-X$ rotation sequence were used in this study to transform the global coordinate system into segmental local coordinate systems. The Euler's $Z-Y-X$ rotational sequence indicated the relationship between the proximal and distal unit vectors, which were expressed in Equation 1.

$$\begin{bmatrix} i_d \\ j_d \\ k_d \end{bmatrix} = \text{Rot (X, } \theta) \quad \text{Rot (Y, } \psi) \quad \text{Rot (Z, } \phi) \quad \begin{bmatrix} i_p \\ j_p \\ k_p \end{bmatrix} \tag{1}$$

The rotational matrices for the component rotations θ,ψ,ϕ were defined in Equation 2.

$$R^{\theta} = \begin{bmatrix} \cos(\theta) & \sin(\theta) & 0 \\ -\sin(\theta) & \cos(\theta) & 0 \\ 0 & 0 & 1 \end{bmatrix} \quad R^{\psi} = \begin{bmatrix} 1 & 0 & 0 \\ 0 & \cos(\psi) & \sin(\psi) \\ 0 & -\sin(\psi) & \cos(\psi) \end{bmatrix} \quad R^{\phi} = \begin{bmatrix} \cos(\phi) & \sin(\phi) & 0 \\ -\sin(\phi) & \cos(\phi) & 0 \\ 0 & 0 & 1 \end{bmatrix} \tag{2}$$

The coordinate transformation matrix was obtained by matrix multiplication (Equation 3)

$$R^{\theta,\psi,\phi} = \begin{bmatrix} \cos(\psi)\cos(\phi) & \cos(\theta)\sin(\phi) + \sin(\theta)\sin(\psi)\cos(\phi) & \sin(\theta)\sin(\phi) - \cos(\theta)\sin(\psi)\cos(\phi) \\ -\cos(\psi)\sin(\phi) & \cos(\theta)\cos(\phi) - \sin(\theta)\sin(\psi)\sin(\phi) & \sin(\theta)\cos(\phi) + \cos(\theta)\sin(\psi)\sin(\phi) \\ \sin(\psi) & \sin(\theta)\cos(\psi) & \cos(\theta)\cos(\psi) \end{bmatrix} \tag{3}$$

The transformation matrix was based on the degrees of freedom describing motion of each segment. The thorax, scapula and humerus were each modeled with three degrees of freedom (rotations about *x, y, and z* axes).

The solutions for the Euler angles were derived based on directional cosines expressed in Equation 4 (a) - (c).

$$\text{(a)} \qquad \cos(\phi) = \frac{a^2 + c^2 - b^2}{2ac}$$

$$\text{(b)} \qquad \cos(\psi) = \frac{a^2 + b^2 - c^2}{2ab} \tag{4}$$

(c) $$\cos(\theta) = \frac{b^2 + c^2 - a^2}{2bc}$$

When the Euler angles were calculated for each segment and at each instant in time, the angular velocities and accelerations were then derived.

3.3.3 Determination of Pushing Phase and Recovery Phase

A conventional wheelchair propulsion cycle consisted of a drive phase and a recovery phase (Sanderson & Sommer, 1985). The drive phase was defined as the portion of time that the hand was in contact with the pushrim. The recovery phase was defined as the portion of time that the hand was not in contact with the pushrim and was returning to the start position to initiate another drive phase (Sanderson & Sommer, 1985).

In this study, the drive phase was defined as the portion of time from the initial contact of the hand on the pushrim to the end contact of the hand on the pushrim. The recovery phase was characterized as the non-propulsive phase when no force is exerted on the pushrim. Therefore, it was determined as the portion of time from the end contact of hand on the pushrim to the next initial contact of hand on the pushrim. The combination of the drive phase and recovery phase represented one propulsion cycle.

3.4 Statistical Analysis

In this study, the orientation of the thorax relative to the global coordinate system, and the orientation of the scapula and the humerus relative to the thorax were analyzed. The independent variables were the two stroke patterns of wheelchair propulsion. The dependent variables included both kinematic and EMG variables. In kinematics, there were (1) drive phase time, recovery phase time, and total cycle time, (2) drive phase time to total cycle time ratio, recovery phase time to total cycle time ratio, and drive phase time to recovery phase time ratio, (3) hand initial contact angles, release angles, and range on the pushrim, (4) three-dimensional joint angles and range of motion in each direction during two phases, (5) maximal and mean shoulder joint angular velocities and accelerations during two phases, and (6) maximal and mean linear velocities and accelerations during two phases. In electromyography, for each muscle, there were (1) average muscle activity during drive phase, (2) maximal muscle activity during drive phase, (3) average muscle activity during recovery phase, and (4) maximal muscle activity during recovery phase.

All kinematic and EMG data were averaged from three cycles in each trial then averaged again from three trials. The statistical package of SPSS for Windows (version 16.0) was used for all statistical procedures. The group means of variable parameters were described and compared between two propulsion stroke patters and two wheeling speeds by using 2 x 2 two-way within-subject repeated measures ANOVA. All statistical analyses were determined for significance at the α level of 0.05.

CHAPTER 4

RESULTS

Twenty graduate physical therapy students (14 females and 6 males) participated in this study (Table 1). There were 14 third-year students, 4 second-year students, and 2 first-year students. The mean age of participants was 27.4 years with a standard deviation of 5.92 years. The mean body mass was 64.41 ± 9.37 Kg and the mean body height was 169.32 ± 9.12 cm. All 20 participants were right-handed. The results of power analyses for within-subject repeated measures ANOVA with given α, sample size and effect size showed a statistical power ranging from 0.8-1.0 for all statistical procedures.

Table 1. Participants demographics (n=20)

SUB#	gender	age(y/o)	Body weight (kg)	Body height (cm)	year	dominant hand
1	F	25	60.78	172.72	3	R
2	F	26	53.07	157.48	3	R
3	F	25	58.97	170.18	3	R
4	M	25	77.11	172.72	3	R
5	F	28	58.97	166.37	3	R
6	F	25	72.57	171.5	3	R
7	M	26	63.50	165	3	R
8	M	47	75.75	179	2	R
9	F	24	58.97	172.09	3	R
10	F	24	54.43	152.4	3	R
11	F	25	64.86	176.53	1	R
12	M	24	68.04	180.34	2	R
13	F	29	63.05	167.64	3	R
14	F	26	62.60	167.64	3	R
15	F	24	58.06	162.56	1	R
16	F	27	61.23	171.45	3	R
17	M	41	83.91	187.96	2	R
18	F	27	49.44	149.86	3	R
19	F	26	61.23	167.64	2	R
20	M	24	81.65	175.26	3	R
mean		27.4	64.41	169.32		
SD		5.92	9.37	9.11		

4.1 Kinematic Analysis

4.1.1 Temporal Variables

It was found that when the wheeling speed increased, the drive phase time (DT) and the recovery phase time (RT) both decreased (p = 0.000) (Table 2). When compared between two stroke patterns, it was found that the semicircular pattern had a longer drive phase (p = 0.004) and a shorter recovery phase (p = 0.005). However, there were no significant differences in the total cycle time (CT) between these two patterns (p = 0.268).

Table 2. Drive phase time, recovery phase time, and total cycle time (seconds). S-S: slow speed, single loop pattern; F-S: fast speed, single loop pattern; S-C: slow speed, semicircular pattern; F-C: fast speed, semicircular pattern

	S-S	F-S	S-C	F-C	P (speed)	P (pattern)
DT (sec)	0.450±0.075	0.274±0.043	0.498±0.062	0.304±0.045	.000*	.004*
RT (sec)	0.611±0.089	0.460±0.129	0.528±0.084	0.394±0.078	.000*	.005*
CT (sec)	1.061±0.143	0.734±0.164	1.026±0.133	0.698±0.105	.000*	.268

For the time ratios, it was found when the wheeling speed increased, DT/CT decreased (p = 0.000) and RT/CT increased (p = 0.000), which means in each cycle (100 %), less percentage of time was spent on the drive phase when wheeling speed went faster (Table 3). The DT to RT ratio was greater for the semicircular pattern when compared to the single loop pattern (p = 0.000), which means in each cycle (100 %), more percentage of time was spent on the drive phase for the semicircular pattern.

Table 3. Drive phase time/cycle time (%), recovery phase time/cycle time (%), and drive phase time/recovery phase time ratio

	S-S	F-S	S-C	F-C	P (speed)	P (pattern)
DT/CT (%)	42.4±3.7	38.1±4.9	48.7±2.9	43.7±4.4	.000*	.000*
RT/CT (%)	57.6±3.7	61.9±4.9	51.3±2.9	56.3±4.4	.000*	.000*
DT/RT	0.742±0.109	0.624±0.129	0.955±0.112	0.788±0.144	.000*	.000*

4.1.2 Pushrim Angles

In all conditions, initial hand contacts occurred behind the TDC (top-dead center), while hand release occurred in front of the TDC (Figure 5). When the wheeling speed increased, the decreased release angle (p = 0.005) caused the increase of propulsion angle range (= initial contact angle - release angle) (p = 0.02) while the initial contact angle did not change (Table 4). When compared between two stroke patterns, the differences of initial contact angles between patterns were not significant (p = 0.938). The release angles were smaller for the semicircular pattern (p = 0.000), which caused the larger propulsion angle range when compared to the single loop pattern (p = 0.000).

Figure 5. Hand initial contact angle, release angle, and range

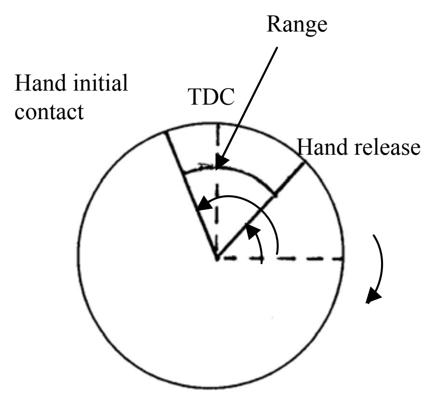


Table 4. Hand contact angle, release angle and range (°)

	S-S	F-S	S-C	F-C	P (speed)	P (pattern)
Contact angle (°)	119.4±12.7	123.1±12.0	123.5±12.6	119.6±13.9	.948	.938
Release angle (°)	40.8±10.2	37.8±13.5	27.6±11.9	21.1±11.1	.005*	.000*
Range (°)	78.6±14.8	85.3±14.9	95.9±12.1	98.4±12.8	.02*	.000*

4.1.3 Joint Angles

In this study, scapular and shoulder angles including minimal angles, maximal angles and range of motion were compared between two patterns and two speeds for each anatomical direction and two phases. For scapular angles, X direction represents scapular upward (positive) and downward (negative) rotations in the frontal plane, Y direction represents scapular tilting up (positive) and downward (negative) movements in the sagittal plane, and z direction represents scapular protraction (positive) and retraction (negative) motions in the transverse plane. For shoulder angles, X direction represents upper arm abduction (positive) and adduction (negative) in the frontal plane, Y direction represents flexion (positive) and extension (negative) movements in the sagittal plane, and Z direction represents humeral internal (positive) and external (negative) rotations in the axial plane. The shoulder angles also including minimal angles, maximal angles and range of motion and they were compared between two patterns and two speeds for each direction and two phases (drive and recovery phases).

When compared between two speeds, the results showed that scapular angles decreased when the wheeling speed increased (Table 5). In the drive phase, the minimal tilting angle (p = 0.049), and maximal tilting angle (p = 0.035) and protraction (p = 0.004) showed a significant decrease when the wheeling speed increased. In the recovery phase, minimal tilting angles (p = 0.003), and protraction angles (p = 0.012) also showed a significant decrease with the fast speed. However, for shoulder angles, some angles increased when the wheeling speed increased while some angles decreased (Table 6).

When compared between two patterns, it was found during the drive phase, the minimal scapular protraction angles for the semicircular pattern were larger than the protraction angles for the single loop pattern (p = 0.047), and the ranges of motion in protraction were smaller (p = 0.043). Also in the drive phase, maximal shoulder abduction angles (p = 0.012) and overall ranges of motion of abduction (p = 0.000) were larger for the semicircular pattern. In the recovery phase, the semicircular pattern showed larger angles in minimal shoulder abduction (p = 0.001) and internal rotation (p = 0.000) when compared to the single loop pattern.

4.1.4 Linear Velocity and Acceleration

The mean and maximal linear velocities and accelerations during the drive phase and the recovery phase in three directions were analyzed and compared between two patterns and two wheeling speeds for each marker, including shoulder, elbow and the 3rd metacarpophalangeal joint. Three directions included 1) X direction: anterior (positive) and posterior (negative) directions, 2) Y direction: medial (positive) and lateral (negative) directions, and 3) Z direction: upward (positive) and downward (negative) directions.

Table 5. Scapular angle: X: upward/downward rotation; Y: anterior/posterior tilting; Z: protraction/retraction (°)

			S-S	F-S	S-C	F-C	P (pattern)	P (speed)
DP	min	X	6.53±3.50	5.46±2.88	5.41±2.97	5.07±2.84	.135	.197
		Y	22.74±5.5	21.56±6.1	22.83±6.9	21.99±6.9	.597	.049*
		Z	21.19±5.7	20.85±6.2	21.90±6.9	21.31±6.9	.047*	.144
	max	X	12.96±4.8	11.57±3.8	11.41±4.2	12.11±4.3	.339	.466
		Y	27.95±6.1	26.55±6.6	27.85±7.5	27.44±7.1	.350	.035*
		Z	26.98±6.8	26.05±6.8	26.92±7.4	26.02±7.0	.903	.004*
	range	X	6.42±3.13	6.11±3.20	6.01±3.02	7.04±4.05	.427	.442
		Y	5.21±2.59	4.99±2.91	5.02±2.77	5.46±2.98	.731	.69
		Z	5.78±2.48	5.19±2.67	4.93±2.06	4.65±1.87	.043*	.105
RP	min	X	6.39±3.4	5.74±2.40	5.38±2.90	4.92±2.50	.059	.303
		Y	22.26±4.8	20.47±6.1	23.06±7.1	22.19±6.5	.033*	.003*
		Z	20.98±5.6	20.03±5.8	22.36±6.9	21.67±6.8	.007*	.012*
	max	X	13.66±5.5	13.24±5.7	11.27±4.2	13.03±8.4	.261	.556
		Y	28.62±6.7	27.9±7.02	28.04±7.5	29.11±9.2	.743	.837
		Z	27.44±7.2	26.99±6.7	27.12±7.5	28.6±11.4	.550	.613
	range	X	7.27±4.08	7.50±5.72	5.89±3.02	8.10±7.57	.694	.209
		Y	6.35±3.72	7.43±3.74	4.98±3.08	6.92±6.99	.405	.085
		Z	6.63±4.18	6.95±4.14	4.76±2.17	6.99±8.91	.467	.229

Table 6. Shoulder angle: X: abduction/adduction; Y: flexion/extension; Z: rotation (°)

			S-S	F-S	S-C	F-C	P (pattern)	P (speed)
DP	min	X	32.07±5.27	35.02±6.68	34.89±5.69	37.52±8	.057	.01*
		Y	-15.17±5.41	-16.57±6.37	-16.17±6.33	-19.01±10.0	.311	.094
		Z	40.76±6.28	38.27±7.95	37.91±7.05	32.78±6.63	.021*	.003*
	max	X	74.40±9.28	73.46±10.39	78.43±8.63	75.97±9.08	.012*	.179
		Y	36.61±8.57	32.12±10.81	33.54±7.08	26.04±7.78	.027*	.003*
		Z	60.31±4.67	60.08±5.58	59.52±4.27	59.39±4.56	.305	.78
	range	X	42.46±8.41	38.48±7.81	43.32±9.30	48.49±10.78	.000*	.295
		Y	51.78±8.52	48.69±10.54	49.71±6.67	45.05±11.58	.212	.162
		Z	19.55±6.74	21.81±8.95	21.61±8.06	26.61±7.98	.095	.011*
RP	min	X	15.49±12.36	14.3±1.86	33.8±3.90	36.83±5.38	.001*	.064
		Y	-15.44±6.01	-16.88±6.68	-15.12±5.46	-16.95±7.11	.918	.173
		Z	17.15±8.73	14.02±8.43	37.67±7.16	32.65±6.57	.000*	.001*
	max	X	78.74±7.57	78.42±10.01	78.10±7.88	74.79±8.08	.053	.601
		Y	55.93±11.46	64.02±15.24	56.16±7.31	62.37±6.88	.799	.000*
		Z	59.71±5.15	60.52±5.87	61.44±4.22	59.33±4.12	.797	.252
	range	X	63.22±9.64	64.12±18.17	44.30±7.04	37.96±10.80	.005*	.056
		Y	70.49±9.46	80.90±13.19	71.28±7.17	79.42±9.03	.832	.011*
		Z	42.55±7.86	46.50±9.42	23.76±7.17	26.67±7.95	.001*	.006*

Linear velocities and accelerations of the shoulder marker increased when the speed increased (p = .000 -.003) (Table 7-8). When compared between two patterns, during the drive phase the maximal medial and mean downward linear velocities for the semicircular pattern were larger than the velocities for the single loop pattern (*p* = .04 and .012). The maximal medial linear accelerations of the shoulder marker were also significantly larger for the semicircular pattern (*p* = .015) in the drive phase. In the recovery phase, the maximal posterior linear velocities (*p* = .021) and mean posterior (*p* = .001), lateral (*p* = .001) and upward velocities (*p* = .001) were larger for the semicircular pattern when compared to the single loop pattern. The maximal and mean posterior linear accelerations (*p* = .03 and .01) as well as the mean lateral linear accelerations (*p* = .03) were also larger for the semicircular pattern in the recovery phase.

Table 7. Linear velocity (m/s) of shoulder marker. X: anterior (+)/posterior (-) direction, Y: medial (+)/lateral (-) direction, Z: up (+)/down (-) direction

			S-S	F-S	S-C	F-C	P (pattern)	P (speed)
DP	max	X	0.164±0.069	0.3±0.142	0.202±0.119	0.32±0.135	.199	.000*
		Y	0.139±0.051	0.202±0.097	0.172±0.063	0.231±0.094	.04*	.000*
		Z	0.169±0.045	0.237±0.075	0.172±0.079	0.274±0.093	.189	.000*
	mean	X	0.1±0.047	0.18±0.092	0.123±0.079	0.185±0.092	.336	.000*
		Y	0.063±0.027	0.124±0.063	0.087±0.04	0.138±0.061	.067	.000*
		Z	0.069±0.018	0.115±0.044	0.087±0.045	0.152±0.054	.012*	.000*
RP	max	X	0.162±0.069	0.238±0.102	0.216±0.127	0.281±0.135	.021*	.003*
		Y	0.133±0.053	0.202±0.093	0.159±0.066	0.222±0.083	.145	.000*
		Z	0.128±0.038	0.248±0.079	0.178±0.07	0.238±0.064	.140	.000*
	mean	X	0.084±0.04	0.124±0.058	0.124±0.074	0.164±0.074	.001*	.002*
		Y	0.062±0.028	0.101±0.045	0.088±0.036	0.13±0.049	.001*	.000*
		Z	0.058±0.017	0.103±0.033	0.088±0.04	0.125±0.043	.001*	.000*

Table 8. Linear acceleration (m/s^2) of shoulder marker. X: anterior (+)/posterior (-) direction, Y: medial (+)/lateral (-) direction, Z: up (+)/down (-) direction

			S-S	F-S	S-C	F-C	P (pattern)	P (speed)
DP	max	X	1.593±0.47	3.457±1.015	1.984±0.888	3.769±1.17	.079	.000*
		Y	1.706±0.552	2.441±0.793	1.911±0.646	2.97±1.162	.015*	.000*
		Z	2.754±0.873	3.904±1.261	2.401±0.889	4.493±1.982	.642	.000*
	mean	X	0.683±0.22	1.885±0.618	0.804±0.381	1.865±0.575	.568	.000*
		Y	0.622±0.116	1.135±0.324	0.665±0.195	1.246±0.387	.198	.000*
		Z	0.979±0.287	1.812±0.611	0.909±0.329	1.898±0.626	.937	.000*
RP	max	X	1.715±0.65	3.019±0.714	1.927±0.855	3.539±1.134	.03*	.000*
		Y	2.214±0.949	3.151±1.38	2.047±0.711	3.469±1.227	.645	.000*
		Z	2.906±0.938	4.996±1.585	2.262±0.803	4.628±1.927	.075	.000*
	mean	X	0.721±0.301	1.431±0.465	0.927±0.491	1.661±0.597	.01*	.000*
		Y	0.688±0.219	1.246±0.455	0.749±0.303	1.449±0.492	.03*	.000*
		Z	0.833±0.272	1.91±0.55	0.887±0.327	1.815±0.54	.804	.000*

It was found that elbow linear velocities and accelerations increased when the speed increased (p = .000 -.001) (Table 9-10). In the drive phase, the downward linear velocities ($p \leq .007$) were larger for the semicircular pattern compared to the single loop pattern. However, the anterior and medial linear velocities were smaller for the semicircular pattern (p = .005 - .01). In addition, maximal anterior linear accelerations were significantly smaller for the semicircular pattern (p = .000) during the drive phase. In the recovery phase, the maximal posterior linear velocities (p = .028) and lateral linear velocities ($p \leq .02$) were smaller for the semicircular pattern, while the maximal upward linear velocities and accelerations were larger for the semicircular pattern ($p \leq .006$). It was also found that the posterior and lateral linear accelerations were smaller for the semicircular pattern ($p = .000 - .028$).

Table 9. Linear velocity (m/s) of elbow marker. X: anterior (+)/posterior (-) direction, Y: medial (+)/lateral (-) direction, Z: up (+)/down (-) direction

			S-S	F-S	S-C	F-C	P (pattern)	P (speed)
DP	max	X	0.891±0.142	1.426±0.17	0.809±0.094	1.361±0.144	.01*	.000*
		Y	0.917±0.196	1.304±0.296	0.763±0.129	1.204±0.183	.006*	.000*
		Z	0.511±0.073	0.903±0.17	0.61±0.11	1.075±0.267	.001*	.000*
	mean	X	0.55±0.067	0.929±0.128	0.504±0.088	0.872±0.122	.005*	.000*
		Y	0.342±0.048	0.553±0.122	0.303±0.059	0.513±0.096	.009*	.000*
		Z	0.306±0.053	0.582±0.092	0.355±0.084	0.709±0.193	.007*	.000*
RP	max	X	0.904±0.168	1.156±1.034	0.796±0.169	1.107±0.255	.028*	.001*
		Y	0.691±0.183	1.269±0.797	0.521±0.143	0.86±0.225	.006*	.000*
		Z	0.469±0.109	0.804±0.302	0.671±0.155	0.979±0.278	.006*	.000*
	mean	X	0.448±0.063	0.683±0.157	0.483±0.096	0.692±0.132	.333	.000*
		Y	0.326±0.054	0.513±0.184	0.297±0.069	0.452±0.099	.020*	.000*
		Z	0.243±0.038	0.485±0.292	0.341±0.083	0.567±0.167	.067	.000*

Table10. Linear acceleration (m/s²) of elbow marker. X: anterior (+)/posterior (-) direction, Y: medial (+)/lateral (-) direction, Z: up (+)/down (-) direction

			S-S	F-S	S-C	F-C	P (pattern)	P (speed)
DP	max	X	9.785±1.73	19.81±4.46	8.638±1.65	17.14±2.66	.000*	.000*
		Y	10.64±2.912	19.32±4.391	10.96±2.914	21.39±3.93	.081	.000*
		Z	6.1±1.518	14.63±10.7	6.975±1.873	16.4±5.571	.302	.000*
	mean	X	3.351±0.488	8.678±1.571	3.315±0.614	8.861±1.49	.742	.000*
		Y	2.886±0.601	7.004±1.234	2.976±0.767	8.016±1.77	.014*	.000*
		Z	2.601±0.489	5.698±1.936	2.768±0.593	6.696±1.94	.041*	.000*
RP	max	X	9.107±2.133	18.554±3.38	7.658±1.533	16.148±3.8	.003*	.000*
		Y	12.302±2.66	22.393±3.99	10.669±2.83	20.721±4.7	.026*	.000*
		Z	6.492±1.777	13.154±4.64	8.026±2.402	16.811±6.2	.002*	.000*
	mean	X	3.777±0.726	7.557±2.407	3.295±0.926	6.348±2.59	.000*	.000*
		Y	3.115±0.542	7.95±4.767	2.5±0.906	5.983±2.80	.028*	.000*
		Z	2.355±0.582	5.811±2.776	2.796±0.948	6.011±2.45	.45	.000*

The linear velocities and accelerations of the 3rd MP joint marker increased when the speed increased (*p* = .000) (Table 11-12). In the drive phase, the mean anterior (*p* = .000) and medial (*p* = .001) linear velocities as well as the maximal anterior linear accelerations (*p* = .000) were smaller for the semicircular pattern when compared the single loop pattern. However, the downward linear velocities and accelerations were larger for the semicircular pattern (*p* ≤ .041). In the recovery phase, maximal velocities in all three directions (*p* = .001 - .011) and mean lateral (*p* = .003) velocities were smaller for the semicircular pattern when compared to the single loop pattern. It was also shown that the posterior and lateral linear accelerations for the semicircular pattern were smaller (p = .000 - .028) while the maximal upward linear accelerations showed a significant increase (p = .002).

Table 11. Linear velocity (m/s) of 3rd MP joint marker. X: anterior (+)/posterior (-) direction, Y: medial (+)/lateral (-) direction, Z: up (+)/down (-) direction

			S-S	F-S	S-C	F-C	P (pattern)	P (speed)
DP	max	X	0.864±0.132	1.535±0.117	0.828±0.074	1.506±0.128	.136	.000*
		Y	0.594±0.121	0.979±0.149	0.587±0.069	0.939±0.145	.372	.000*
		Z	0.707±0.14	1.116±0.254	0.922±0.19	1.412±0.283	.000*	.000*
	mean	X	0.648±0.041	1.153±0.086	0.601±0.053	1.006±0.091	.000*	.000*
		Y	0.415±0.054	0.725±0.109	0.39±0.052	0.605±0.089	.001*	.000*
		Z	0.299±0.072	0.523±0.132	0.408±0.085	0.745±0.179	.000*	.000*
RP	max	X	1.195±0.322	1.687±0.299	1.085±0.226	1.505±0.323	.011*	.000*
		Y	1.03±0.269	1.347±0.254	0.77±0.178	1.096±0.207	.001*	.000*
		Z	1.04±0.42	1.433±0.535	0.721±0.141	1.008±0.283	.003*	.000*
	mean	X	0.65±0.115	0.954±0.143	0.664±0.138	0.909±0.16	.554	.000*
		Y	0.564±0.151	0.765±0.148	0.451±0.104	0.622±0.128	.003*	.000*
		Z	0.493±0.246	0.662±0.318	0.367±0.082	0.575±0.18	.072	.000*

Table 12. Linear acceleration (m/s^2) of 3rd MP joint marker. X: anterior (+)/posterior (-) direction, Y: medial (+)/lateral (-) direction, Z: up (+)/down (-) direction

			S-S	F-S	S-C	F-C	P (pattern)	P (speed)
DP	max	X	14.33±2.378	25.81±6.032	9.695±2.192	17.77±3.29	.000*	.000*
		Y	7.011±2.395	12.84±3.472	6.837±1.73	11.29±3.02	.15	.000*
		Z	13.61±3.481	25.36±5.907	9.644±2.899	17.45±4.79	.000*	.000*
	mean	X	3.128±0.61	7.969±2.726	2.811±0.527	7.598±1.93	.194	.000*
		Y	1.854±0.437	4.538±1.257	2.342±0.479	5.016±1.28	.021*	.000*
		Z	3.693±0.54	8.962±1.554	4.076±0.963	9.488±1.90	.116	.000*
RP	max	X	15.585±3.56	27.102±5.78	12.111±3.03	22.09±6.16	.000*	.000*
		Y	9.346±2.132	16.798±3.10	7.904±2.173	13.098±3.9	.001*	.000*
		Z	13.378±4.07	25.315±6.36	7.045±2.076	14.58±5.65	.000*	.000*
	mean	X	5.156±1	10.78±3.346	5.213±1.668	10.3±3.496	.617	.000*
		Y	4.885±1.183	9.053±2.002	3.515±1.116	6.972±2.09	.000*	.000*
		Z	4.904±2.201	8.82±3.691	2.697±0.703	5.536±2.54	.001*	.000*

4.1.5 Shoulder Angular Velocity and Acceleration

The mean and maximal shoulder angular velocities and accelerations during the drive phase and the recovery phase in three directions were analyzed. Angular velocities and accelerations were compared between two patterns and two wheeling speeds for shoulder joint. Three directions were 1) X direction: abduction (positive) and adduction (negative) directions in the frontal plane, 2) Y direction: flexion (positive) and extension (negative) directions in the sagittal plane, and 3) Z direction: internal (positive) and external (negative) rotations in the axial plane. The shoulder angular velocities and accelerations increased when the speed increased ($p \leq .001$) (Table 13-14).

Table 13. Angular velocities of shoulder joint (°/sec). X: abduction (+)/adduction (-) directions, Y: flexion (+)/extension (-), Z: internal (+)/external (-) rotations

			S-S	F-S	S-C	F-C	P (pattern)	P (speed)
DP	max	X	177.75±28.1	341.77±74.4	205.37±48.5	418.4±107.3	.000*	.000*
		Y	159.28±20.1	225.53±28.3	139.21±24.9	227.3±36.9	.237	.000*
		Z	362.65±65.1	439.7±84.82	247.44±61.7	340.8±60.4	.000*	.000*
	mean	X	122.53±20.9	227.69±47.3	123.15±26.7	254.8±86.3	.165	.000*
		Y	80.30±9.06	137.74±18.9	74.25±16.2	142.6±34.2	.913	.000*
		Z	164.45±20.6	247.51±42.29	120.69±27.8	197.0±35.1	.000*	.000*
RP	max	X	230.11±42.0	340.95±67.53	214.16±59.6	360±110.4	.905	.000*
		Y	124.83±22.2	217.85±76.85	113.03±28.3	190.7±50.3	.063	.000*
		Z	276.76±55.5	458.24±255.7	197.86±61.4	273.4±73.2	.000*	.001*
	mean	X	102.43±18.6	165.19±45.16	116.63±30.3	201.1±73.3	.001*	.000*
		Y	73.16±10.35	117.93±34.26	72.38±16.90	116.2±29.3	.801	.000*
		Z	153.8±23.3	222.85±75.7	118.9±33.6	158.7±36.0	.000*	.000*

In the drive phase, the angular velocities of external rotation were smaller for the semicircular pattern when compared to the single loop pattern ($p = .000$) while the maximal adduction velocities were larger for the semicircular pattern ($p = .000$). Furthermore, adduction accelerations as well as the mean flexion angular accelerations were larger for the semicircular pattern when compared to the single loop pattern ($p \leq .016$). In the recovery phase, the angular velocities and accelerations of internal rotation direction were smaller for the semicircular pattern when compared to the single loop pattern ($p \leq .005$). The mean abduction velocities and maximal abduction accelerations were larger for the semicircular pattern ($p \leq .027$).

Table 14. Angular accelerations of shoulder joint (°/sec^2). X: abduction (+)/adduction (-) directions, Y: flexion (+)/extension (-), Z: internal (+)/external (-) rotations

			S-S	F-S	S-C	F-C	P (pattern)	P (speed)
DP	max	X	1716.5±499.3	3663.9±1202	1862.2±641.8	5018.3±238	.007*	.000*
		Y	1331.4±295.1	2222±550.3	1310.3±330.4	2533.7±631.8	.081	.000*
		Z	2951.1±818.9	4126.9±1259	2677.9±762.1	4057.5±910.4	.469	.000*
	mean	X	651.0±212.5	2071.8±695.1	874.5±301.7	2743±1124	.001*	.000*
		Y	504.5±121.9	1184.5±370.4	549.8±141.2	1380.2±395.9	.016*	.000*
		Z	1104.2±238.2	2242.5±691.7	924.0±260.4	2107.1±483	.178	.000*
RP	max	X	2019.8±538.2	3819.1±1122	2017.7±657.3	5168.4±2571	.027*	.000*
		Y	1536.3±233.8	2936.1±1239	1308.7±322.1	2538±621.3	.038*	.000*
		Z	3328.7±781.8	6073.7±3979	2685.1±796.9	3991.2±920	.005*	.001*
	mean	X	917.8±195.7	1994.3±803.8	867.5±334.4	2218.1±1326	.323	.000*
		Y	599.3±101.8	1430.7±716.4	516.1±186.9	1216.3±515	.074	.000*
		Z	1274.9±224.8	2676.6±1493	909.1±355.1	1584.1±534	.000*	.000*

4.2 Muscle Activity

Muscle activities in each muscle were compared between two speeds and between two patterns in terms of mean EMG in DP, maximum EMG in DP, mean EMG in RP, and maximum EMG in RP. The muscle activities were normalized and presented in percentage of MVC.

4.2.1 Pushing Phase Muscles

Push phase muscles included pectoralis major, anterior deltoid, infraspinatus, and the long head of triceps muscle. The mean and maximal EMGs in both phases for all muscles increased when wheeling speed increased ($p \leq 0.038$) (Table 15-18). In the drive phase, pectoralis major EMGs for the semicircular pattern was lower than that for the single loop pattern ($p \leq 0.002$). However, maximal anterior deltoid EMGs ($p = 0.036$) and triceps EMGs ($p = 0.000$) where higher for the semicircular pattern. During the recovery phase, mean triceps EMGs were also higher for the semicircular pattern ($p = 0.04$). There were no significant differences in infraspinatus muscle activity when comparing between two patterns.

Table15. EMG for pectoralis major muscle (% of MVC)

	S-S	F-S	S-C	F-C	P (speed)	P (pattern)
Mean_DP	5.15±4.29	13.46±8.47	3.75±2.94	9.05±6.25	.000*	.001*
Max_DP	27.45±23.16	66.41±35.99	20.63±16.6	47.33±28.26	.000*	.002*
Mean_RP	1.91±3.19	4.46±6.46	1.53±2.18	3.44±4.34	.006*	.118
Max_RP	16.4±29.6	32.77±45.88	8.97±11.21	21.79±24.01	.003*	.126

Table 16. EMG for anterior deltoid muscle (% of MVC)

	S-S	F-S	S-C	F-C	P (speed)	P (pattern)
Mean_DP	4.07±1.51	11.18±4.97	3.9±2.07	10.75±5.3	.000*	.536
Max_DP	19.19±7.35	42.84±19.17	21.04±10.23	50.13±24.1	.000*	.036*
Mean_RP	1.49±0.01	2.86±1.82	1.49±0.01	3.18±1.92	.000*	.478
Max_RP	8.09±3.52	18.32±10.39	8.69±4.89	22.21±12.19	.000*	.105

Table 17. EMG for infraspinatus muscle (% of MVC)

	S-S	F-S	S-C	F-C	P (speed)	P (pattern)
Mean_DP	2.97±1.61	8.27±4.51	2.69±1.96	6.25±4.88	.000*	.052
Max_DP	14.2±7.57	36.52±20.62	14.47±11.88	31.97±28.75	.000*	.431
Mean_RP	1.31±0.01	2.68±1.87	1.42±0.01	2.2±1.28	.000*	.352
Max_RP	6.45±3.56	16.32±14.04	7.96±4.95	13.22±8.47	.001*	.648

Table 18. EMG for triceps muscle (% of MVC)

	S-S	F-S	S-C	F-C	P (speed)	P (pattern)
Mean_DP	2.42±0.01	7.46±6.23	3.43±1.92	10.99±5.99	.000*	.000*
Max_DP	13.19±13.28	39.78±3.48	23.09±13.67	62.96±33.09	.000*	.000*
Mean_RP	2.18±1.62	4.5±3.82	3.3±1.72	4.66±3.65	.009*	.04*
Max_RP	13.06±10.83	28.51±23.52	22.46±23.25	31.16±29.81	.038*	.153

4.2.2 Recovery Phase Muscles

Recovery phase muscles included posterior deltoid, middle trapezius, and the long head of biceps muscle. The mean and maximal EMGs in both phases for all muscles increased when wheeling speed increased ($p \leq 0.015$) (Table 19-21). In the drive phase, posterior deltoid EMGs for the semicircular pattern were higher than muscle activities for the single loop pattern ($p \leq 0.009$). However, middle trapezius EMGs for the semicircular pattern were lower than muscle activities for the single loop pattern ($p \leq 0.039$). In the recovery phase, posterior deltoid EMGs were also higher for the semicircular pattern ($p = 0.004$). When comparing between two patterns, there were no significant differences found in the EMGs of biceps.

Table 19. EMG for posterior deltoid muscle (% of MVC)

	S-S	F-S	S-C	F-C	P (speed)	P (pattern)
Mean_DP	1.39±0.01	5.34±2.98	2.24±1.05	6.62±3.63	.000*	.009*
Max_DP	12.38±5.5	34.65±18.18	17.8±7.43	41.95±20.44	.000*	.006*
Mean_RP	4.45±2.65	6.86±4.47	6.25±3.41	8.07±4.66	.001*	.004*
Max_RP	22.65±12.23	34.69±16.44	29.36±13.85	39.68±19.55	.000*	.004*

Table 20. EMG for middle trapezius muscle (% of MVC)

	S-S	F-S	S-C	F-C	P (speed)	P (pattern)
Mean_DP	2.55±1.21	6.81±4.04	2.35±1.35	5.01±2.77	.000*	.037*
Max_DP	15.98±7.74	35.94±20.62	14.38±7.94	25.77±13.23	.000*	.039*
Mean_RP	5.99±2.57	8.63±4.8	6.84±4.03	8.29±4.1	.001*	.566
Max_RP	31.17±13.32	42.47±20.91	32.88±17.92	38.76±19.21	.006*	.628

Table 21. EMG for biceps muscle (% of MVC)

	S-S	F-S	S-C	F-C	P (speed)	P (pattern)
Mean_DP	1.79±1.12	4.88±3.49	2.26±1.89	6.46±8.67	.003*	.184
Max_DP	8.09±5.28	23.97±18.25	12.64±9.99	34.13±47.26	.009*	.104
Mean_RP	1.62±1.38	2.63±2.18	1.89±1.81	4.48±6.51	.012*	.127
Max_RP	9.85±13.02	13.68±11.01	11.06±11.99	23.01±28.71	.015*	.164

CHAPTER 5

DISCUSSION

Manual wheelchair users rely on their wheelchairs as a major means of locomotion and mobility. Wheelchair propulsion involves repetitive shoulder range of motion and muscular activities. Particularly, active stability over the shoulder girdle during trunk movements requires coordination between shoulder joint movements and muscle activities. It is believed that repetitive stress in the shoulder joint due to inefficient and uncoordinated muscle activation along with awkward shoulder joint position and movement during wheelchair propulsion contribute to shoulder injury in manual wheelchair users (Bayley et al., 1987; Helm & Veeger, 1996; Boninger et al., 1997; Kulig et al., 1998; Cooper et al., 1999).

Studies have reported that experienced wheelchair users optimize wheelchair mobility and function by adapting their stroke techniques and patterns while accomplishing task requirements (Robertson et al., 1996; Boninger, Souza, Cooper, Fitzgerald, Koontz & Fay, 2002). For example, a longer drive phase was found in experienced wheelchair users, which may decrease the repetition of shoulder stress without decreasing wheeling speed (Robertson et al., 1996). However, information provided to manual wheel users about the appropriate way to push their wheelchairs was limited.

The purpose of his study was to investigate three-dimensional shoulder joint kinematics and electromyography between two stroke patterns (semicircular and single loop patterns). All kinematic and EMG data were compared in order to find the effects of stroke patterns on shoulder position, movement patterns and muscle activities. To verify the findings of this study, and to compare with previous literature in terms of kinematic and electromyographic characteristics of wheelchair propulsion, the following are discussed: (1) temporal and spatial characteristics associated with the pushrim, (2) joint angles and angular accelerations (3) joint linear accelerations, and (4) muscle activities. These considerations enhance the understanding of the factors related to wheelchair stroke patterns.

It should be emphasized that studies investigating wheelchair propulsion between different stroke patterns are limited and it is challenging to compare the results from this study to the findings of other studies. These difficulties come from individual variations among our participants due to body size and gender difference (14 females and 6 males in this study). Most studies involved in different stroke patterns recruited experienced manual wheelchair users as their participants while in this study physical therapy students served as our testing subjects. The experience of wheelchair use varies among different participants even though each of them has received wheelchair propulsion instruction which is part of their physical therapy curriculum. It has been documented that experienced manual wheelchair users adapt their stroke techniques and patterns accompanied by the employment of wheelchairs. The longer they use the wheelchair as their major means of locomotion and mobility, the smoother and more consistent their body segment movements and muscle activities are during wheelchair propulsion. Furthermore, muscle strength and muscular activation patterns are different between experienced manual wheelchair users and physical therapy students due to physiological differences, such as disability and body composition, as well as experience of wheelchair use.

Therefore, it may be found that some of the results of kinematic and EMG variables in this study are different from the reports in literatures. This comes from the differences of sample populations between studies as well as the variations among participants in this study. In this study, there are significantly more female participants than male participants (14 versus 6). It indicates that our results represent the female population better than the male population. How females push wheelchairs in terms of body movements and muscle activities may be different from the way males do. This would make it difficult to compare our findings to other studies since most previous studies recruited more male participants than female participants.

5.1 Temporal and Spatial Characteristics Associated with the Pushrim

In this study, the semicircular pattern with the hand below the pushrim during the recovery phase is associated with a longer drive phase and more time spent in the drive phase relative to the recovery phase. These two findings may imply to less loads per unit of time during the drive phase, which may reduce the risk of shoulder injuries. A shorter recovery phase may lead to increased efforts in accelerating/decelerating movements during the recovery phase. Nevertheless, a shorter recovery path is the nature of the semicircular pattern which can be explained by the elliptical path. Since an elliptical path is associated with less extra hand movements during the recovery phase, increased efforts in accelerating/decelerating are not expected in the semicircular pattern. Although in this study, no difference in total cycle time was found between two stroke patterns, other studies reported a lower cadence and higher ratio of push time to recovery time related to the semicircular pattern (Sanderson & Sommer, 1985; Shimada et al., 1998; Boninger et al., 2002). The increased drive phase time with no change of total cycle time corresponds to the increase of hand contact angle range on the pushrim. It was accomplished by decreasing the hand release angle without changing the initial hand contact angle on the pushrim.

It is suggested that the increased drive phase time in the participants with the semicircular pattern is caused by the delay of hand release on the pushrim. During the late part of the drive phase, the hand continues propelling the wheel and moving down along with the pushrim until the hand cannot reach further and has to let go of the wheel. In order to follow the elliptical path associated with the semicircular pattern, the upper trunk leans forward during the drive phase so the hand can reach down. During the recovery phase, participants with the single loop pattern rise their hands forward, then redirect their hand movement so the hands can come back down on the pushrims. The decelerating follow-through movement during the early recovery phase is not seen in the participants with the semicircular pattern. The follow-through movement as well as the change of the direction of hand movements may result in the longer recovery phase time associated with the single loop pattern when comparing to the semicircular pattern. As a result, less muscle activities are required to slow down the arm or to change the direction of hand movement during the recovery phase. Less muscle activities could lead to decreased shoulder joint stress which may diminish the risk of shoulder joint pain. Although a lower cadence related to the semicircular pattern was found in other studies, this study did not find a difference of cadence between two stroke patterns. This may be attributed to the differences of participants among studies. Again, this study recruited able-bodied participants and comparisons between stroke patterns were based on within-subject analyses. Most previous studies recruited experienced manual wheelchair users and data were compared between different groups of participants.

5.2 Joint Angles and Angular Accelerations

In the drive phase, our results showed no significant difference between the two stroke patterns in maximal scapular protraction while the minimal protraction was larger in the semicircular pattern, which caused the smaller range of motion in protraction. In this study, it was expected to find larger scapular protraction in the single loop pattern which was associated with the larger shoulder flexion at the end of the drive phase. Although the larger scapular protraction was found in the single loop pattern, it was not statistically significant ($p = 0.903$). Even though the minimal protraction was larger in the participants with the semicircular pattern in the beginning of the drive phase ($p = 0.047$), the difference was small. These findings of scapular protractions may be affected by the coordinate system of the trunk defined in this study.

Veeger and Rozendal (1992) described the global shoulder kinematics of five able-bodied participants during the drive phase. It was showed in the drive phase that the shoulder started with a flexion from an extended position combined with abduction, changing to flexion and adduction. These findings were supported by Boninger, Cooper & Shimada (1998) and Newsam et al. (1999).

In this study, the semicircular pattern showed greater shoulder abduction during the early drive phase due to trunk flexion. Shimada et al. (1998) also reported significantly larger shoulder abduction/adduction ranges of motion in the subjects with the semicircular stroke pattern when compared to the other stroke patterns. The mean maximum shoulder abduction/adduction angles for their subjects were approximately 75 degrees, and the minimum angles were approximately 35 degrees. In this study, the mean maximum shoulder abduction/adduction angles for the semicircular pattern were 78.43 degrees at the slow speed and 75.97 degrees at the fast speed; the minimum angles were 33.80 degrees and 36.83 degrees, respectively. Although increased shoulder abduction is found in the semicircular pattern, this does not necessary lead to the high risk of shoulder pain, since in this study the smaller scapular protraction is associated with the semicircular pattern with no differences found in scapular tilting and shoulder internal rotation.

Three-dimensional shoulder kinematics in wheelchair propulsion has been reported by Rao, Bontrager, Gronley, Newsam & Perry (1996), Newsam et al. (1999), and Finley, Rasch, Keyser & Rodgers (2004). It was found that movements of the humerus in all three directions reached maximal and minimal positions during the recovery phase. The end range positions during the recovery phase were also found in this study. For example, in slow speed single loop pattern, shoulder abduction and internal rotation reached relative minimum at 15.5º and 17.2º, respectively, while the peak flexion was 55.9º during the early part of the recovery phase. During the later part of recovery, shoulder abduction and internal rotation reached relative maximum at 78.7º and 59.7º while the flexion decreased to a minimum value (-15.5º). The mean ranges of motion for shoulder abduction, flexion/extension and internal rotation were 63.2 º, 70.5 º, and 42.6º, respectively. Boninger, Cooper and Shimada (1998) reported shoulder motion at low speed (1.3 m/sec) to be within the range of 64 to -11° for flexion/extension in the sagittal plane, 21 to 47° for abduction/adduction and 54 to 91° for internal/external rotation. Rao et al. (1996) reported that mean humeral elevation and internal rotation reached relative minimum at 22.5º and 11.6º, respectively, while the peak humeral plane was 23.2º during the early part of the recovery phase at 40-42% of the cycle. During the later part of recovery at 93-95% of the cycle, humeral elevation and internal rotation reached relative maximum at 56.6º and 86.2º while the plane angle decreased to a minimum value (-57.3º). The mean ranges of motion for the humeral plane, rotation and elevation were 81.6, 76.1, and 35.4º, respectively (Rao et al., 1996). It is difficult to compare the results of shoulder kinematics between studies; and the differences between studies may come from the differences of pushing speeds, stroke patterns, participants and definitions of coordinate systems among studies.

It is suggested that accelerating and decelerating the arms during the recovery phase requires an amount of energy. Since the hand is not on the pushrim, this energy does not directly contribute to propelling the wheelchair (de Groot, Veeger, Hollander & van de Woude, 2005). Since energy is lost in decelerating and accelerating the arms during the recovery phase, a high cycle frequency would lead to more acceleration and deceleration of the arms and therefore a higher energy loss. Although in this study no differences in frequency is found between two patterns, a stroke pattern with lower frequency may save manual wheelchair users energy that must come from muscle activities (Boninger et al., 2002).

Shimada et al. (1998) found that subjects with the semicircular pattern had smaller shoulder abduction/adduction acceleration values, when compared to the subjects with the single loop stroke pattern. In this study, however, the semicircular stroke pattern was associated with higher shoulder abduction/adduction angular accelerations during both phases. During the recovery phase, shoulder angular accelerations in internal/external rotation and flexion/extension were lower with the semicircular pattern.

5.3 Linear Accelerations

In this study, it was found that during the recovery phase, the third MP joint marker had lower linear accelerations in all three directions. The results of linear accelerations showed that the semicircular stroke pattern was associated with lower linear accelerations over the shoulder and arm except for the shoulder movements on the transfer plane and elbow vertical movements. This can be explained by the nature of the semicircular stroke pattern which is associated with an elliptical path without extra hand movements.

Based on the findings from kinematic data, it can be concluded that the semicircular pattern has advantages in drive phase time and accelerations of the arms. When the cadence does not change, a longer drive phase may refer to less work per unit of time required to maintain the pushing speed. It means less muscle activities are required for propelling the wheel during the drive phase. A smaller acceleration value of the arms during the recovery phase indicates less energy loss even when the cadence does not change. According to these findings, it is suggested that during the entire cycle the semicircular stroke pattern may save manual wheelchair users energy that comes from muscle activities.

The lower cycle cadence which was found in previous studies did not show in the semicircular pattern in the current study. This may indicate that our participants have not developed proper propelling skill which is associated to the lower cadence. In addition, the comparisons were made between two stroke patterns within the same group. Some participants may do better with the single loop pattern while the others are good at the semicircular patter. Their experience in wheelchair propulsion can be affected by the years of study in physical therapy and where or how long they have been completed their clinical training. As a result, the future studies should recruit separate groups of experienced wheelchair users. Each participant would be good at only one stroke pattern. Another option will be training each able-bodied participant for only one stroke pattern then comparing the differences between training groups.

5.4 Muscle Activities

The EMG intensities of shoulder muscles during wheelchair propulsion have been reported in several studies, in combination with three-dimensional kinematic parameters to describe the role of different muscles (Veeger & Rozendal, 1992; Veeger et al, 1998; Finley et al, 2004; van Drongelen et al, 2005).

Trunk instability due to the absence or impairment of abdominal and back muscle control or the long period of sitting usually leads to an increased kyphotic posture with flattened lumbar spine among individuals with spinal cord injury (Hobson & Tooms, 1992, Yang et al., 2006). This functional sitting posture allows individuals with spinal cord injury to shift the trunk center of gravity back and secure it within their base of support without losing balance in a wheelchair (Curtis, Drysdale, Lanza, Kolber, Vitolo & West, 1999; Sinnott, Milburn & McNaughton, 2000; Samuelsson, Tropp & Gerdle, 2004). In addition, lack of trunk stability, which resulted in less erect posture and poor support of the shoulder girdle complex, may limit production of maximal shoulder strength (Powers, Newsam, Gronley, Fontaine & Perry, 1994). Due to the differences of participant population between this study and others, the results of muscle activities in this study are difficult to compare with the findings in other studies.

Mulroy et al. (1996) identified two synergies of shoulder muscle function during wheelchair propulsion. The push phase synergy was dominated by muscles with shoulder flexion (anterior deltoid, pectoralis major), external rotation (supraspinatus, infraspinatus) and scapular protraction (serratus anterior) functions. Pectoralis major and supraspinatus had the highest peak (58% and 67% MAX) and average (35% and 27% MAX) EMG intensities in this group. In this study, we tested pectoralis major, anterior deltoid and infraspinatus, three muscles in the push phase synergy muscle group. Our results also showed that pectoralis major had the highest peak and average EMG intensities in this group. However, the EMG amplitudes reported in this study was lower than the intensities reported by Mulroy et al. (1996) due to the difference of participant populations between the two studies. It is suggested that since our participants have better trunk control and usually lean their trunk forward during the drive phase which generates body momentum to assist pushing, less shoulder muscle activity is required for the propulsive force in the drive phase.

The dominant functions of the recovery synergy were extension (posterior deltoid), abduction (medial deltoid, supraspinatus), internal rotation (subscapularis) and scapular retraction (middle trapezius). This group of muscles had EMG onsets in late push (17% to 26% cycle) with moderate average intensities (21% to 32% MAX). In this study, we tested posterior deltoid and middle trapezius in the recovery phase synergy muscle group. Again, the EMG intensities found in this study were lower (4.5% to 8.6% MVC) than the intensities reported by Mulroy et al. (1996). It is suggested that the lower EMG intensities found in this study may be due to the difference of participant populations between the two studies. In this study, our able-bodied participants had better trunk control, which may diminish the co-contraction of shoulder muscles during the early stage of the recovery phase. When compared with two stroke patterns, less middle trapezius muscle activities were required for scapular retraction due to the smaller range of motion in scapular protraction during the drive phase. In addition, the semicircular pattern was associated with less shoulder extension combined with the trunk flexion which could link to the smaller scapular retraction during the recovery phase. More posterior deltoid muscle activities for shoulder extension against the gravity were also associated to the forward bending of the trunk in the semicircular pattern.

The push phase muscles were also activated in the late recovery phase to decelerate the back swing of the arm and to prepare the hand (increasing hand speed) for the contact on the pushrim (Mulroy et al., 1996). The same phenomenon has been described for the recovery phase muscles, activated already at the end of the push phase to restrain shoulder flexion (posterior deltoid), adduction and external rotation.

At the elbow joint, biceps brachialis was activated in the late recovery phase and continued its action over a period where an elbow flexion torque would contribute to the propulsion. After the hand has made contact with the pushrim, the pull phase starts with an initial elbow flexion, accompanied by activity of the biceps brachialis (Veeger, van der Woude & Rozendal, 1989). Likewise, triceps brachialis only became active when elbow extension would contribute to a propulsive force on the pushrim (Rodgers et al., 1994; Mulroy et al., 1996; Schantz, Björkman, Sandberg & Andersson, 1999; Guo, 2003). In this study, the semicircular pattern was associated with higher triceps muscle intensities. When triceps became active, the elbow was in a more flexed position due to the forward bending of the trunk during the drive phase. This position combined with an extended shoulder would result in more muscle efforts required on triceps in order to contribute to a propulsive force on the pushrim. In addition, it is believed that triceps brachialis is necessary for an effective force direction (Guo, 2003). Anterior deltoid has a high activity at the beginning of hand contact, whereas pectoralis major has a more constant activity of longer duration. These two muscles are considered to be the prime movers in wheelchair propulsion (Veeger, van der Woude & Rozendal, 1989; Rodgers et al., 1994; Mulroy et al., 1996; Guo, Zhao, Su & An, 2003). Thus, the higher anterior deltoid EMG intensities and lower pectoralis major EMG intensities in the semicircular pattern may be associated with the hand path during the end of recovery phase. Although there is an increasing abduction of the arm after the hand contact, it is suggested that this shoulder abduction is not an active movement (Veeger, van der Woude & Rozendal, 1989). It is assumed to be caused by the high activity of the prime movers at the start of the push phase, causing a flexion and internal-rotation torque in a closed chain (Veeger & Rozendal, 1992). In this study, it is noticed that the trunk leans down forward during the early stage of the drive phase, which may increase shoulder abduction. Increased shoulder abduction combined with extension would put anterior deltoid muscle in a stretched, elongated position, which may be associated with higher muscle activities in this muscle. It was suggested trunk flexion at initial hand contact could induce an internal rotation position at the shoulder joint accompanied by eccentric work of the external rotators. A repetitive condition of an internal rotation position, combined with high external rotator and abductor load, would be a serious risk factor for supraspinatus tendon impingement (Boninger, Cooper & Shimada, 1998).

It was concluded by Schantz et al. (1999) that the brachial biceps and triceps, anterior deltoid and pectoralis major muscles could be anticipated to propel the wheelchair forward, whereas the posterior deltoid and trapezius muscles could be expected to play a role, especially during the recovery phase. However, individual differences exist. Regardless of the type of recovery movement (semicircular and single loop), the posterior deltoid and trapezius muscles in particular were active during part of, or the whole, recovery phase (Schantz et al., 1999). The general order of activation of, first, the brachial biceps, thereafter the pectoralis major and anterior deltoid, and then the triceps brachial muscle during the push phase is constant with findings in other studies (Veeger, van der Woude & Rozendal, 1989; Mulroy, 1996).

5.5 Applications of the Study

Twenty physical therapy students who have been trained in wheelchair propulsion as part of physical therapy program of study were recruited in the current study. Although experience and the skill level might be different among participants, all participants must meet the requirements in order to be included in this study. In other words, each participant was verified as an expert in wheelchair propulsion by the competency assessment. It is suggested that performance of wheelchair propulsion is related to factors such as age, body type, disability type, and injury level. Therefore, based on the characteristics of our participants, the results of this study may be generalized to manual wheelchair users who have independent upper extremity function such as diabetics, lower-extremity amputees and lower-level paraplegics.

Since there is no documentation in the literature showing that techniques of wheelchair propulsion are different between two genders or are taught differently between genders, in this study we assume that our female and male participants employ the same techniques to accomplish the tasks. As a result, the findings from this study should be applied to both genders. However, the generalizability of our results to people with tetraplegia, high-level paraplegics, multiple sclerosis or severe obesity is limited. The limitation is due to the differences of body type, body composition, as well as muscle strength and functional level of upper extremity or trunk between our participants and these populations. In particular, our participants have much better trunk control than individuals with a high level of spinal cord injury. Trunk control is required when wheelchair users lean their upper body forward which is seen in the semicircular stroke pattern. These differences can result in the changes in kinematic and EMG variables so it might be erroneous to explain how these people maneuver their wheelchairs using the data from this study. It is also inappropriate to relate the results of this study to wheelchair propulsion in the elderly since most likely they propel wheelchairs with their hands and legs. In addition, muscle strength and flexibility (range of motion) declines as people age, which can affect the parameters in our analyses.

It is suggested that physical and occupational therapists, including students, educators and clinicians may benefit from the information in this study. Our results may provide a better understanding in wheelchair propulsion between different stroke patterns in terms of kinematics and electromyography, and provide some guidelines of what to emphasize on wheelchair skill training for patients and students. Since the semicircular stroke pattern with long strokes and an elliptical path is recommended for manual wheelchair users, clinicians and PT educators should spend time on teaching this particular pattern during the early stage of clinical rehabilitation and in school courses.

5.6 Limitations and Future Studies

This study has several limitations which may compromise the reliability, validity or generalizability of our findings. First, our findings were generalized from the comparisons between two stroke patterns. Although the results in this study demonstrate that the semicircular pattern is more advantageous, we don't know if this will still apply when we include double loop and arcing stroke patterns in the study. It is reported in the literature that the single loop pattern is the most popular pattern, followed by the double loop pattern. In order to have an acceptable statistical power (0.8) with a sample size of 20 people, we did not compare all of these four stroke patters. We chose the semicircular patterns instead of the other two was because in the literate this particular pattern has been reported to be the most beneficial pattern.

Second, convenient subjects were recruited in this study. Although each participant was qualified for the eligibility before data were collected, some participants had more experience than the others at the time of their participation. It was noticed that some participants pushed better in the semicircular pattern which might affect the results in general when comparisons were made within the same sample group. Additionally, our participants are physical therapy students who have been trained in wheelchair propulsion for weeks; therefore, it is difficult to generalize our results to long-term wheelchair users.

Third, some marker points were not based on true bony landmarks; therefore, real joint centers were not computed or used for retrieving kinematic data. Since the motion capturing system we used in this study had only two cameras, our marker points were limited to the sagittal view. We avoided the positions on the chest and the medial side of the arm since our cameras could not detect them. This would affect the orientation of coordinate system for each segment and the values of joint angle, angler velocity and angular acceleration.

Fourth, it was hard to precisely monitor the pushing speed during data collections. During each trial, participants were asked to maintain their pushing speed by looking at the speedometer. It was noticed that during the slow speed, sometimes the pushing speed was not picked up by the speedometer due to the inconsistency of the wheel speed. This inconsistency may result from the friction between wheels and the roller. When participants kept their hands on the pushrims during the drive phase, the propulsive force kept the wheels running; once their hands left the pushrims, the wheel speed decreased. At the end of the recovery phase, the wheel speed went down more, so the speedometer lost the speed before the sensor re-detected the magnet which was attached on the spoke of the right wheel. To eliminate the effects on kinematic data due to inconsistency of the pushing speed, three trials were collected for each condition. All kinematic data were averaged from three cycles in each trial then averaged again from three trials.

Last, the training of wheelchair propulsion was not controlled for this study. We don’t know how well our participants have been trained or either how long or how often they have practiced. We don’t know how well they know about wheelchair propulsion or their knowledge about wheelchair users. We did testify our participants to see if they could keep up the speed and stay consistent. However, we don’t know if it is good enough to determine an expert in wheelchair use solely based on the competency assessment used in this study.

From this study, we have learned that wheelchair propulsion is a complex task and involved in many factors which may affect the results of the study. To make the findings meaningful, it is required to control these factors so the variety among participants can be reduced. However, the generalizability of the findings may be compromised when most factors are controlled. It really depends on what the researcher wants to look and how broad the results need to be generalized. In this study, we did not have the control of wheelchair propulsion training and this could weaken the significance of this study. If this study could have been done differently, we would have conducted the training by ourselves, recruited more participants, and used a motion capturing system with at least six cameras. We would have four different groups of participants and train each group for only one of these four stroke patterns. The training time would be controlled for each group with a larger sample size. The number of participants in each gender would be equal in each group. Since six digital cameras would be used so markers could be attached on true anatomical landmarks in order to calculate true joint centers and determine real body segment orientations.

Future studies should be granted to recruit experienced manual wheelchair users matched with age, gender, injury level and body size for each group. For example, if individuals with spinal cord injury will be recruited, participants in each group should have their counterparts in the other groups to minimize the effects of related factors such as age, injury level and body size. Biomechanical characteristics such as shoulder joint reaction forces and muscle moments should be investigated and related to shoulder joint kinematics and electromyography in order to understand the mechanism of shoulder injuries.

5.7 Conclusion

It is logical that the semicircular pattern is advantageous because this pattern follows an elliptical pattern. An elliptical path avoids abrupt changes in direction and minimizes the need for extra hand movement. In this study, lower hand linear velocities and accelerations are found during the recovery phase. The results of scapulothoracic and glenohumeral joint angles only show an increase in shoulder abduction which is within the physiological limit when comparing between two stroke patterns.

Although in this study the semicircular pattern is associated with a longer drive phase and larger ratio of drive phase time to the recovery phase time, the total cycle time is found to be no different between the two patterns. For muscle activities, the EMG intensities of pectoralis major and middle trapezius are found to be lower during the drive phase while others are found to be higher. In conclusion, the semicircular pattern is recommended in wheelchair propulsion training for rehabilitation.

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