THE BIOMECHANICS AND NEURAL CONTROL OF MANUAL WHEELING: AN EXPLORATION OF CUTANEOUS REFLEXES

by

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ABSTRACT

There are many health complications resulting from manual wheelchair use after spinal cord injury (SCI). Biomechanical and neural control components are critical in teaching wheelchair skills and developing efficient wheeling strategies. Thus, the overall goal of this thesis was to gain a comprehensive understanding of the biomechanics and neural control underlying upper limb movements during manual wheeling.

A) Many studies examining the biomechanical and physiological characteristics of manual wheeling have examined able-bodied subjects, however, it is unknown if this data can be applied to manual wheelchair users (MWUs) with SCIs. Thirteen able-bodied subjects and 9 MWUs participated in this study. Kinetic, kinematic, and electromyography (EMG) data were collected while subjects wheeled for several minutes at a self-selected cadence. The MWUs demonstrated different wheeling strategies, significantly larger wrist range of motion, larger average forces, larger percentage of the wheeling strategy spent in propulsion and larger push angles. These differences may be key in developing effective wheeling strategies.

B) The neural modulation of upper limb movements during manual wheeling was investigated by examining reflex responses to cutaneous nerve stimulation. Cutaneous reflexes from the superficial radial nerve were elicited while subjects wheeled for several minutes at a self-selected cadence. Subjects also performed a symmetrical arm cycling task at the same cadence while receiving nerve stimulation. EMG was recorded from 6 upper limb muscles. The data were divided into cycles and then all cycles were divided into 8 chronological bins. All reflexes occurring from stimuli in a specific bin were averaged together for each individual and then reflex averages were determined for the able-bodied and MWU groups. No significant differences were found in the amount of reflex modulation between the groups, but there were significant differences between tasks in the early latency response of the triceps brachii and the middle latency response of the posterior deltoid. There was also a significant correlation in the amplitude of the early latency reflex of the triceps brachii between amount of modulation and years of manual wheeling experience. Manual wheeling, like arm cycling and walking, demonstrates examples of both phase dependent and task dependent reflex modulation.

TABLE OF CONTENTS

LIST OF TABLES

LIST OF FIGURES

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CO-AUTHORSHIP STATEMENT

The supervisory committee and all authors on the manuscripts were involved in the design of the research projects. The MSc candidate was responsible for collecting the data, analyzing the data, and writing the manuscripts. The supervisory committee and other authors of the manuscripts from this study provided direction, support and critical feedback on the design of the study. The supervisory committee provided critical feedback on the manuscripts that were prepared. The MSc candidate was responsible for writing the final content of all manuscripts.

1 AN INTRODUCTION TO THE BIOMECHANICS AND NEUROMECHANICS OF RHYTHMIC UPPERLIMB TASKS

1.1 Overview

Locomotion can be defined as the act of moving from one point to another. Humans, as well as other animals, have evolved to optimize the efficiency of locomotion and continue to adapt their locomotion strategies throughout life. Locomotion varies greatly from one animal to another, however all animal locomotion involves cyclical movements of limbs or body segments. Locomotion consists of many forms, from primitive oscillatory movements observed in the lamprey to complex quadrupedal and bipedal gait observed in cats and humans.

Not all people are able to partake in typical gait due either to congenital diseases and disorders or traumatic events to the spinal cord. Compensatory forms of human locomotion may include the use of walkers and crutches. When walking is not possible or too fatiguing, manual wheelchair use is often the most efficient form of locomotion. Manual wheelchair use is one of the most common and more permanent forms of adapted mobility. There are an estimated 155,000 Canadians living in private households who need a wheelchair (Shields 2004). One of the largest populations using manual wheelchairs are people with spinal cord injuries. More than 41,000 Canadians are living with a spinal cord injury (SCI) with approximately 1100 new cases each year (Rick Hansen Foundation). SCIs offer challenges for the individual as well as society. The individual is challenged to adapt to the change in their body, which often requires that they adjust or change their mode of mobility to maintain their independence while society must work to maintain accessibility and inclusion of people with physical disabilities. Society is affected by the high costs to the health care system including the large costs of wheelchairs. Every year, SCIs cost the Canadian healthcare system 750 million dollars (Rick Hansen Foundation). This value also includes overuse injuries to the shoulder joint as a result of manual wheeling.

Manual wheelchair locomotion offers many potential problems. Many manual wheelchair users (MWUs) will develop shoulder pain and injuries as a result of increased demand on the upper body. Many people dependent on manual wheelchairs will experience rotator cuff problems throughout their lives (Silfverskiold and Waters 1991). There is little understood about ideal wheeling strategies and the literature has provided conflicting results, therefore the guidelines for ideal wheeling strategies are vague (Consortium of Spinal Cord Medicine, 2005). There is also very little understood about whether adaptation or reorganization of the nervous system occurs with manual wheeling experience. Furthering our understanding of the neural control of upper limb movement during manual wheeling may help us better understand the neural adaptations that occur as a result of prolonged manual wheelchair use. In this study, we measured how sensory input is processed during movement (wheeling) to gain an understanding of the neural strategies and adaptations that accompany manual wheeling experience and different types of manual wheeling strategies.

1.2 Literature review

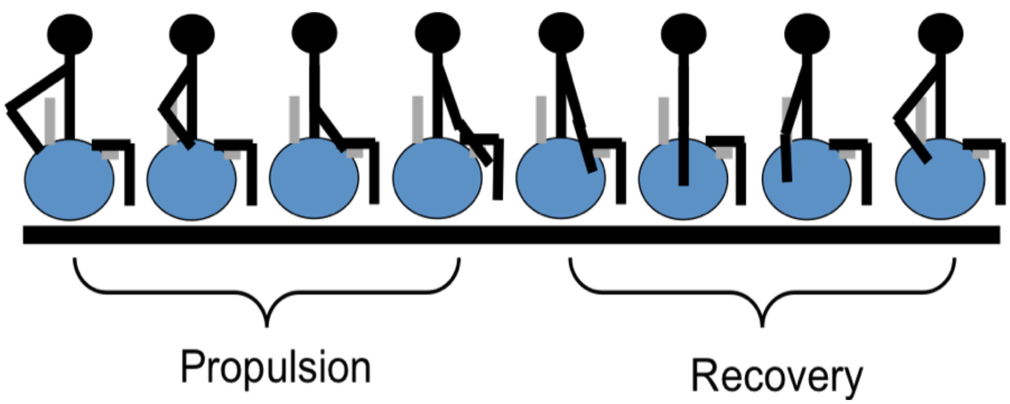
Manual wheelchairs are the main form of mobility for many people in Canada and the United States. Manual wheelchairs offer a unique form of mobility that is only dependent on the functioning of the upper limbs. The concept of a wheelchair is believed to date back to the 6th century B.C., as evident in a stone sculpture from China in which a chair on three wheels is carved (Kamenetz 1969). By the time of the 16th century AD, wheelchairs had gained popularity in Europe and were observed in drawings and literature (Kamenetz 1969). In 1896, the first wheelchair was patented in the United States in response to an increased demand for an adapted form of locomotion due to injuries resulting from the civil war (Hotchkiss 2000).

Although wheelchairs offer a means to improve mobility for people with lower limb disabilities, many if not most, MWUs will experience shoulder pain throughout their lives as a result of some element of wheelchair use. Manual wheeling as well as other wheelchair related tasks, including transfers, are thought to contribute to shoulder injury and pain. Any loss of upper arm function or associated problems puts the independence of a MWU at risk (Pentland and Twomey 1994; Curtis, Roach et al. 1995). Previous studies have shown that anywhere from approximately 31% to 73% of spinal cord injured MWUs will experience shoulder pain (Curtis, Drysdale et al. 1999; Boninger, Souza et al. 2002). Interestingly, it has also been found that people who begin using manual wheelchairs as children report less shoulder pain than people who began using wheelchairs as adults (Sawatzky, Slobogean et al. 2005). The lack of shoulder pain in people who began using a wheelchair as a child may be influenced by an increased mechanical efficiency developed earlier in life.

1.2.1 Wheeling strategies

The wheeling cycle is composed of two main phases: the propulsion phase (when the hands are in contact with the pushrim), and the recovery phase (when the hands are not in contact with the pushrim) (Figure 1.1).

Figure 1.1 Phases of the manual wheeling cycle

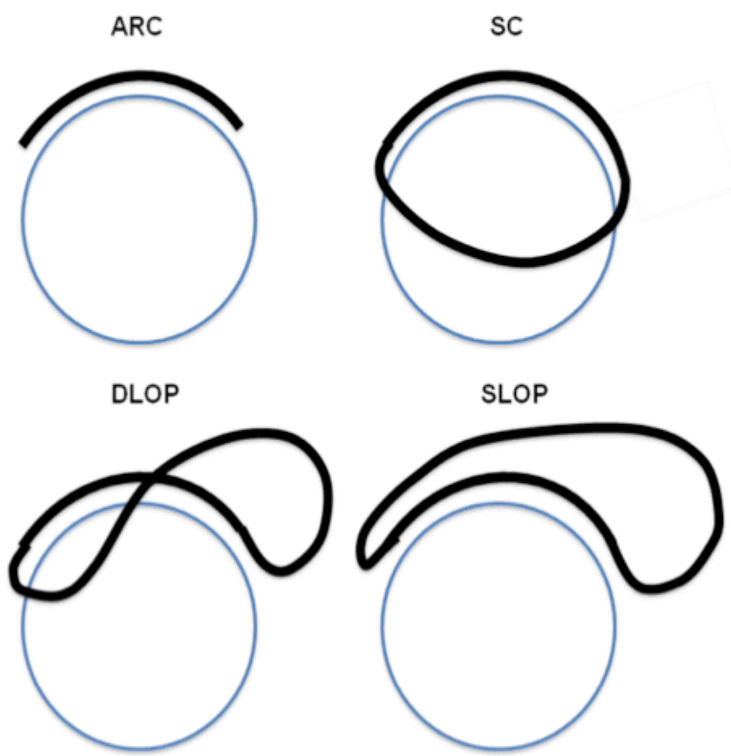


This figure represents the two distinct phases of manual wheeling. The first phase, propulsion, occurs when the hand is in contact with the pushrim and serves to push the wheel forward. The second phase of wheeling, recovery, occurs when the hand is not in contact with the pushrim. The recovery phase is also known as the gliding phase.

The specific movement pattern of the upper limbs during manual wheeling can take various forms depending on the strength and flexibility of the individual as well as the incline and speed of wheeling (Richter, Rodriguez et al. 2007). Previously, there were only three distinct wheeling strategies, which were categorized by the trajectory of the wrist through space during the recovery phase. These included *semi-circular* (hands falling below the propulsion pattern during the recovery period), *single looping over propulsion* (SLOP) and *double looping over propulsion* (DLOP) (Figure 1.2) (Shimada, Robertson et al. 1998).

The SLOP is defined by the hands rising above the path of the pushrim (Shimada, Robertson et al. 1998) while the DLOP consists of the hands initially rising above the pushrim and then dropping below the pushrim (Richter, Rodriguez et al. 2007). More recently, a fourth wheeling strategy, *arcing*, was identified (Boninger, Souza et al. 2002). The arcing pattern describes a recovery pattern that followed the wheeling strategy in reverse along the pushrim (Boninger, Souza et al. 2002).

Figure 1.2 Manual wheeling strategies



The four commonly classified wheeling strategies: arcing (ARC), semi-circular (SC), double looping over propulsion (DLOP), and single looping over propulsion (SLOP).

The wheeling strategy appears to be largely dependent on the incline of the wheeling surface rather than the speed of wheeling. In a recent study, only three of the four wheeling strategies were observed when individuals wheeled at their self-selected speeds (Richter, Rodriguez et al. 2007). The 3 most common strategies at self-selected wheeling speeds were arcing (42% of the individuals), followed by SLOP (31%) and DLOP (27%). However, when wheeling up a 6° grade, 73% of individuals performed the arcing pattern (Richter, Rodriguez et al. 2007). This study did not find any statistically significant differences in handrim biomechanics (including speed, peak force relative to body weight, push angle, and push frequency) between the various wheeling strategies (Richter, Rodriguez et al. 2007).

Boninger and Colleagues found that the choice of wheeling strategy appears to be independent of wheeling speed, however differences in the biomechanical pattern of each wheeling strategy were observed (Boninger, Souza et al. 2002). When subject characteristics, including age, years with SCI, gender, height and weight were controlled, significant differences were observed in cadence and the propulsion:recovery time ratio between the different wheeling strategies. With increasing speed, there was an increase in the time spent in recovery as well as increases in force and cadence. However, the mechanical efficiency (which was defined by: tangential force^2 ÷ resultant force^2), decreased as speed increased (Boninger, Souza et al. 2002). The cadences for the semicircular and DLOP wheeling strategies were significantly lower than the SLOP and arcing wheeling strategies. The semicircular and arcing wheeling strategies had significantly greater relative time spent in propulsion than recovery compared to the DLOP and SLOP wheeling strategies. The semicircular wheeling strategy demonstrated the lowest cadence and the highest ratio of propulsion time to recovery time (Boninger, Souza et al. 2002). Boninger and colleagues (2002) advocated that the semicircular wheeling strategy may reduce over-use shoulder injuries and therefore should be taught in rehabilitation programs.

Another study found that arcing, among able-bodied (AB) subjects, is the wheeling strategy with the highest physiological efficiency (defined by: mean power output × energy expenditure^{-1} × 100) while the semicircular stroke pattern was found to have the lowest physiological efficiency irrespective of velocity in able-bodied (AB) subjects (de Groot 2004). The conflicting results of these two studies indicate that either mechanical efficiency with respect to force does not equate to physiological efficiency based on torque and energy expenditure (oxygen uptake and respiratory exchange) or that AB subjects cannot be compared to MWUs in terms of wheeling strategies. This finding may also indicate that AB and experienced MWUs utilize different kinematics while manual wheeling. Despite the observed mechanical efficiency by Boninger et al, (2002) the semicircular stroke pattern was found to have the lowest physiological efficiency in AB subjects (de Groot 2004). It was found that wheeling strategy did not explain the difference in physiological efficiency (de Groot 2004). Furthermore, no significant differences in the magnitude of pushrim force between the 4 wheeling strategies have been reported (Boninger, Souza et al. 2002). There is an accepted relationship that high pushrim forces are associated with shoulder pain and injury. Curtis and colleagues found that the highest incidents of shoulder pain were experienced during wheeling up inclines requiring much stronger forces to be exerted on the pushrim (Curtis, Roach et al. 1995). Therefore evidence for possible biomechanical correlates to shoulder pain is inconclusive and it remains unclear whether there is a ‘preferred’ wheeling strategy for minimizing injury.

Many studies have previously examined AB subjects, rather than experienced MWUs, and it is unknown if there are major differences in manual wheeling strategies between these two populations and if the data from the AB studies can be applied to MWUs. To date, the majority of research has focused on the biomechanics of propulsion rather than understanding the neural strategies underlying manual wheeling. Studies investigating the neural mechanisms underlying wheeling will further our understanding of how the nervous systems adapts to long-term wheelchair use and provide new perspectives on rehabilitation training for MWUs.

1.2.2 Muscle activation patterns

It has been shown that muscle activity patterns differ between paraplegic MWUs and AB (non-MWUs) during manual wheeling. Paraplegic MWUs have been shown to use a greater percentage of their maximal voluntary contraction in their posterior deltoids, biceps and triceps during wheeling (Dubowsky, Sisto et al. 2009). Peak muscle activity during wheeling also occurred at different positions in MWUs and AB subjects. MWUs reached peak anterior deltoid electromyographic (EMG) activity 10 degrees earlier on the pushrim while AB subjects reached peak posterior deltoid EMG 10 degrees later on the pushrim.

AB subjects also had no triceps activity in the early stages of propulsion whereas the MWUs group had triceps activity throughout (Dubowsky, Sisto et al. 2009). This study clearly indicates that MWUs and AB subjects use different wheeling strategies and that some neuromuscular adaptations, resulting in changes in coordination patterns, occur as a result of manual wheeling experience. Dubowsky et al (2009) also concluded that the greater percentage of maximum voluntary contraction achieved in the MWUs during manual wheeling may be responsible for greater joint forces and fatigue which result in shoulder joint pathology and pain. Although Dubowsky and colleagues found differences between groups, they used a very small sample size and only looked 10 wheeling cycles. Therefore, It is possible that the natural wheeling strategy was not obtained within 10 wheeling cycles. This study did not examine many of the variables commonly observed in the literature or consider the propulsion strategies that were used.

Other researchers have compared shoulder and arm muscle activation patterns between paraplegics and tetraplegics (Schantz, 1999). People with paraplegia and tetraplegia have very different functioning, including differences in trunk control as well as arm control. These differences in control make it difficult to categorize muscle activation patterns into a single group.

1.2.3 The neural control of locomotion

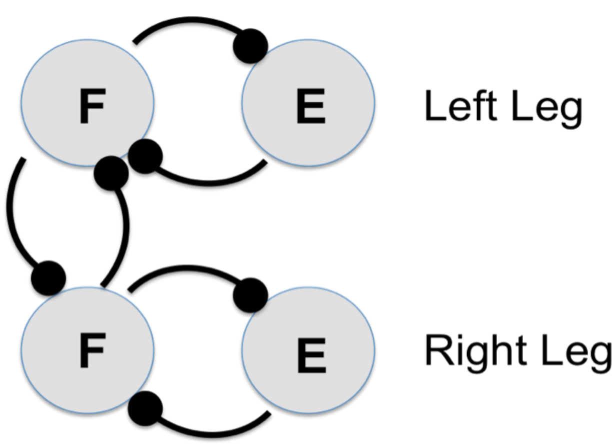
One compelling concept is that, like walking, manual wheeling can be considered an upper limb correlate of locomotion. While there has been much research on the neural control of walking, less research has been done on the neural control of wheeling. Research on the neural control of rhythmic upper limb tasks has emerged in recent years (Zehr and Chua 2000; Zehr, Carroll et al. 2004; Carroll, Zehr et al. 2005; Zehr and Hundza 2005) and some of the concepts garnered from studies of the neural control of walking have also been consistent for upper limb tasks.

Central pattern generators for locomotion

A key concept in the neural control of locomotion is the idea that neural circuits in the spinal cord, known as central pattern generators (CPGs) are capable of generating the basic features of the locomotor pattern without the influence of sensory feedback. Many forms of rhythmic human movement are thought to be generated by CPGs. Nearly all forms of human locomotion, including crawling and creeping in infants, as well as walking and swimming, are cyclical in nature. These forms of locomotion share common characteristics in that contralateral limb movements are out of phase from each other, arm and leg movements are frequency locked, and muscle activation patterns within a limb exhibit alternating contraction of flexors and extensors (Wannier, Bastiaanse et al. 2001).

Locomotion research conducted over the past century has led to the theory that a CPG or series of CPGs underlie the patterning and adaptability of locomotion. Brown (1911) proposed a schematic of a CPG consisting of oscillating neural circuitry or ‘half-centers’ based on reciprocal inhibition that resides in the lumbar spinal cord (Brown 1911). This half-center model can be used to account for the rhythmic and reciprocal relationship of the flexor and extensor motor activity. Through reciprocal inhibition, activation of the flexor half-center would cause inhibition of the extensor half-center and vice versa. Interneurons in these half-centers therefore provide the 'locomotor drive' to corresponding motoneuron pools, resulting in the alternating flexion and extension characteristic of locomotor patterns. Brown's half-center model can also account for the interlimb coordination observed during locomotion by mutual inhibition between contralateral flexor half-centers. The mutual inhibitory connection between the flexor half-centers ensures that both limbs will not undergo flexion at the same time (Figure 1.3). Rather, as one limb undergoes flexion, the flexors of the opposing limb will be inhibited, which facilitates extension in that limb.

Figure 1.3 Flexor-extensor half-center model



Schematic diagram depicting the neuronal model for a CPG. When flexor (F) neuronal pools are excited within a leg, the extensor (E) neuronal pools are inhibited and vice versa. When one leg is extended the other leg must be flexed to allow for the progression of gait. Mutual inhibition of flexor neuronal pools of contralateral limbs ensures that both legs will not initiate flexion at the same time (critical for safe bipedal walking).

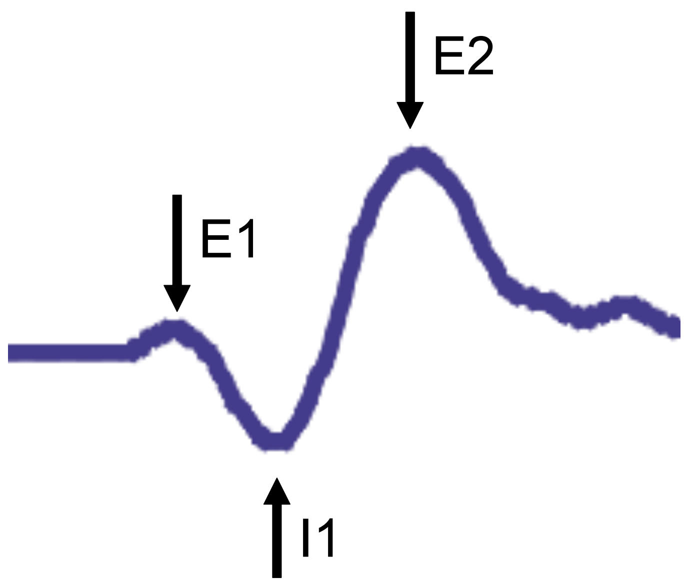
Evidence for a CPG exists for many animal models, from insects (Wilson 1964; Wolf and Pearson 1988) to the lamprey (Wallen 1997; Grillner 2006) to cats (Forssberg and Grillner 1973; Forssberg, Grillner et al. 1980; Forssberg, Grillner et al. 1980; Perret and Cabelguen 1980; Grillner and Zangger 1984). However, there is less direct evidence for a CPG underlying human locomotion. The best evidence for this comes from human infants and individuals with spinal cord injury. Human infants offer a unique perspective into the early ability of the human nervous system in producing locomotion. The neocortex and descending motor tracts are immature in infants prior to one year of age (reviewed in (Yang, Lam et al. 2004)). Infants who are unable to walk on their own are capable of stepping at various speeds and even different directions on a treadmill, as long as their body weight is partially supported (Lamb and Yang 2000). Although the infants do not have the ability to walk independently, they are capable of initiating gait-like cyclical movements given the appropriate sensory input, such as leaning them forwards, or increasing their general excitability (Yang, Stephens et al. 1998). Infants are also able to generate complex interlimb coordination patterns during stepping. When placed on two separate treadmill belts, one for each leg, the leg on the faster belt often takes 2 or more steps, for every step on the slower belt to match the speed of the belts (Yang, Lamont et al. 2005). Therefore infants appear to have two distinct and independent pattern generators for each leg. However, infants will not perform two different stepping patterns if it would result in two legs simultaneously performing swing (Yang, Lamont et al. 2005). This is consistent with Brown's half-center model (Figure 1.3) and functionally would be critical in maintaining balance and preventing falls.

Further evidence for a locomotor CPG in humans comes from observations in individuals with severe incomplete spinal cord injury (Calancie, Needham-Shropshire et al. 1994; Dimitrijevic, Gerasimenko et al. 1998). Calancie et al (1994) reported data from an individual who experienced bouts of involuntary stepping-like movements when he lay on his back with his hips extended after participating in an intense body weight supported treadmill training regime. Muscle recruitment patterns during the involuntary stepping were consistent within and between recording sessions (Calancie, Needham-Shropshire et al. 1994). The individual was unable to willfully terminate stepping, however certain postural adjustments could cease the movement. Interestingly, it was also found that the subject could influence the rhythmic stepping patterns voluntarily as long as the contraction matched the phase of the movement sequence. For example, the subject was unable to voluntarily flex his leg when the leg was undergoing extension in the movement phase, however if he tried to flex the leg undergoing flexion, a greater amount of flexion occurred. These findings indicate that the production of locomotor patterns can proceed even with little cortical descending input and may require training to strengthen neural pathways (Calancie, Needham-Shropshire et al. 1994). Another study showed that application of tonic electrical stimulation to the posterior portion of the upper lumbosacral cord in complete paraplegics elicited gait-like movement and EMG patterns (Dimitrijevic, Gerasimenko et al. 1998). This study supports the idea that the human spinal cord can generate fundamental locomotor patterns without any supraspinal influence.

Influence of sensory input during locomotion

As the previous observations from human infants (Lamb and Yang 2000; Yang, Lam et al. 2004) and the spinal cord injury case report (Calancie, Needham-Shropshire et al. 1994) demonstrate, sensory input plays an important role in shaping or adapting the locomotor output from the CPG. A large amount of research has been conducted to improve the understanding of the sensory mechanisms that influence the patterning and adaptability of locomotion. Reflexes represent motor responses to a given sensory stimuli. Investigations of changes in reflex responses to different circumstances (reflex modulation) has been frequently used to gather insight into the neural pathways underlying (Zehr, Komiyama et al. 1997; Zehr and Kido 2001; Zehr and Haridas 2003). Studying reflex responses to cutaneous stimuli is one approach to understanding the nervous system's ability to perform and adapt locomotor strategies. Modulation of the cutaneous reflex response provides evidence to substantiate the level of neural command resulting in the EMG responses observed in muscles throughout the body based on the latency of the reflex response post stimulation. The modulation of the cutaneous reflex response also allows researchers to explore the interdependency of moving and static limbs as well as the influence of phase and task dependency of rhythmic movement (Zehr, Komiyama et al. 1997; Zehr and Kido 2001; Zehr and Haridas 2003). Stimulation of a cutaneous nerve elicits a cutaneous reflex, which can be observed as a muscle response recorded using either surface or indwelling EMG. An adult cutaneous reflex response is triphasic (Figure 1.4). There is often an increase in electrical activity produced by the muscle during the first phase (early latency response), followed by an inhibition (middle latency response) and lastly by a larger increase in muscle electrical activity (late latency response) (Rowlandson and Stephens 1985).

Figure 1.4 A triphasic cutaneous reflex response



The above diagram represents a triphasic cutaneous reflex response including an early latency excitation (E1), middle latency inhibition (I1) and secondary late latency excitation (E2).

The cutaneous reflex response changes throughout development and tracking the changes of the cutaneous reflex response over the course of development has provided some insight into the neural centers mediating this reflex. At birth the reflex response is monophasic and has only one excitation phase, however by 8-12 years of age, the cutaneous reflex response develops into a triphasic (adult) response (Rowlandson and Stephens 1985). It is believed that this change in cutaneous reflex response is a result of changes in the neural circuitry (Rowlandson and Stephens 1985). Evidently, many changes in the nervous system are observed within the first two years of life (Issler & Stephens, 1983) during the critical growth period. Infants are assumed to have only the most basic level of neural control with immature supraspinal pathways. Throughout development, as these supraspinal pathways develop, accompanying changes are evident in the cutaneous reflex response. Short-latency excitatory cutaneous reflex responses produced in the long flexor and extensor muscles reduce in size over the first few years of life (Issler & Stephens, 1983). Over the next few years the reflex response is replaced by the 'adult' pattern response with an additional inhibitory and long-latency excitation phase (Issler and Stephens 1983). Lesions to the motor cortex have been found to result in the loss of the middle-latency inhibitory component of the cutaneous reflex response as well as the loss of the secondary (late-latency) excitation phase while the short-latency (early-excitation) phase becomes exaggerated (Jenner and Stephens 1982). The similarities between infant cutaneous reflex responses and reflexes of people with lesions to the cortex are quite similar.

Further evidence that the long-latency component of the cutaneous reflex response involves a trans-cortical loop comes from studies of individuals with Klippel-Feil syndrome. This syndrome is characterized by the fusion of the cervical and cervico-thoracic vertebrae, and is often associated with bilateral ‘mirror’ movements. It has been found that when a single hemisphere of the brain is stimulated, patients with Klippel-Feil syndrome exhibit an early latency excitation occurring at approximately 27ms and an early latency inhibition occurring at approximately 44ms (on the side of stimulation) and bilateral long-latency components of the cutaneous reflex response including middle excitation occurring at approximately 53ms and late excitation occurring at approximately 83ms (Farmer, Ingram et al. 1990). The cutaneous reflex response was only observed on the side of stimulation in healthy individuals (Farmer, Ingram et al. 1990). This further supports the concept that the long-latency phase of the cutaneous reflex response occurring at approximately 53ms and 83ms of the cutaneous reflex response occurs as a result of interneurons affected by descending regulation (Farmer, Ingram et al. 1990).

During locomotion, cutaneous reflex responses are important for ensuring appropriate responses to perturbations in the external environment, such as an unexpected perturbation that could elicit tripping. Indeed, perturbations that stimulate cutaneous receptors on the foot will elicit the well-described tripping or stumbling corrective response. When the foot encounters an obstacle in its path during the swing phase, the reflex response consists of increased leg flexion that would effectively raise the foot to avoid the obstacle. This response, which has been well described in cats (Forssberg 1979) as well as humans, (Schillings, Van Wezel et al. 1996), has a strong cutaneous contribution.

It is very difficult, although possible, to provide unexpected mechanical perturbations during gait to elicit the stumble response (Forssberg 1979;

Forssberg 1979; Schillings, Van Wezel et al. 1996). Producing unseen obstacles requires that modifications be made to obstruct the obstacle from the subject's line of vision and therefore may create an unrealistic environment. As a result of the difficulty in using mechanical perturbations to elicit reliable and consistent cutaneous reflex responses, researchers frequently use cutaneous nerve stimulation.

Electrical cutaneous nerve stimulation is not normally experienced, however it has been shown to produce analogous reactions to mechanical cutaneous stimulation. The same qualitative responses were observed when cats were either electrically or mechanically stimulated on the dorsum of the paw (Forssberg 1979). Analogous responses are observed if the nerve stimulated is functionally similar to the cutaneous stimulus. This infers that similar pathways may be activated by electrical nerve stimulation that are also activated during mechanical stimulation (Forssberg 1979). This allows researchers to use electrical stimulation to simulate mechanical perturbations found in nature. To simulate mechanical perturbations to the anterior aspect of the foot, the superficial peroneal nerve can be targeted, while stimulating the sural nerve is analogous to mechanical perturbations to the posterior aspect (sole) of the foot.

Studies employing either mechanical or electrical stimuli to elicit cutaneous reflex responses have demonstrated that these reflex responses are flexible and may be modulated according to the phase of the locomotor cycle in which they are presented (phase-dependency), the movement task during which they are presented (task-specificity), and the location of the stimuli (site-specificity).

Phase-dependent reflex responses are exhibited when cutaneous stimuli delivered during different phases of the locomotor cycle trigger different responses. For example, an excitation of the extensors occurs when the hindlimbs are stimulated by a touch to the paw dorsum during stance, while an excitation of the flexors occurs when the same stimulus is delivered during the swing phase in chronic spinal cats (Forssberg, Grillner et al. 1975). Similarly, perturbations to the dorsum of the foot experienced during the swing phase of infant stepping resulted in increased flexion of the knee joint, however, when the perturbation occurred during stance phase, infants tended to prolong stance phase duration (Lam, Wolstenholme et al. 2003; Pang, Lam et al. 2003). In adults, it was also found that the responses to electrical stimulation of the superficial peroneal nerve and the tibial nerve were larger during the swing phase and transition phase compared to the stance phase (Zehr, Komiyama et al. 1997). These findings indicate that the efficacy of a given reflex pathway can change (modulate) according to the phase of rhythmic movement and that such sensory gating could be mediated by central pattern generating circuits.

Task-specificity of the cutaneous reflex response has been demonstrated by examining forward and backward walking in the cat (Buford and Smith 1993). It was observed that forward and backward walking elicited differences in stumbling corrective reactions when the dorsal aspect of the paw was stimulated during forward walking and the ventral aspect of the paw was stimulated during backward walking. However, when stimulation was applied in the same location during both forward and backward walking, responses were different (Buford and Smith 1993). This indicates that the response to the stimulation is dependent on the direction of walking. Therefore forward and backward walking can be considered to be different tasks.

The reflex response is also dependent on the location of stimulation. A perturbation to the dorsal aspect of the foot during forward walking will elicit increased flexion while a perturbation to the lateral surface of the leading foot during sideways walking will also elicit flexion (Lam, Wolstenholme et al. 2003). A greater response (ie flexion) is observed when the perturbation occurs in the direct line of movement whereas a perturbation to the side of the foot during forward walking does not have any effect on the movement (Lam, Wolstenholme et al. 2003). In adults, it has been demonstrated that stimulation of the sural, posterior tibial, and superficial peroneal nerve elicits distinct reflex responses during walking (Van Wezel, Ottenhoff et al. 1997). This further supports that location of stimulation on the foot influences the cutaneous reflex response. Large ipsilateral biceps femoris responses were observed after sural nerve stimulation during stance phase, however much smaller responses were observed during the same phase when the peroneal nerve was stimulated (Van Wezel, Ottenhoff et al. 1997). The biceps femoris response during swing was very similar between stimulation of the same 2 nerves (Van Wezel, Ottenhoff et al. 1997). There did not appear to be any consistent relationship in cutaneous reflex response in the biceps femoris between sural nerve stimulation and peroneal nerve stimulation. These findings indicate that there is separate control over the reflex pathways of each of the three nerves. During locomotion, the reflex modulation was not related to the background EMG activity. However during static conditions, there was a strong relationship between the reflex amplitude and background EMG, therefore it has been suggested that these differences occur at a pre-motoneuronal level (Van Wezel, Ottenhoff et al. 1997).

Neural control of upper limb rhythmic movements

The majority of research on the neural control of human rhythmic movement has been conducted in the context of walking. However, recent research has explored other forms of rhythmic motion, such as arm and leg cycling (Zehr, Klimstra et al. 2007). Recently the relationship between various rhythmic human movements was analyzed. Leg and arm cycling were compared to walking and inclined stepping. Factor analysis demonstrated that four principal components accounted for 93% of the variance in background EMG and middle-latency cutaneous reflex amplitude between the three tasks (Zehr, Klimstra et al. 2007). These results are consistent with the concept of a common neural pattern generator for rhythmic movement and help to justify the use of rhythmic upper limb movements as a means of accessing human CPG networks. Similar to what has been observed during walking, the processing of sensory input from cutaneous receptors in the upper limbs during rhythmic movement also exhibits phase dependent and task-dependent modulation.

Phase-dependent cutaneous reflex modulation

Upper arm cycling involves using the hands to move handgrips of an arm ergometer in a cyclical motion and is performed in a seated position with the ergometer secured to a table. The handgrips are fixed in length to ensure a constant cyclical path. Upper limb cycling is frequently used because it is a simple movement exhibiting phase-dependent characteristics. Not only does the movement-related background EMG influence cutaneous reflex modulation, but there are also some muscles that show evidence of phase dependent cutaneous reflex modulation (Zehr and Chua 2000; Zehr and Kido 2001). This indicates that differences in cutaneous reflex modulation may be dependent on the functional state of the arm during stimulation. Different muscles will demonstrate different changes in cutaneous reflexes as a result of phase dependency rather than the muscle activity produced by the movement (Zehr and Chua 2000). Similar to Zehr and colleagues earlier study (Zehr, Komiyama et al. 1997) which examined the effect of stimulating different nerves in the leg, this study examined 10 muscles with stimulation to the superficial radial nerve, median nerve and ulnar nerve, it was also discovered that over half of the cutaneous reflex responses were not significantly dependent on background EMG (Zehr and Kido 2001). It was found that early latency reflexes for the ipsilateral and contralateral anterior deltoids, ipsilateral posterior deltoids, ipsilateral triceps, ipsilateral extensor carpi radialis and first dorsal interosseous, were significantly modulated according to position after stimulation of each of the three nerves (Zehr and Kido 2001). Significant modulation was also observed in the ipsilateral and contralateral anterior deltoids, the ipsilateral triceps and the ipsilateral first dorsal interosseous during middle latency (Zehr and Kido 2001).

Task-specific cutaneous reflex modulation

Cutaneous reflex modulation is not only dependent on the phase of a movement, but also the type or form of movement. For example, differences in cutaneous reflex modulation have been observed between active vs. passive cycling. Passive bilateral arm cycling elicits small cutaneous reflexes in response to superficial radial nerve stimulation and only showed significant modulation of reflex amplitude across the cycle in the arm receiving the stimulation in 8% of the 40 cases studied (Carroll, Zehr et al. 2005). The results for passive cycling had the same incidence of statistically significant modulation as the stationary limbs (Carroll, Zehr et al. 2005). This indicates that afferent information received during passive cycling is not enough to elicit the modulation of cutaneous reflexes.

Significant task dependent modulation is also evident when comparing static contraction versus rhythmic arm cycling. This was directly compared by matching reflex responses to the 3, 6, 9 and 12 o'clock positions for static contractions and dynamic cycling. Task dependency was defined as both changes in reflex amplitude as well as reversals in sign of the reflex amplitude. It has been shown that when either the superficial radial, median or ulnar nerve is stimulated, cutaneous reflex responses are generally larger in amplitude during the cycling task compared to the static task (Zehr and Kido 2001).

In contrast to the differences between active cycling, passive cycling, and static contractions, there is little difference in modulation of cutaneous reflex responses when comparing out-of-phase (asymmetrical) vs. in-phase (symmetrical) cycling in the six muscles studied (Carroll, Zehr et al. 2005). The amplitudes of cutaneous reflexes in response to stimulation of the superficial radial nerve recorded while bilaterally cycling 180 degrees out-of-phase (asymmetrical) are largely similar to those elicited while bilaterally cycling 180 degrees in-phase (symmetrical) (Carroll, Zehr et al. 2005). Some early-latency EMG responses, on the ipsilateral side of the stimulus were significantly larger for the out-of-phase (asymmetrical) cycling compared to in-phase (symmetrical) cycling for each arm position. These differences were observed in the posterior deltoids (shoulder extensor) and the extensor carpi radialis (wrist extensor). In contrast, the ipsilateral anterior deltoids (shoulder flexor) had significantly smaller early- and middle-latency cutaneous reflex responses during the out-of-phase (asymmetrical) condition, compared to the in-phase (symmetrical) condition (Carroll, Zehr et al. 2005). This study has demonstrated that the synchronous in-phase (symmetrical) motion does not produce different responses to stimulation of the superficial radial nerve, in most muscles, compared to the alternating out of phase (asymmetrical) motion. Considering the fact that there was not a large difference in modulation of the cutaneous reflex response between the in-phase (symmetrical) and out of phase (asymmetrical) movements, it is possible that these two movements have similar neural control or that there is little interaction across the spinal cord in the production of these movements.

The movement of one arm does not appear to influence the modulation of cutaneous reflex responses in the contralateral limb (Carroll, Zehr et al. 2005). To test the influence of the movement of one arm on the stationary limb, two conditions of unilateral arm cycling were compared to bilateral cycling. In the first unilateral cycling condition, the arm receiving cutaneous nerve stimulation was cycling while the other remained stationary. In the other condition, the arm receiving cutaneous nerve stimulation maintained a tonic contraction while the other arm was cycling. The amplitude of the cutaneous reflex responses in the arm on the contralateral side of the stimulus during the bilateral task (both arms cycling) was similar at each arm position to the single-arm task (one arm cycling) in which the arm stimulated was contralateral to the arm moving. Therefore the only difference between these trials was that in the bilateral cycling task both arms cycled while in the unilateral task, the stationary arm was stimulated while the contralateral arm cycled. The only significant difference in reflex amplitude between these two conditions occurred in the anterior deltoid, contralateral to the stimulus, during middle latency (Carroll, Zehr et al. 2005).

This same trend was present when the bilateral condition was compared to single-arm cycling ipsilateral to stimulation. The reflex amplitude of the posterior deltoids was significantly smaller compared to the bilateral task during early latency while the reflex amplitude of the anterior deltoid and triceps were larger compared to the bilateral task during early/middle and early latencies respectively (Carroll, Zehr et al. 2005). These differences between unilateral and bilateral cycling conditions indicate that the neural control differs between these two tasks.

1.3 Summary and thesis objectives

In summary, the modulation of the cutaneous reflex response during upper limb cycling is influenced by the phase of the cycling movement as well as the type of cycling task, and the specific nerve being stimulated. However, there appear to be weak interlimb connections since reflex responses in an arm are primarily influenced by the ipsilateral (stimulated), moving arm, and less so by the movements of the contralateral side. This may allow for the complexity of humans' bimanual dexterity. If there was a strong inter-arm connection and the functional state of one arm was highly dependent on the other, humans would not have the ability to perform such diverse motions. Arm cycling could be considered a novel rhythmic movement because it is not a form of locomotion people usually use. However, there are other more common circumstances where rhythmic movements of the arm are used for locomotion, such as people who require a manual wheelchair. There are already indications from a recent study for differences in the neuromuscular regulation of muscle activity patterns between experienced MWUs and able-bodied (naïve MWUs) subjects (Dubowsky, Sisto et al. 2009). Could it be possible that wheeling strategies in long-time manual wheelchair-users harness CPG circuits? One approach to investigate this possibility is to examine the modulation of the cutaneous reflex response during manual wheeling.

Although many manual wheeling studies use able-bodied subjects, it is unknown if the wheeling strategy and biomechanics differ between these two groups. If there are major differences in kinetics and kinematics between the two groups, that could influence differences in muscle activation patterns as well as cutaneous reflexes during manual wheeling. Thus the first objective of this thesis is to determine the biomechanical differences between AB and experienced MWUs to validate the use of AB subjects in manual wheeling studies (Chapter 2). The second objective of this thesis is to determine whether cutaneous reflex responses during manual wheeling follow similar patterns of phase-dependent and task-specific modulation as observed in previous arm cycling studies. In addition, the question of whether manual wheeling experience affords any difference in the modulation pattern of these reflex responses will also be explored (Chapter 3).

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2 THE BIOMECHANICS OF MANUAL WHEELING IN ABLE-BODIED SUBJECTS AND EXPERIENCED MANUAL WHEELCHAIR USERS[1]

2.1 Introduction

Currently in Canada there are 41,000 individuals with spinal cord injuries (SCI) and 155,000 individuals living in private homes who depend on manual wheelchairs for locomotion (Shields 2004). Many individuals with mobility impairments, including people with SCI, are able to regain functional independence through the use of a wheelchair (Chaves, Boninger et al. 2004). However, the incidence of shoulder joint injuries is high among manual wheelchair users (MWUs), potentially debilitating their ability to move independently and participate in activities of daily living (Silfverskiold and Waters 1991). More than two-thirds of spinal cord injured MWUs experience shoulder pain throughout their lives (Curtis, Drysdale et al. 1999) which are often associated with repetitive upper limb movements and large forces exerted by the shoulder joint.

Many studies have investigated the biomechanical characteristics of manual wheeling (Sanderson and Sommer 1985; Robertson, Boninger et al. 1996; Shimada, Robertson et al. 1998; Boninger, Souza et al. 2002; Koontz, Cooper et al. 2002; Richter, Rodriguez et al. 2007; Collinger, Boninger et al. 2008; Rice 2008), however only a few have compared able-bodied (AB) subjects

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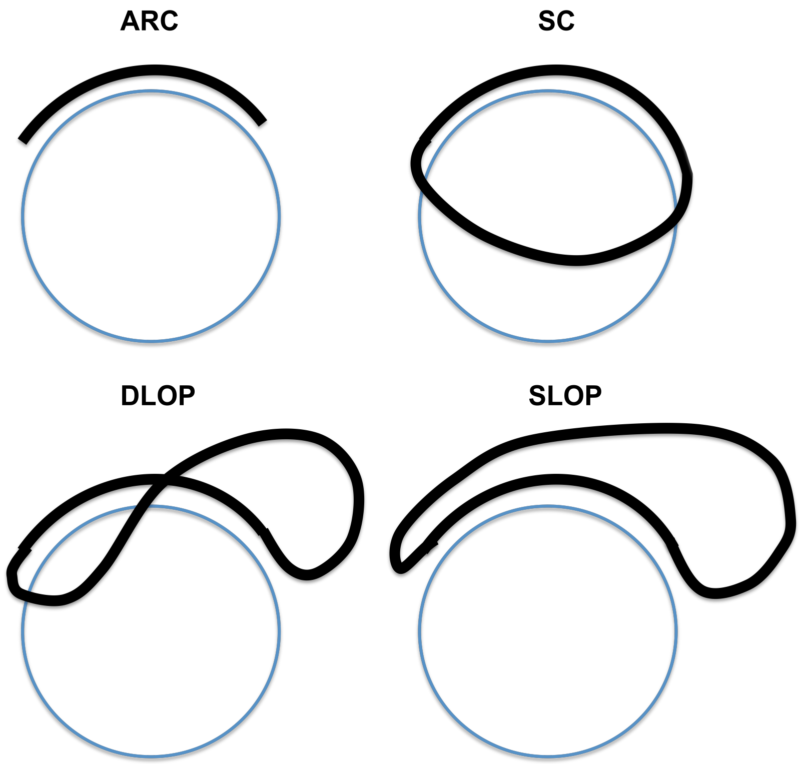
to experienced MWUs (Robertson, Boninger et al. 1996; Dubowsky, Sisto et al. 2009). Much of the manual wheeling literature examined the biomechanics of AB subjects and assumed that the data can be applied to manual wheeling populations (Veeger, van der Woude et al. 1989; de Groot 2004). Rarely are all necessary elements studied, including kinetics, kinematics, and electromyography (EMG), and consequently important differences between these two groups could easily be missed.

Previous studies have shown that AB subjects demonstrate different propulsion strategies from experienced MWUs (Veeger, van der Woude et al. 1989; Chou 1991; Shimada, Robertson et al. 1998), however, it is unknown if biomechanical characteristics change with manual wheeling experience and one's dependence on manual wheeling for locomotion. The current literature has defined four wheeling strategies (Figure 2.1) based on the path of the hands during the recovery phase of the wheeling cycle. These include *semi-circular* (SC), where the hands fall below the propulsion pattern during the recovery period) *single looping over propulsion* (SLOP) and *double looping over propulsion* (DLOP) (Shimada et al, 1998). The SLOP is defined by the hands rising above the path of the pushrim (Shimada, Robertson et al. 1998), while the DLOP consists of the hands initially rising above the pushrim and then dropping below the pushrim (Richter, Rodriguez et al. 2007). More recently, a fourth wheeling strategy, *arcing* (ARC), was identified (Boninger, Souza et al. 2002). The arcing pattern describes a recovery pattern that follows the reverse of the propulsion phase along the pushrim (Boninger, Souza et al. 2002).

Figure 2.1 Manual wheeling strategies



The four commonly classified wheeling strategies: arcing (ARC), semi-circular (SC), double looping over propulsion (DLOP), and single looping over propulsion (SLOP).

One earlier study comparing wheeling strategies in AB vs. experienced MWUs observed that all AB subjects used ARC, while the experienced MWUs used SC (Chou 1991). More current literature has debated which propulsion strategy is most efficient and should be taught by rehabilitation therapists (Boninger, Souza et al. 2002; de Groot 2004). The SC strategy has been found to have the lowest cadence and highest ratio of propulsion time to recovery time in MWUs (Boninger, Souza et al. 2002). These biomechanical characteristics of the SC strategy were considered to impart greater mechanical efficiency to this wheeling strategy, compared to the other wheeling strategies. Thus, the SC strategy was considered the preferred strategy in terms of reducing the incidence of shoulder injury (Boninger, Souza et al. 2002). However, another study examining physiological efficiency (which was defined by the mean power output $\times$ energy expenditure^{-1} $\times$ 100), found the SC strategy to be the least efficient and the ARC strategy to be the most efficient among AB subjects (de Groot 2004). The opposing conclusions from these studies suggest that either the different methods used to determine efficiency are not comparable or that the manual wheeling strategies of AB subjects cannot be compared to those of experienced MWUs. As mentioned earlier, experienced manual wheelers tend to use different wheeling strategies compared to AB subjects. Therefore, it would be expected that experienced MWUs also exhibit different levels of efficiency based on differences in wheeling strategy.

Previously, no significant differences in force were observed between the four wheeling strategies among manual wheelchair users (Boninger, Souza et al. 2002). Another important indicator of biomechanical effectiveness, during manual wheeling, examines the ratio between the tangential forward force to the overall force applied to the pushrim. This method of calculating mechanical efficiency looks at the ratio of the tangential force^2/ resultant force^2 to determine what fraction of the total force is responsible for propelling the chair forward. It has been assumed that AB individuals are less efficient manual wheelers than experienced MWUs, however this has only been a qualitative observation (Robertson, Boninger et al. 1996; Dubowsky, Sisto et al. 2009).

Despite the vast biomechanics literature, few studies have looked at muscle activation patterns as well as the biomechanical characteristics of manual wheeling (Veeger, van der Woude et al. 1989; Schantz, Bjorkman et al. 1999; Dubowsky, Sisto et al. 2009). Muscle activation patterns may differ during the recovery phase of the wheeling cycle with different propulsion strategies because the arm is performing slightly different movements. By furthering the understanding of shoulder muscle activation and identifying normative data, inefficient wheeling strategies or problems such as muscle imbalances may be recognized.

More research needs to be conducted to understand the muscle activation patterns and biomechanical differences during manual wheeling between AB and experienced MWUs before findings from studies using AB subjects are applied to individuals with SCI. Therefore the purpose of this study was to determine if the propulsion strategies, kinematics, kinetics, and muscle activation patterns during manual wheeling, are consistent between AB individuals with no manual wheeling experience and individuals who have been dependent on manual wheelchairs for at least one year.

2.1.1 Hypothesis

It is hypothesized that individuals who are experienced and depend on manual wheelchairs will exhibit different biomechanical characteristics compared to the able-bodied group. Specifically, it is hypothesized that:

1. Manual wheelchair users will exhibit smaller average forces applied to the pushrim compared to the able-bodied group.

2. Manual wheelchair users will demonstrate greater joint range of motion compared to the able-bodied group.

3. Manual wheelchair users will use different wheeling strategies and therefore display different muscle activation patterns compared to the able-bodied group.

2.2 Methods

2.2.1 Participants

A total of 22 individuals (13 able-bodied inexperienced MWUs and 9 experienced MWUs) were recruited to participate in this study. The inclusion criteria for the manual wheeling group included using a manual wheelchair on a regular basis for more than one year and having a spinal cord injury at T1 or below (paraplegia). All participants were over 19 years of age, capable of wheeling for 5-10 minute spans, and provided informed consent. People were excluded from the study if they had any injury to their upper-limbs or if they had upper-limb or nerve pain that would prohibit them from wheeling a manual wheelchair. All experimental procedures were approved by the UBC Behavioural Research Ethics Board. All participants provided informed and written consent to participate in this study

2.2.2 Data collection

Demographic information including age, height, weight, sex, injury level, age of injury and years since injury, was collected from all subjects. All participants with SCI were given the Wheelchair Users Shoulder Pain Index (WUSPI) (Curtis, Roach et al. 1995) to determine if they experienced shoulder pain related to manual wheelchair use.

The SmartWheel (Three Rivers Holdings, Mesa, Az), which is a specialized wheel that measures 3-dimensional forces, was exchanged with the right wheel of the MWUs wheelchair or secured to the laboratory wheelchair that was used by the AB subjects. Tire air pressure was pumped to the recommended level. The participant's weight while in the wheelchair was measured. MWUs used their own manual wheelchair while AB participants used a fitted ELEVATION™ wheelchair (Instinct Mobility, Vancouver, BC, CA). Participants' wheelchairs were secured to the rollers with adjustable straps (Figure 2.2). All participants were given adequate time to become familiar with the rollers and find their self-selected wheeling cadence. To ensure that the cadence did not change throughout the course of the study, the subjects were asked to maintain their frequency by synchronizing with a metronome. All subjects wheeled for approximately 8 minutes and were provided with breaks as necessary to reduce fatigue.

Figure 2.2 Experimental set-up: manual wheeling

This photograph represents the experimental set up. The wheelchair is safely secured to the rollers and the SmartWheel replaces the right wheel. The electrodes and Optotrak sensors are placed on the right arm.

Kinetics

Data from the SmartWheel was collected at 240Hz using proprietary software (Three Rivers Holdings, Mesa, AZ, USA). Participants began wheeling prior to initiating data collection so that the recorded data would be representative of continuous wheeling rather than initial acceleration to begin propulsion. The following variables were collected from the SmartWheel: the 3-dimensional forces applied to the pushrim, cadence, velocity, push angle (the portion of the handrim in which the hand was in contact, in degrees).

Kinematics

A motion analysis system, Optotrak 3020 (NDI, Waterloo, ON, CA) was used to record sagittal-plane upper limb kinematics on the right side of the body during wheeling. Two cameras positioned to create a 90° angle between them were used to collect the data. Active infrared markers were placed on the following landmarks: 3rd metacarpophalangeal joint, radial styloid, lateral epicondyle, and acromion to represent the hand, wrist, elbow, and shoulder, respectively. The marker on the 3rd metacarpophalangeal joint was used to define the wheeling strategy for each subject (Richter, Rodriguez et al. 2007). Kinematic data was collected at a sampling frequency of 200 Hz. All data was filtered using a fourth order, 7 Hz low-pass Butterworth filter.

The shoulder joint angle was calculated using the upper segment of the arm (lateral epicondyle to acromion) and the vertical (projecting down from the acromion) while the elbow joint was calculated using the lower segment of the arm (lateral epicondyle to radial styloid) and upper segment of the arm (lateral epicondyle to acromion). Lastly the wrist angle was calculated using the lower segment of the arm (lateral epicondyle to radial styloid) and the hand (lateral styloid to 3rd metacarpophalangeal joint).

Electromyography (EMG)

The skin above the muscles of interest was shaved, exfoliated and wiped with an alcohol pad to help remove dirt and oils that may interfere with the EMG signal. EMG activity was recorded from the anterior (AD) and posterior deltoids (PD), biceps brachii (BB), triceps brachii (TB) (long head), flexor carpi radialis

(FCR) and extensor carpi radialis (ECR) (brevis) with ground electrodes placed over the olecranon process and clavicle. These muscles (AD, PD, BB, TB) were selected because of their role in flexion and extension of the shoulder and elbow joints during wheeling (Dubowsky, Sisto et al. 2009). EMG data from the AD and PD, BB and TB were collected with a 4-lead Grass amplifier (7HP511, Grass Technologies, Warwick RI) while the FCR and ECR were collected from a Life Electronics 2 lead system. EMG signals were collected at 1000Hz, pre-amplified (×2000-5000), band-pass filtered between 100 and 300Hz. Further offline analysis was done with a custom made program using MATLAB (Mathworks, Natwick, MA, USA). Data were filtered using a fourth order, 100 Hz low pass Butterworth filter and full-wave rectified.

2.2.3 Data analysis

Kinetic data from the SmartWheel were broken into cycles with the start of each cycle defined by initial hand-contact on the pushrim. All cycles were normalized in time to 100% of the cycle and averaged together to determine a representative force profile for each individual. Several variables were obtained from the SmartWheel including the 3-dimensional forces applied to the pushrim, cadence, velocity, push angle (the portion of the handrim in which the hand was in contact, in degrees), and percentage of cycle spent in propulsion. From the SmartWheel data, the resultant and tangential forces were calculated. The average forces were obtained by calculating the mean over the course of the full wheeling cycle. Lastly mechanical efficiency was calculated using the following equation: $F_{tangential}^2/F_{resultant}^2$ (Shimada, Robertson et al. 1998). Average mechanical efficiency was calculated across the entire wheeling cycle.

Kinematic and EMG data were synchronized by an external trigger pulse signal and divided into cycles based on the maximum shoulder extension. All data were normalized in time to 100% of the cycle. For each subject, the EMG data for each muscle was normalized to the peak rectified EMG value of that muscle during the average wheeling cycle. EMG activity during wheeling was then averaged across subjects in the able-bodied and experienced MWUs groups. The time to peak muscle activity (with respect to the onset of each cycle) was calculated and compared between groups. Shoulder, elbow and wrist angles were calculated and averaged for both groups. Kinematic data were quantified by the range of motion (ROM) of each joint across the wheeling cycle as well as peak flexion and extension of the shoulder and elbow and peak ulnar and radial deviation at the wrist. Hand trajectory was plotted for all cycles to ensure that one wheeling strategy was used for the majority of the data collection. The averaged hand trajectory was then used to categorize the individual's wheeling strategy into one of four groups. One investigator categorized the wheeling strategies according to the definitions mentioned earlier (Shimada, Robertson et al. 1998; Boninger, Souza et al. 2002; Richter, Rodriguez et al. 2007). An additional researcher also categorized the wheeling strategies according to the same definitions. This was done to ensure that all wheeling strategies were properly categorized.

2.2.4 Statistical analysis

A Chi-squared test of independence was used to determine if wheeling strategy was dependent on manual wheeling experience. Means and standard deviations (SD) were calculated for all kinetic, kinematics, and EMG variables. Two-tailed non-parametric tests (Mann-Whitney U) were used to determine if any of the kinetic, kinematics or EMG variables were different between the experienced MWUs and the able-bodied groups. Using averaged data from each subject, Spearman's rank correlation coefficients (r_s) were calculated and tested for significance between percent time spent in propulsion and push angle as well as percent time spent in propulsion and wrist ROM. Lastly, the Spearman's rank correlation was performed for each group on force (tangential and resultant) and bodyweight. All statistical computations were completed with SPSS (SPSS Inc. Chicago, Il, USA). Statistical significance was evaluated at an alpha level of 0.05.

2.3 Results

2.3.1 Subjects

Nine experienced manual wheelers (5 males, 4 females) with spinal cord injuries at or below T1, and 13 able bodied subjects (4 males, 9 females) volunteered to participate in the study. The average age of the experienced MWU's and the able-bodied subjects was 44.7±11.9 and 26.1±1.45, respectively. The MWUs WUSPI scores were quite low indicating that shoulder pain was very minimal and experienced rather infrequently. The specific characteristics of the MWU group and AB group are summarized in Table 2.1 and Table 2.2. The average self-selected wheeling velocities for subjects in the AB and MWUs group were 0.75 (0.10) m/s and 0.82 (0.32) m/s respectively (Appendix 1).

Table 2.1 Manual wheelchair users participant data

Subject	Sex	Age (yr)	Injury level	years since injury	Dominant hand	Propulsion strategy	Weight (kg)	Height (cm)
1	M	36	T3-T6	18	R	SC	88.5	188.0
2	M	48	T4	7	L	ARC	79.4	188.0
3	M	39	T4	20	R	SC	68.0	182.9
4	F	34	T5	29	R	SC	47.6	163.0
5	M	27	T5	6	R	SC	59.0	177.8
6	F	46	T10-11	27	R	ARC	40.9	162.6
7	F	54	T7	19	R	SC	65.3	160.0
8	F	52	T1	14	R	DLOP	72.6	165.1
9	M	66	T2	2	R	SLOP	90.7	190.5

Table 2.2 Able-bodied participant data

Subject	Sex	Age (yr)	Height (cm)	Weight (kg)	Dominant hand	Propulsion strategy
1	F	26	175.3	63.5	R	SLOP
2	F	28	172.7	61.2	R	ARC
3	M	25	179.1	77.6	R	ARC
4	F	28	172.7	65.8	R	SLOP
5	F	24	160.0	65.8	R	SLOP
6	M	28	183.0	69.3	R	ARC
7	F	25	154.9	50.8	R	SLOP
8	M	27	180.0	62.0	R	SLOP
9	F	25	157.5	56.7	R	ARC
10	F	24	175.3	64.9	R	ARC
11	M	25	180.3	72.6	R	ARC
12	F	26	170.0	54.0	R	ARC
13	F	24	167.6	70.3	R	ARC

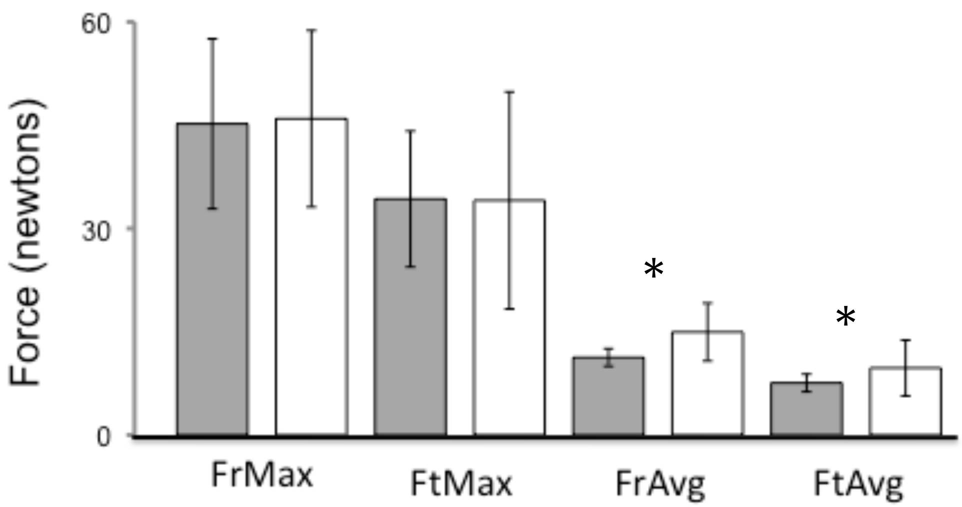
2.3.2 Propulsion strategies

All four of the commonly classified propulsion strategies were observed in this study. In the AB group, only ARC (61.5%) and SLOP (38.5%) were observed while in the experienced MWUs group, all strategies were utilized including ARC (22%), DLOP (11%), SLOP (11%), and SC (55%). According to the chi-squared test of independence, wheeling strategy was found to be dependent on subject group ($p<0.01$).

2.3.3 Kinetics

The only significant differences in forces between the AB and MWUs group were in average resultant force and average tangential force (Figure 2.3). Results are presented in means±standard deviation (Appendix 1). MWUs (9.7±4.1) demonstrated significantly larger average tangential forces compared to the AB group (7.6±1.3). MWUs also (14.9±4.2) demonstrated significantly larger average resultant forces compared to the AB group (11.3±1.3). Significant differences were also found in percentage of the wheeling cycle spent in propulsion and push angle between the two groups (Figure 2.4) ($p<0.05$). MWUs (50.5±9.9) demonstrated significantly larger percentage of the wheeling cycle spent in propulsion compared to the AB group (42.0±6.1). MWUs also (78.4±15.3) demonstrated significantly larger push angles compared to the AB group (56.8±8.8). The percent time spent in propulsion was significantly correlated with push angle, when collapsed across groups ($r_s = 0.57$, $p<0.05$). Percent time spent in propulsion was also significantly correlated with wrist ROM, when collapsed across groups ($r_s = 0.65$, $p<0.05$). When the forces demonstrating significant differences between groups (force resultant average and force tangential average) were correlated with body weight for each group, only the average force tangential in the MWUs group was found to be significantly correlated with bodyweight ($r_s = 0.71$, $p<0.05$).

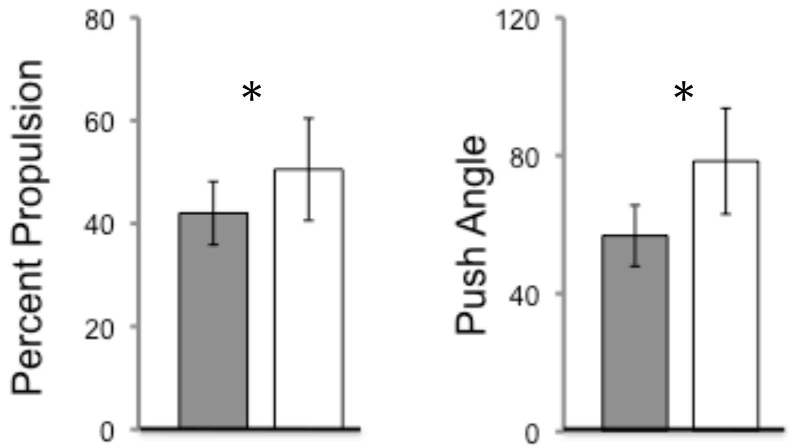
Figure 2.3 Maximum and average resultant and tangential forces in able-bodied subjects and manual wheelchair users



Maximum resultant force (FrMax) and tangential force (FtMax) as well as average resultant force (FrAvg) and tangential force (FtAvg) for both the experienced MWUs (white) and AB (grey) groups. The bars represent group averages and the error bars represent 1 standard deviation. The MWUs demonstrated significantly greater FrAvg and FtAvg compared to the AB group (represented by the asterisk).

Means and standard deviations for all kinetic variables are presented in Table 2.3 according to group and wheeling strategy.

Figure 2.4 Percent propulsion and push angle in able-bodied subjects and experienced manual wheelchair users



The bar graph on the left presents the average percent time spent in propulsion and the bar graph on the right represents the average push angle for the MWU (white) and AB (grey) groups. The error bars represent 1 standard deviation. The MWUs demonstrated significantly greater percent propulsion and push angle compared to the AB group (represented by the asterisk).

2.3.4 Kinematics

Significant differences were found in wrist ROM, peak wrist radial deviation and peak elbow flexion between AB and MWU groups ($p<0.05$).

Figure 2.5 illustrates the average kinematic profile of the kinematics grouped according to wheeling strategy and subject group. Due to the low numbers, only descriptive statistics (means and standard deviations) were used to compare data across the different wheeling strategies (Table 2.4). As observed in Figure 2.5, the time to peak radial deviation of the wrist joint appeared to be longer in the experienced MWUs who used the arcing strategy, however, the AB group did not show this same pattern. Interestingly the time to peak elbow extension and shoulder flexion also appeared longer among the experienced MWUs who used the ARC and SLOP strategy and again the AB group did not show this same pattern (Figure 2.5 and 2.6).

Table 2.3 Kinetic variables according to wheeling strategy

	ARCING		**DLOP**		**SEMI-CIRCULAR**		**SLOP**	
Variable	AB	SCI	AB	SCI	AB	SCI	AB	SCI
Avg F_{tan} (N)	7.5 (1.2)	9.7 (2.9)		1.5		10.4 (1.3)	7.6 (1.6)	15.5
Max F_{tan} (N)	33.7 (11.4)	31.1 (13.4)		5.5		39.2 (11.9)	35.4 (7.2)	48.1
Avg F_{res} (N)	10.8 (1.3)	16.0 (4.0)		6.2		16.4 (2.5)	12.1 (1.2)	15.6
Max F_{res} (N)	43.2 (13.8)	47.1 (17.4)		27.2		50.5 (11.7)	49.1 (9.0)	44.2
Mechanical Efficiency	0.4 (0.1)	0.5 (0.4)		0.5		0.3 (0.2)	0.4 (0.2)	0.6
Cadence	0.8 (0.0)	0.8 (0.1)		0.7		0.8 (0.0)	0.7 (0.0)	0.7
push/rec	0.8 (0.2)	1.3 (0.3)		0.5		1.2 (0.5)	0.7 (0.2)	1.0
% Prop	43.5 (5.6)	55.5 (6.4)		32.0		52.8 (8.9)	39.0 (7.0)	50.0
Velocity (m/s)	0.7 (0.1)	0.6 (0.0)		0.6		1.0 (0.4)	0.9 (0.1)	0.6
Push angle (deg)	52.6 (7.3)	64.0 (3.2)		-		87.3 (14.5)	65.2 (4.3)	71.7
	n=8	n=2	n=0	n=1	n=0	n=4	n=4	n=1

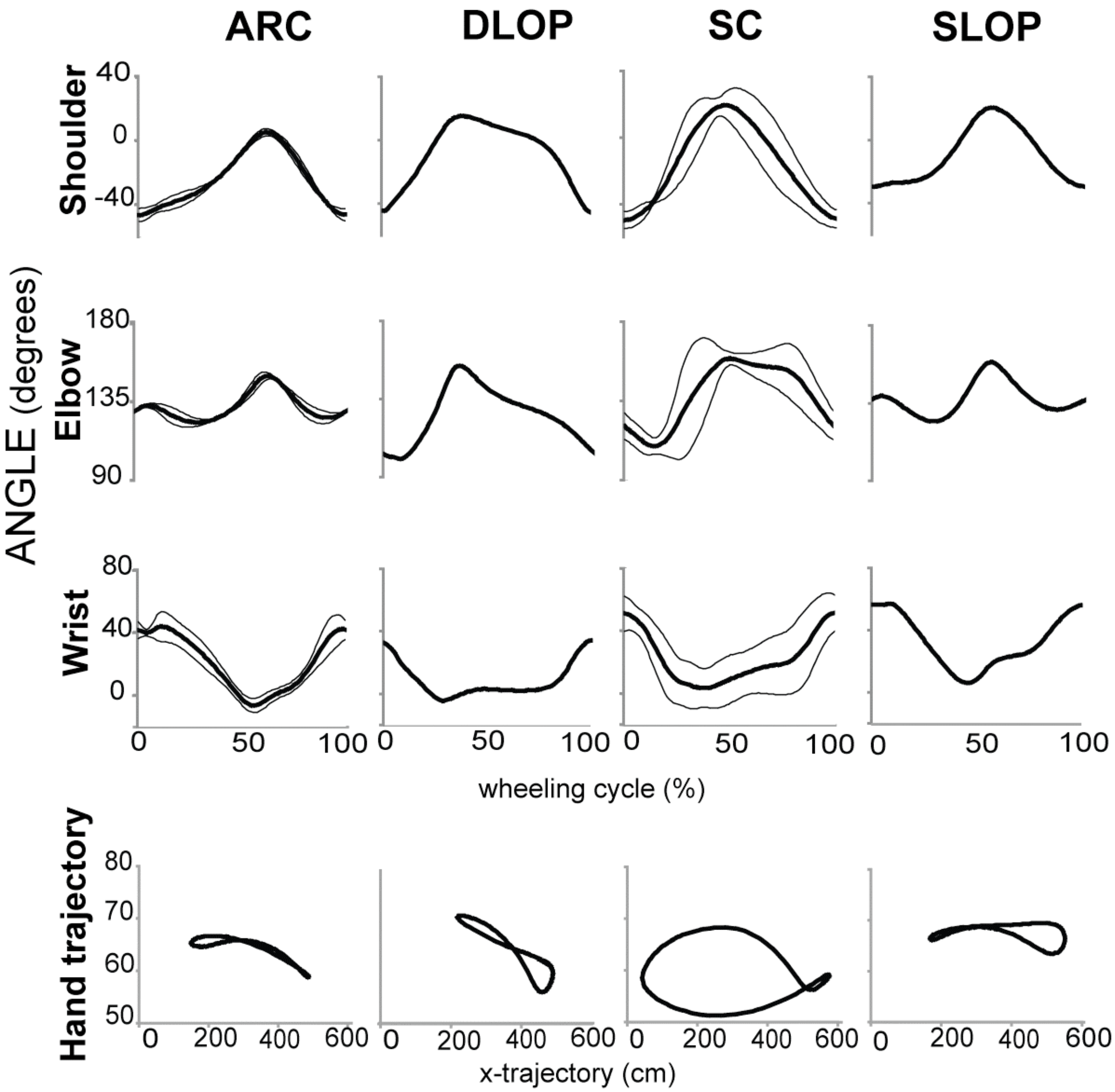
Values are expressed as mean (SD); push/rec, ratio of propulsion to recovery; % Prop, % of wheeling cycle spent in propulsion

Table 2.4 Kinematic variables according to wheeling strategy

	ARCING		DLOP		SEMI-CIRCULAR		SLOP	
Variable	AB	SCI	AB	SCI	AB	SCI	AB	SCI
Shoulder ROM (deg)	62.5 (4.2)	51.5 (6.2)		60.2		72.9 (6.4)	67.5 (7.6)	49.7
Elbow ROM (deg)	52.1 (7.7)	27.2 (3.5)		53.5		56.5 (7.5)	50.1 (7.0)	34.1
Wrist ROM (deg)	40.9 (7.2)	51.1 (4.0)		38.4		64.3 (7.7)	46.3 (6.3)	50.6
peak Shoulder Flexion (deg)	20.6 (6.1)	5.0 (2.1)		15.2		23.1 (6.2)	23.7 (10.0)	20.3
peak Elbow Flexion (deg)	98.2 (3.8)	122.2 (1.5)		100.7		106.9 (6.3)	101.5 (5.3)	124.4
peak Wrist Ulnar Deviation (deg)	-9.4 (7.0)	-6.7 (4.4)		-4.4		-9.1 (5.1)	-5.9 (6.3)	6.1
peak Shoulder Extension (deg)	-41.8 (6.1)	-46.4 (4.0)		-45.0		-49.8 (5.0)	-43.8 (4.2)	-29.4
peak Elbow Extension (deg)	150.3 (7.0)	149.5 (2.0)		154.1		163.4 (7.5)	151.6 (7.4)	158.5
peak Wrist Radial Deviation (deg)	31.5 (10.7)	44.4 (8.5)		34.0		55.2 (9.5)	40.5 (7.6)	56.7
	n=8	n=2	n=0	n=1	n=0	n=5	n=5	n=1

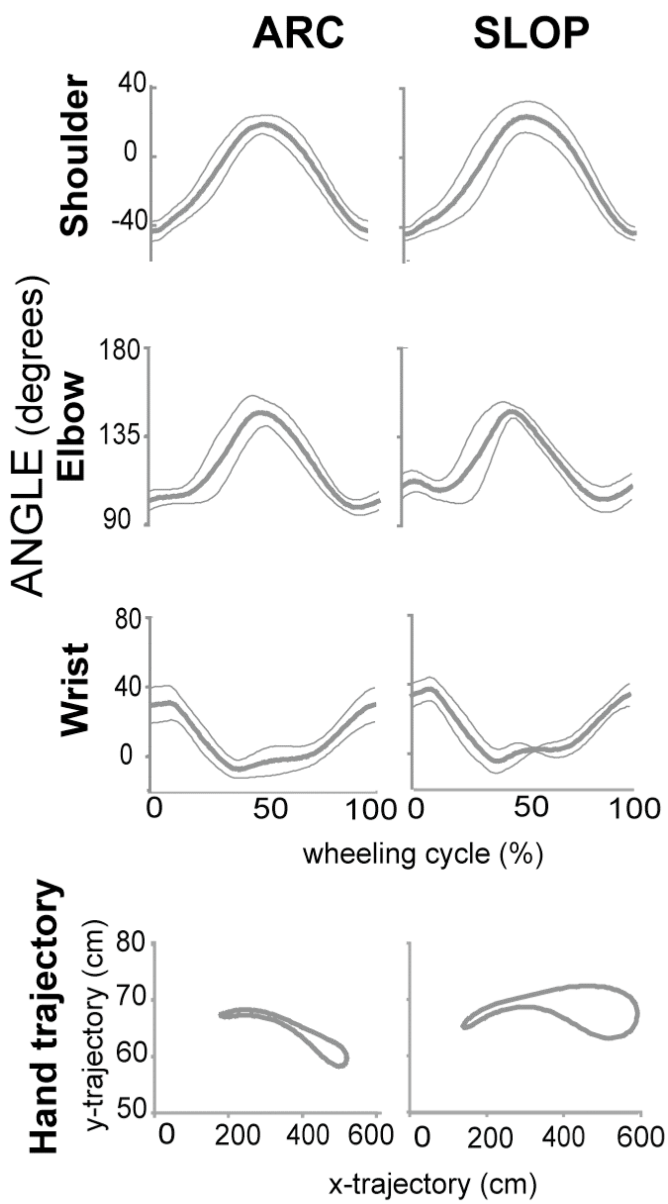
Values are expressed as mean (SD).

Figure 2.5 Manual wheelchair users' kinematics categorized according to wheeling strategy



Averaged experienced MWUs kinematics categorized by wheeling strategy. The plots above are organized according to joint angle measured in degrees (row) and wheeling strategy (column). For the shoulder angle positive values represent flexion, for the elbow angle upward deflections represent extension and for the wrist angle upward deflections represent radial deviation. The thick black line represents the average profile and the thin black lines represent ±SD across the wheeling cycle. The bottom row displays representative hand trajectories that demonstrate the four wheeling strategies.

Figure 2.6 Able-bodied subjects' kinematics categorized according to wheeling strategy

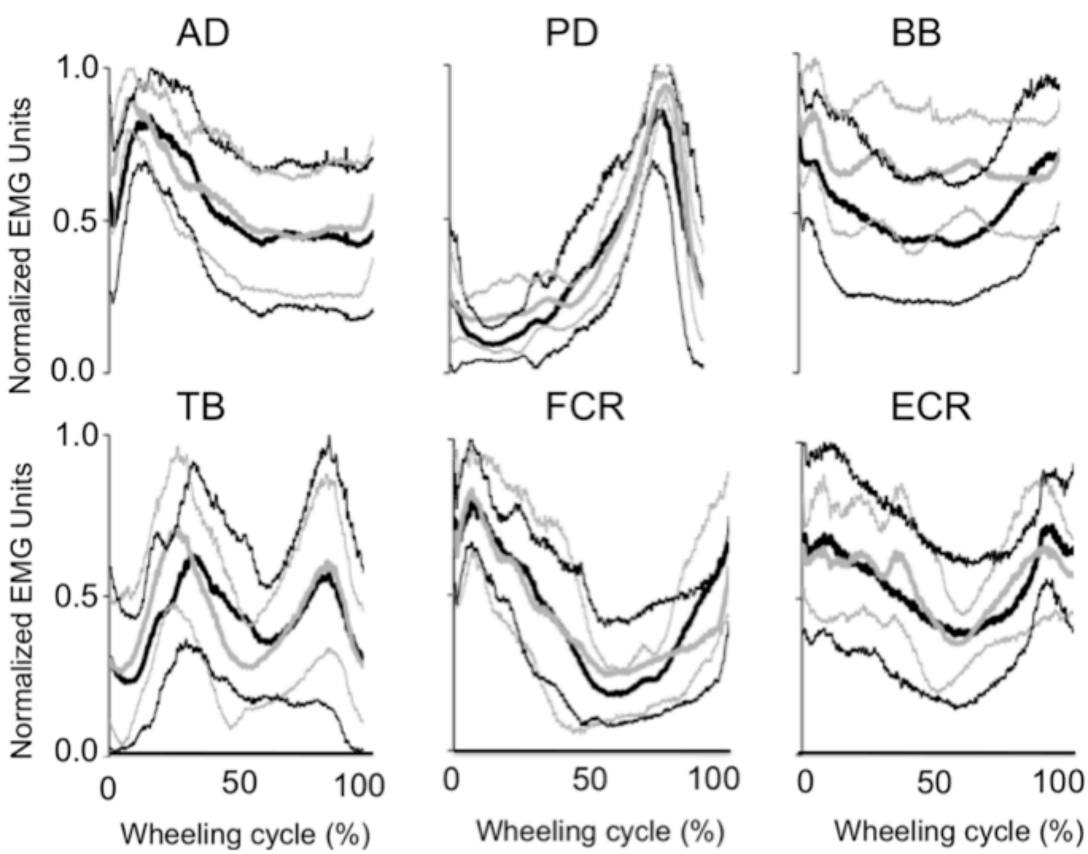


Averaged AB kinematics categorized by wheeling strategy. The plots above are organized according to joint angle measured in degrees (row) and wheeling strategy (column). For the shoulder angle positive values represent flexion, for the elbow angle upward deflections represent extension and for the wrist angle upward deflections represent radial deviation. The thick grey line represents the average profile and the thin grey lines represent SD across the wheeling cycle. The bottom row displays representative hand trajectories that demonstrate the ARC and SLOP wheeling strategies.

2.3.5 Electromyography

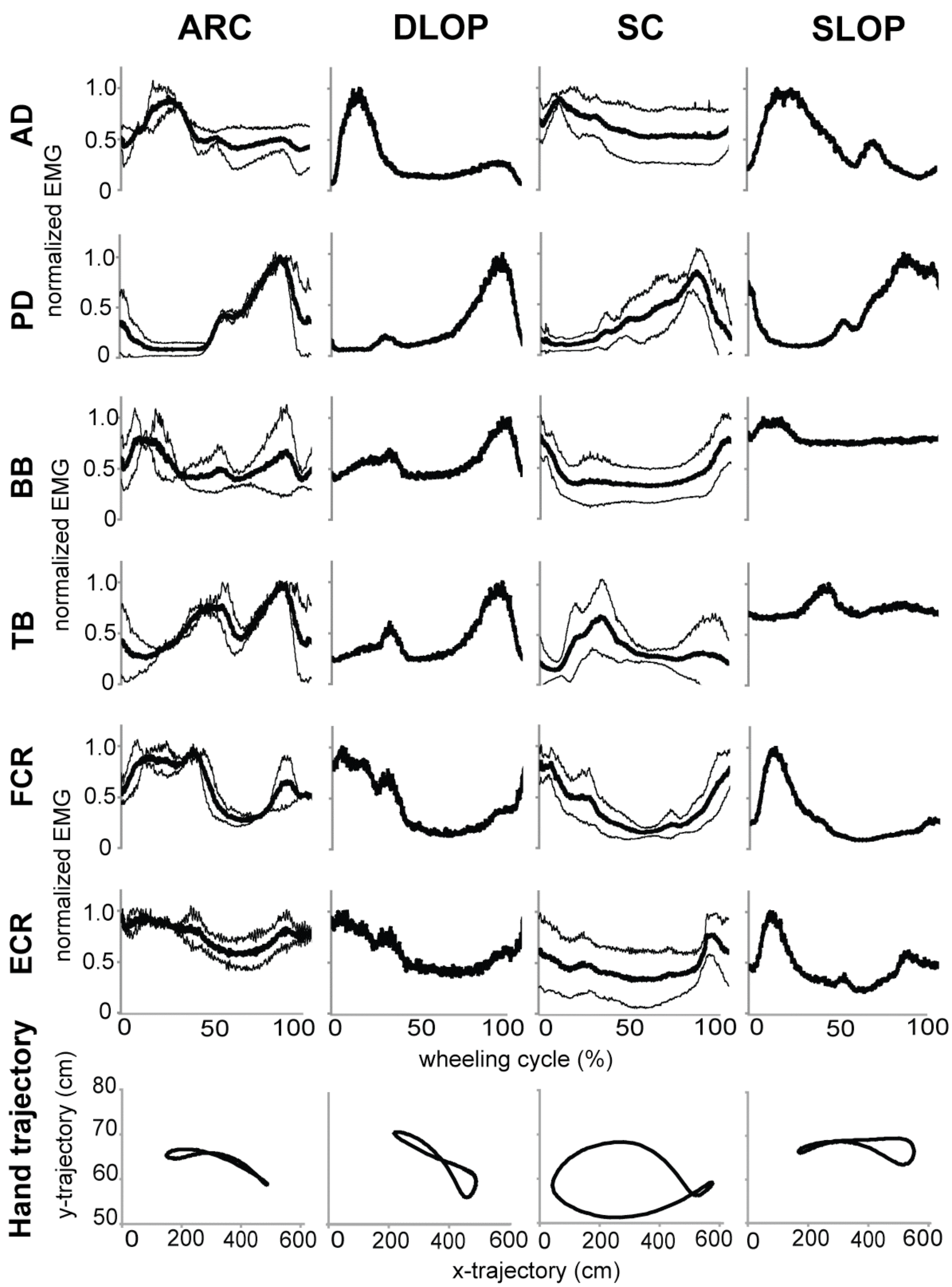
Both the MWUs and able-bodied groups showed similar bursting patterns including similar timing and amplitude of muscle bursts (Figure 2.7). The biceps brachii muscle activation appeared higher during the middle of the wheeling cycle in the AB group. Figures 2.8 and 2.9 illustrate the average muscle activity profiles grouped according to wheeling strategy and subject group.

Figure 2.7 Manual wheelchair users' and able-bodied subjects' normalized EMG during manual wheeling



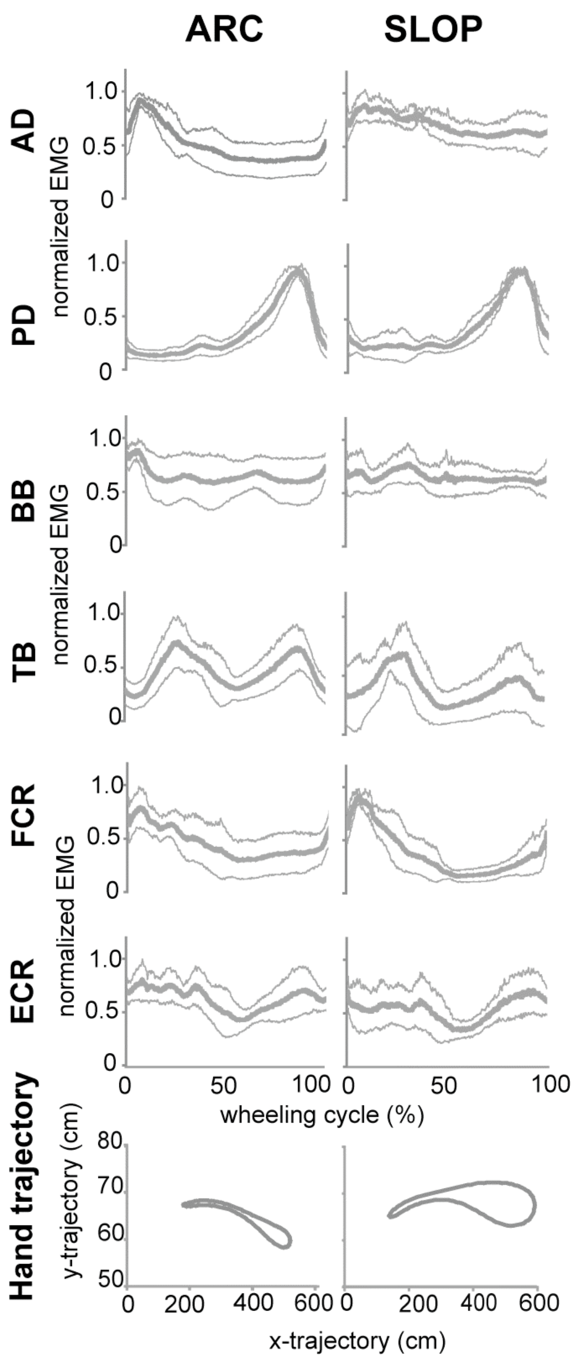
Average normalized EMG for experienced MWUs (thick black) and AB (thick grey) during an averaged wheeling cycle. The thin grey lines represent SD for the AB group and the thin black lines represent SD for the MWUs group.

Figure 2.8 Able-bodied subjects' EMG categorized according to wheeling strategy



Average normalized EMG, with 1 SD if n>1, observed in experienced MWUs categorized by wheeling strategy. The data represent an average wheeling cycle. The bottom row displays representative hand trajectories that demonstrate the four wheeling strategies including ARC (n=2), DLOP (n=1), SC (n=4), and SLOP (n=1).

Figure 2.9 Manual wheelchair users' EMG categorized according to wheeling strategy



Average normalized EMG, with 1SD, observed in the AB group categorized by wheeling strategy. The data represent an average wheeling cycle. The bottom row displays representative hand trajectories that demonstrate the ARC and SLOP wheeling strategies.

Qualitative observations describing muscle activation patterns, based on amplitude and timing of muscle bursts, are described below. Statistics were not performed on this data because of the low numbers of individuals performing each strategy.

Anterior deltoid

The anterior deltoid showed one primary burst occurring in the propulsion phase of the wheeling cycle in all groups. The MWUs who used the ARC, DLOP and SLOP wheeling strategies showed a secondary smaller burst later in the wheeling cycle (Figure 2.7).

Posterior deltoid

The posterior deltoid showed a single major burst in the second half of the wheeling cycle. The burst occurred after the anterior deltoids major burst during the recovery phase of the wheeling cycle (Figure 2.7).

Biceps brachii

Activation patterns in the biceps brachii showed more variability between wheeling strategies. In the MWUs group, there was a single burst during the SLOP and a double burst during the DLOP. In subjects who exhibited the ARC strategy, the biceps brachii showed three bursts of activity, however there was a great amount of variability between these two subjects. Subjects who used the SC wheeling strategy showed a single burst in biceps brachii, beginning before maximum shoulder extension and continuing into the beginning of the wheeling cycle (Figure 2.9). In contrast, in the AB group, biceps brachii showed less bursting activity and appeared to be constantly activated throughout the duration of the wheeling cycle (Figure 2.8).

Triceps brachii

The triceps brachii showed a double bursting pattern in both groups in all wheeling strategies (Figure 2.8 and Figure 2.9).

Flexor carpi radialis

The flexor carpi radialis showed a greater amount of muscle activity in the first half in all wheeling strategies in both groups (Figure 2.8 and Figure 2.9).

Extensor carpi radialis

The extensor carpi radialis showed steady muscle activity followed by a drop at the beginning of the recovery phase and a burst at the end of the cycle in the AB group (Figure 2.9). However, there was more variability in the ECR activation patterns between the different wheeling strategies among the MWUs group. During SLOP, the ECR showed a distinct early burst, whereas during SC there is a distinct late burst in ECR. ECR activity during DLOP and ARC showed a more gradual dip in the second half of the wheeling cycle (Figure 2.8).

2.4 Discussion

Although many manual wheeling studies use AB subjects, it is unknown how similar their biomechanical characteristics are to MWUs and if the results from these studies can be applied to experienced MWUs. This study provides useful information in comparing the kinetics, kinematics and EMG data between AB subjects and experienced MWUs with spinal cord injuries. This study was unique because it compared biomechanical and EMG characteristics between AB and experienced MWUs groups as well as across wheeling strategies. The information gathered from this study should be taken into consideration when applying data from AB subjects to experienced MWUs.

2.4.1 Comparison to the SmartWheel database

There has been a recent effort to describe a standard clinical protocol for the objective assessment of manual wheelchair propulsion through creating a manual wheeling database using only four main kinetic factors obtained by the SmartWheel (Cowan, Boninger et al. 2008). The four selected parameters were velocity, average peak resultant force, push frequency (cadence) and stroke length (push angle). The database contains information from 128 manual wheelchair users and describes the four parameters on tile, carpet and a standard ramp. These four parameters are good general descriptors, however there are some specific information including wheeling strategies as well as specific joint ROM that is lost when only looking at those four parameters.

The findings from the database are fairly different from the results of this study. The self-selected velocities ranged from 1.1-1.2m/s on tile and 0.9-1.0m/s on carpet, whereas the AB and MWUs in this study self-selected average velocities of 0.75 m/s and 0.82 m/s respectively. Peak resultant force was between 67.6 and 77.1 N on tile surface in the database, however the results of this study found that the AB group ranged from 28.7 to 68.8 N and the MWUs group ranged from 27.2 to 63.1. Push angle in the database study was found to be between 97.2 and 104.0 degrees on tile and between 92.9 and 101.5 degrees on carpet. This contrasts the push angles of 44.4 to 70.9 degrees and 61.7 to 104.9 degrees observed in the AB and MWUs groups in our study, respectively. The AB group does not come within the range of the database for velocity or push angle and the MWUs group appears to be slower and utilize shorter push angles compared to the wheelchair users in this study. One reason for this may be that the rollers provide more resistance compared to wheeling on tile floor. This may be more representative of frequently encountered environments such as carpet. Previous studies have demonstrated that steady-state wheeling occurs at 0.8 to 1.6 m/s in level surfaces (Dallmeijer, Kappe et al. 1994; Newsam, Mulroy et al. 1996; Kotajarvi, Sabick et al. 2004; Richter, Rodriguez et al. 2007), therefore the MWUs in this study fell within the range of previously observed velocities.

Differences may also arise in these factors due to the length of the trials. The data for the database is collected for approximately 10 seconds and the first 5 strokes were used for the analysis versus the 10 minutes that subjects wheeled for in this study. It is likely that some individuals will wheel for 10 seconds at a pace that they cannot maintain for longer trials. The results of this study are likely more representative of wheeling outdoors for longer distances and therefore may provide a more representative sampling of biomechanics compared to observing wheeling over a 10 second trial. Within a 10 second trial there is higher chance that subjects will wheel faster than their normal daily speed.

The data observed in this study is more comparable to a study completed by Boninger and colleagues (Boninger, Souza et al. 2002). In that study, the subjects wheeled at 0.9 and 1.8 m/s. It was found that the peak resultant force ranged from 50.4 to 60.9 N at 0.9 m/s (Boninger, Souza et al. 2002). The majority of subjects used the same wheeling strategy for both the 0.9 and 1.8 m/s trials, however, several subjects changed their wheeling strategy with changes in speed. Consequently, we used self-selected wheeling speeds for this study to elicit the most natural wheeling strategy that was comfortable for the individual.

2.4.2 Kinematic differences between AB and experienced MWUs groups

In this study, experienced MWUs primarily used the semi-circular wheeling strategy while most AB subjects used the arcing wheeling strategy. This is consistent with previous literature (Chou 1991). Interestingly, despite using different wheeling strategies, the AB and MWUs groups did not show significant differences in most kinematic variables. Only wrist ROM, maximum wrist radial deviation and maximum elbow flexion showed significant differences between the two groups (Figure 2.5 and Figure 2.6).

MWUs showed larger average values for ROM and maximum radial deviation of the wrist joint indicating that increased wrist ROM, and potentially wrist flexibility, may have more functional relevance to manual wheeling. The larger average wrist ROM among the MWUs may be a result of increased push angle or percent of the cycle spent in propulsion. Increased wrist ROM has been shown to be associated with better nerve conduction of the median and ulnar nerves, less force applied to the pushrim and fewer strokes required to propel a wheelchair a specific distance (Boninger, Impink et al. 2004). Increased ROM of the wrist joint allows for longer wheeling push angles and smoother wheeling cycles. If the wrist has a much smaller ROM, it limits the push angle, as well as the ROM of other joints including the elbow and shoulder angles. Therefore wrist ROM may act as the limiting factor in wheeling kinematics.

2.4.3 Kinetic differences between AB and experienced MWUs groups

The MWUs group spent a longer average percentage of time in propulsion than the AB group. Greater percentage of time spent in propulsion phase is thought to increase efficiency because more work is done to propel the wheelchair forward per cycle and the cadence is often reduced (Boninger, Souza et al. 2002). However, there were no differences observed in cadence in this study. The MWUs group also showed larger average push angles than the AB group. Larger push angle significantly increases time spent in the propulsion phase and is thought to increase wheeling efficiency. The push angle is related to the wrist ROM therefore differences in wrist ROM and push angle between AB and MWUs groups would be expected.

Although there were significant differences in the average tangential force and average resultant force between groups, there was a significant correlation between the average tangential force and body weight for the MWUs group. This indicates that body weight may play a role in generating tangential force in experienced MWUs, however body weight does not appear to influence the resultant force.

2.4.4 EMG differences between AB and experienced MWUs groups

Overall EMG activity was very similar between the MWUs and AB groups. No differences were observed in time to peak EMG amplitude between groups indicating that the general EMG patterns remain consistent across different wheeling strategies. However, the DLOP wheeling strategy did exhibit some different muscle activation patterns in some muscles. The anterior and posterior deltoids, and biceps and triceps brachii displayed two distinct peaks rather than the one seen in some of the other wheeling strategies (Figure 2.8).

Muscle activation patterns have been shown to differ between paraplegic MWUs and AB subjects during manual wheeling (Dubowsky, Sisto et al. 2009). Paraplegic MWUs have been shown to use a greater percentage of their maximum voluntary contraction (MVC), however this cannot be compared to the current study because the subjects' MVCs were not evaluated. Dubowsky and colleagues also observed differences in triceps muscle activation, however these differences were not observed in the current study. It was concluded that the greater percentage of MVC observed in the MWUs group may be responsible for greater joint forces and fatigue which may result in shoulder injury or pain (Dubowsky, Sisto et al. 2009). Greater average forces were observed in the MWUs compared to the AB group, therefore this could be linked to a greater percentage of MVC, however it is uncertain because that variable was not measured in the current study. Despite the MWUs greater forces, they experienced very minimal shoulder pain according to the WUSPI. Future research should examine the relationship between force and percentage of MVC during manual wheeling.

2.4.5 Limitations

The use of rollers in this experiment makes it difficult to compare to other terrains naturally observed in the environment. It appears that the data is not comparable to tile surface based on the data from the SmartWheel data base study (Cowan, Boninger et al. 2008). The length of time spent wheeling in this study is unique and all efforts were made to reduce fatigue in individuals, however it is possible that some people may not have admitted to feeling fatigued and continued with the data collection. Administering a scale of perceived exertion could have been useful in determining subjects' level of effort and if they felt fatigued. It is very possible that people will wheel differently when completing a 10 second trial versus a steady-state 10 minute trial, therefore it may be insufficient to compare this study to other studies consisting of very short trials. Although short bouts of manual wheeling may be indicative of wheeling indoors and have high functional relevance, longer bouts of wheeling are quite important. Longer bouts of wheeling would be used when exercising, or traveling longer distances in the community and would be critical in maintaining aerobic capacity and maintaining a high level of functional independence.

The kinematic analysis were based on 2-dimensional calculations of joint movements. We assumed that wrist movements were mainly occurring along the radial/ulnar deviation axis. However, there may have been some degree of wrist flexion and extension that we could not account for. In addition, the shoulder angle calculated was relative to the vertical, so possible contributions of trunk movements to the wheeling strategies could not be accounted for. The biomechanics of manual wheeling can be influenced by the length of the arms and by the wheelchair configuration including the seat height (Consortium for Spinal Cord Medicine, 2005). Seat height was not recorded in this study therefore it is uncertain if the seat height may have influenced the wheeling strategy, push length or other biomechanical characteristics of manual wheeling. Seat height was not controlled because we wanted to examine wheeling strategies that were natural to the MWU subjects. The seat height for subjects in the AB group was adjusted to a level that they reported was comfortable for them. Generally, all subjects in the AB group were seated so that the iliac crest was at the same height as the top of the wheel. The clinical guidelines for the Paralyzed Veterans of America have described that lower seat heights are believed to facilitate greater movement of the arms, allow for a lower wheeling cadence and a higher mechanical efficiency (Consortium for Spinal Cord Medicine, 2005).

It is also possible that the difference in age between the groups influenced the results, however all subjects were healthy and had begun wheeling at fairly young ages. There is no known literature to suggest that the biomechanics between these two age groups should differ. A study examining elderly manual wheelchair users with a mean age of 68.2 found that 56.3% of subjects wheeled with strategies utilizing arcing characteristics (Aissaoui and Desroches 2008). We found that 22% of experienced manual wheelchair users the arcing strategy, however none of these individuals were elderly. The range of wheeling strategies in the experienced MWUs group observed in our study were similar to those presented in other studies involving younger subjects (Chou, Su et al, 1991; Boninger, Souza et al, 2002). This suggests that the age of the subjects in the two groups did not affect wheeling strategy.

2.5 Conclusion

It is clear that there are some major differences between AB subjects and experienced MWUs, therefore, when possible research should be conducted on experienced MWUs. These differences in kinetic (average force, push angle and percent time spent in propulsion) and kinematic (wrist ROM, maximum wrist radial deviation, and maximum elbow flexion) variables should be a focus in future research and may provide insight into new teaching strategies when learning to manual wheel. These differences observed may also be key in the prevention of overuse injuries.

2.6 References

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3 CUTANEOUS REFLEXES DURING MANUAL WHEELING AND ARM CYCLING IN ABLE-BODIED SUBJECTS AND EXPERIENCED MANUAL WHEELCHAIR USERS[2]

3.1 Introduction

There are many systems that are involved with producing, controlling and adapting rhythmic human movement such as walking or manual wheeling. The three core components required in the control of locomotion involve supraspinal input, spinal input, and sensory feedback (Zehr 2005). With increasing complexity, more specialized networking is required between various systems. Vision, propriocpeption, vestibular input are only some of the systems involved in completing successful movement. However, rhythmic movement, in its simplest form, is thought to be derived from the oscillation of membrane potentials among collections of neurons in the spinal cord known as central pattern generators (CPGs) (Brown 1911; Pearson 1993). The CPG is believed to generate the basic features of human locomotion without requiring sensory feedback.

Nearly a century ago, Brown (1911) proposed a schematic of a CPG consisting of oscillating neural circuitry or ‘half-centers’ based on reciprocal inhibition. This CPG was thought to reside in the lumbar spinal cord (Brown 1911). This half-center model can be used to describe the rhythmic and applied to the alternating flexion and extension of the legs during human walking.

[2] A version of this chapter will be submitted for publication. MacGillivray, M., Klimstra, M., Zehr, E.P., Sawatzky, B. and Lam, T. (2010) “Cutaneous Reflexes During Manual Wheeling and Arm Cycling in Able-Bodied Subjects and Experienced Manual Wheelchair Users.”

Excitation of a flexor half-center in one leg will cause inhibition of the flexors in the alternating leg so that contact with the ground is maintained. Other forms of locomotion, including manual wheeling, involve symmetrical bilateral movement, however the flexors and extensors alternate within the limb.

Since Brown's early discoveries in 1911, much research has been dedicated to looking for evidence of the CPG within animals and humans. Direct evidence for a CPG exists for many animal models, from insects (Wilson 1964; Wolf and Pearson 1988) to the lamprey (Wallen 1997; Grillner 2006) to cats (Forssberg and Grillner 1973; Forssberg, Grillner et al. 1980; Perret and Cabelguen 1980; Grillner and Zangger 1984). Direct evidence in these models can be achieved through eliminating supraspinal input and sensory feedback pathways to ensure that the spinal cord is responsible for generating the rhythmic movement.

There is far less substantiating evidence for a human CPG. Infants are of particular interest when exploring evidence for a human CPG because the neocortex and descending motor tracts are immature in infants prior to one year of age (reviewed in (Yang, Lam et al. 2004)). Although the infants do not have the ability to walk independently, they are capable of initiating gait-like cyclical movements given the appropriate sensory input (Yang, Stephens et al. 1998). Further evidence for a locomotor CPG in humans comes from observations in individuals with severe incomplete spinal cord injury (Calancie, Needham-Shropshire et al. 1994; Dimitrijevic, Gerasimenko et al. 1998). Calancie et al (1994) reported data from an individual with SCI who experienced bouts of involuntary stepping-like movements when he lay on his back with the hips extended. Another study showed that application of tonic electrical stimulation of the posterior portion of the upper lumbosacral cord in complete paraplegics elicited gait-like movement as well as gait-like EMG patterns (Dimitrijevic, Gerasimenko et al. 1998). This study supports the idea that the human spinal cord can generate fundamental locomotor patterns without any supraspinal influence.

The studies mentioned above are difficult to perform and often require invasive measures, therefore other methods and protocols have been developed to investigate the properties of putative central pattern generators in humans. More recent studies have looked at changes in cutaneous reflexes throughout a movement cycle. Reflex responses are measured through the use of electromyography and typically demonstrate a triphasic response to stimulation of a cutaneous nerve. The early latency response of the cutaneous reflex occurs in the range of 50 to 75 milliseconds (Zehr and Chua 2000; Zehr and Kido 2001). The short-latency of this response suggests changes within this period would indicate involvement at the spinal level. Middle latency responses are seen from 75 to 120 miliseconds following stimulation (Zehr and Chua 2000; Zehr and Kido 2001) are considered to likely involve supraspinal input. There is a third (late) latency reflex response that occurs after 120ms post stimulation, however that response may incorporate many pathways and is therefore more difficult to interpret. Neural control, possibly regulated by a CPG or series of CPGs, of the upper limbs has been explored over the past decade. Similar to what has been observed during walking, the processing of sensory input from cutaneous receptors in the upper limbs during rhythmic movement (arm cycling) also exhibits phase dependent and task-dependent modulation (Zehr and Chua 2000; Zehr and Kido 2001).

A cutaneous reflex may be modulated by the phase of a movement as well as the type or form of movement. For example, within active arm cycling, many muscles exhibit reflexes of varying amplitudes across the movement cycle which are significantly related to movement position (Zehr and Chua 2000; Zehr and Kido 2001). This indicates that the reflex responses are phase modulated according to movement position within a cycle. Differences in cutaneous reflex modulation has also been observed between active versus passive cycling. Passive bilateral arm cycling elicits small cutaneous reflexes in response to superficial radial nerve stimulation and rarely showed significant modulation of reflex amplitude across the cycle (Carroll, Zehr et al. 2005). In fact, the lack of modulation for passive cycling was similar as observed if the limb was stationary (Carroll, Zehr et al. 2005). This indicates that afferent information received during passive cycling is not enough to elicit modulation of cutaneous reflexes. Presumably the involvement of CPG circuits during active cycling could underlie the modulation of reflexes when the arms are actively engaged in a rhythmic movement.

It has been concluded that reflexes modulate similarly during rhythmic movements of the upper and lower limbs (Zehr and Kido 2001). Rhythmic EMG patterns, phase and task dependent reflex modulation, as well as variations in reflex amplitude in active and passive cycling have all been exhibited in both upper and lower limb rhythmic movements (Zehr, Komiyama et al. 1997; Zehr and Chua 2000; Zehr, Collins et al. 2001; Zehr and Kido 2001; Zehr, Collins et al. 2003; Zehr and Haridas 2003; Carroll, Zehr et al. 2005; Zehr, Klimstra et al. 2007). However, differences have been found when examining contralateral limb movement. Movement of one arm does not modulate reflex amplitude in the contralateral arm (Tax, Van Wezel et al. 1995; Carroll, Zehr et al. 2005), however movement of one lower limb shows modulation of reflex amplitude in the contralateral limb (Tax, Van Wezel et al. 1995; Van Wezel, Ottenhoff et al. 1997). This suggests that the legs are more strongly coupled than the arms in humans. The lack of interlimb coupling in the arms may serve the important function of bimanual dexterity or provide the ability to more easily perform asynchronous movements with the upper limbs.

Previous studies have consistently shown that cutaneous reflex responses can be measured during upper limb cycling and rhythmic arm swing (Carroll, Zehr et al. 2005; Zehr and Kido 2001; Zehr and Chua 2000), however no studies have examined cutaneous reflexes during manual wheeling. This task differs from other upper limb tasks studied in the past because it has the functional goal of producing locomotion. There are many people who become dependent on manual wheelchairs for mobility following some type of injury. Are there differences in afferent modulation of upper extremity movements once someone becomes reliant on using rhythmic arm movements on a daily basis for mobility? Therefore, the overall purpose of this study is to determine whether there are any differences in cutaneous reflex modulation across manual wheeling and arm cycling tasks and between able-bodied subjects and experienced manual wheelchair users.

3.1.1 Hypothesis

It is hypothesized that:

1. reflex responses to stimulation of the superficial radial nerve during manual wheeling will be modulated according to the phase of the wheeling cycle.

2. reflex responses to stimulation of the superficial radial nerve during manual wheeling will exhibit a larger modulation Index (demonstrate larger changes in amplitude) compared to synchronous arm cycling.

3. reflex responses to stimulation of the superficial radial nerve during manual wheeling will exhibit a larger modulation index (demonstrate larger changes in amplitude) in experienced manual wheelchair users compared to able-bodied inexperienced manual wheelchair users.

3.2 Methods

3.2.1 Participants

A total of 22 individuals 13 able-bodied (AB) inexperienced manual wheelchair users (MWUs) and 9 experienced MWUs were recruited to participate in this study. The inclusion criteria for the MWUs included using a manual wheelchair for more than one year and having a SCI at T1 and below. Additional inclusion criteria required that all subjects were over 19 years of age, capable of wheeling for 5-10 minute spans, and provided informed consent. People were excluded from the study if they had any injury to their upper-limbs or if they had upper-limb or nerve pain that would prohibit them from wheeling a manual wheelchair. Data from the right side of the body was used in this analysis. Demographic information including age, height, weight, sex, injury level, age of injury and years since injury, was collected from all subjects. All participants with SCI were given the Wheelchair Users Shoulder Pain Index (WUSPI) (Curtis, Drysdale et al. 1999) to determine if they experienced shoulder pain related to manual wheelchair use.

3.2.2 Experimental set-up

After providing informed consent and filling out the demographic questionnaires and WUSPI, the subjects' right arms were prepared for EMG, nerve stimulation, and kinematic data collection. All subjects completed both manual wheeling and arm cycling tasks. For the purpose of this study, arm cycling and manual wheeling were considered different tasks because they demonstrate different kinematics and muscle activation patterns.

Manual wheeling

MWUs used their own manual wheelchair while able-bodied participants used a fitted ELEVATION™ wheelchair (Instinct Mobility, Vancouver, BC, CA). Participants' wheelchairs were secured to the rollers (Figure 3.1). They were asked to wheel at a comfortable, self-selected, wheeling speed. All participants were given adequate time to familiarize themselves with the rollers and find their self-selected wheeling cadence. To ensure that the cadence did not change throughout the course of the study, the subjects were asked to maintain their frequency by synchronizing with a metronome.

Electrical stimuli was delivered to the superficial radial nerve pseudorandomly, approximately every 3-5 seconds (not more than once per wheeling cycle) using a custom-made LabView program (National Instruments Corporation, Austin, TX, USA). EMG and motion capture data were collected simultaneously and later synchronized with a square-wave pulse sent to all collection devices. To ensure adequate stimuli were delivered to each phase of the wheeling cycle to see a clear reflex response, at least 8 minutes of data were collected. Rest breaks were provided as necessary to reduce fatigue.

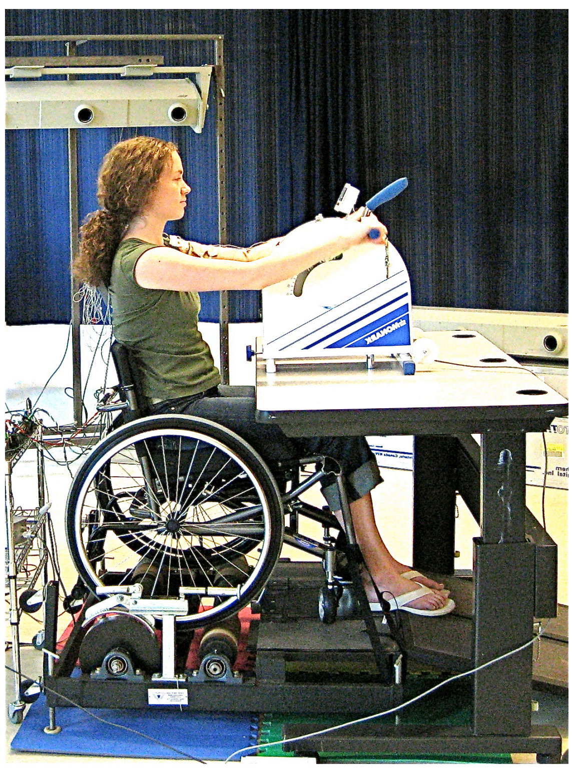
Figure 3.1 Experimental set-up: manual wheeling

This photograph represents the experimental set up. The wheelchair was safely secured to the rollers. The electrodes, motion capture sensors and the stimulating electrode were all placed on the right arm.

Arm cycling

Subjects also performed symmetrical (in-phase) arm cycling on an instrumented and modified cycle ergometer (Monark) (Figure 3.2). The ergometer was mounted on a table of adjustable height so that the table was level with the forearm when the elbow was flexed at 90 degrees and the upper arm was perpendicular to the floor. Subjects cycled at the same frequency as wheeling so that differences could not be attributed to changes in frequency.

Figure 3.2 Experimental set-up: arm cycling



This figure shows the arm cycling task. The electrodes, motion capture sensors and the stimulating electrode were all placed on the right arm as indicated in the photograph.

Kinematics

A motion analysis system Optotrak 3020 (NDI, Waterloo, ON, CA) was used to record sagittal-plane upper limb kinematics on the right side of the body during wheeling. Two cameras positioned at approximately 45° angles between the horizontal and sagittal planes, creating a 90° angle between the two cameras, were used to collect the data. Active infrared markers were placed on the following landmarks: 3rd metacarpophalangeal joint, radial styloid, lateral epicondyle, and acromion to represent the hand, wrist, elbow, and shoulder, respectively. Kinematic data was collected at a sampling frequency of 200 Hz. All data was filtered using a fourth order, 7Hz low-pass Butterworth filter. The shoulder joint angle was calculated using the upper segment of the arm (lateral epicondyle to acromion) and the vertical (projecting down from the acromion) while the elbow joint was calculated using the lower segment of the arm (lateral epicondyle to radial styloid) and upper segment of the arm (lateral epicondyle to acromion). Lastly the wrist angle was calculated using the lower segment of the arm (lateral epicondyle to radial styloid) and the hand (lateral styloid to 3rd metacarpophalangeal joint).

Electromyography (EMG)

The skin above the muscles of interest was shaved, exfoliated and wiped with an alcohol pad to help remove dirt and oils that may interfere with the EMG signal. EMG activity was recorded from the anterior deltoids (AD), posterior deltoids (PD), biceps brachii (BB), triceps brachii (TB) (long head), flexor carpi radialis (FCR) and extensor carpi radialis (ECR) (brevis) (Carroll, Zehr et al.

2005) with ground electrodes placed over the olecranon process and clavicle. EMG data from the AD, PD, BB and TB were collected with a 4-lead Grass amplifier (7HP511, Grass Technologies, Warwick, RI) while EMG data from the FCR and ECR muscles were collected with a Life Electronics 2 lead system. EMG signals were collected at 1000 Hz, pre-amplified (×2000-5000), and band-pass filtered between 100 and 300 Hz. Further offline analysis was done with a custom made program using MATLAB (Mathworks, Natwick, MA, USA). Data was filtered using a fourth order, 100 Hz low pass Butterworth filter and full-wave rectified.

Nerve stimulation

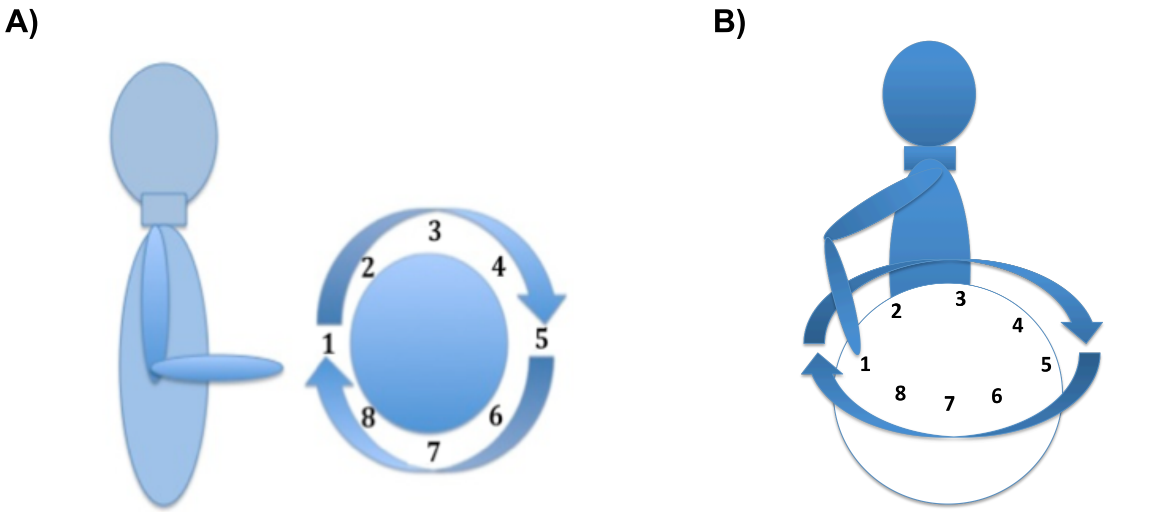
The superficial radial nerve, which innervates the skin of the dorsal-lateral aspect of the hand, was identified on the right side of the body over the lateral portion of the wrist. The skin was shaved as necessary and cleaned. Flexible 1-cm disposable Ag-AgCl surface EMG electrodes (Grass Instruments, AstroMed Inc., West Warwick, RI, USA) were placed on the skin on the dorsal surface of the forearm just proximal to the radial head and the crease of the wrist joint. Electrical stimuli was delivered with trains of 5 x 1.0-ms pulses at 300 Hz (Grass S88 Stimulator, Grass Instruments, AstroMed Inc., West Warwick, RI, USA). For each participant, the radiating threshold was identified. Radiating threshold was defined as the minimum stimulus intensity required to elicit a non-painful sensation spreading over the skin area supplied by the nerve. The stimulus amplitude for the experiment was approximately 2 times the radiating threshold.

3.2.3 Data analysis

Kinematic, EMG and nerve stimulation data were synchronized by an external trigger pulse signal and were cut into cycles based on maximum shoulder extension. For graphical purposes, data were normalized in time to 100% of the cycle. For each subject, the EMG data for each muscle was normalized to the peak rectified EMG value of that muscle during the non-stimulated wheeling or arm cycles for later descriptive group analyses. Shoulder, elbow and wrist angles were also calculated from non-stimulated cycles and averaged for both groups for descriptive purposes.

For each individual, all cycles containing a single stimulation were categorized into 8 bins (Figure 3.3A and 3.3B). All reflex responses occurring in a given bin were averaged together to produce 8 averaged reflex responses for the 8 bins of the movement cycle. The early and middle latencies of the cutaneous reflex response were identified. The early latency reflex responses were those that occurred in the range of 50 to 75 milliseconds post stimulation and middle latency reflex responses were those that occurred in the range of 75 to 120 milliseconds post stimulation (Zehr and Chua 2000; Zehr and Kido 2001). The peak reflex response was identified and a 20-ms window around the peak was averaged to provide a representative amplitude of the reflex response. EMG amplitudes were recorded for stimulated trials (EMG_{STIM}) and then the background EMG (EMG_{BG}) was subtracted from the stimulated data to achieve a subtracted value (EMG_{SUB}), which was representative of the reflex response ($EMG_{STIM} - EMG_{BG} = EMG_{SUB}$). Background EMG was defined as the EMG amplitude of the average non-stimulated cycles at the same time of the cycle as when the stimulus occurred.

Figure 3.3 Dividing manual wheeling and arm cycling into 8 chronological bins



This figure represents how the arm-cycling cycle and manual wheeling cycle were divided into 8 bins. Bin 1 corresponds with maximum shoulder extension occurring at the beginning of the cycle. The cycles each consist of 8 chronological bins.

3.2.4 Statistical analysis

The significance of all statistical tests was evaluated at $p < 0.05$ and conducted using SPSS (SPSS Inc. Chicago, Il, USA).

The Spearman's Rank correlation (r_S) was used to determine if EMG_{SUB} was correlated with EMG_{BG}. Significant correlation would indicate that the reflex amplitude was dependent on the amplitude of the EMG_{BG}.

The Wilcoxon signed ranks test was performed to compare EMG_{BG} with EMG_{STIM} for each muscle for each group (AB cycle, AB wheel, MWU cycle and MWU wheel). This test was conducted to examine whether there was a difference between the EMG_{STIM} and EMG_{BG}. Differences indicated a reflex response. Next the Friedman's ANOVA was used to detect if there was significant modulation of the reflex responses across the 8 bins of the movement cycles. To ensure that the reflex responses were being evaluated rather than the change in EMG_{BG}, the EMG_{SUB} was used for this test.

A Modulation Index (MI) was calculated by taking the maximum reflex amplitude achieved for a particular latency of a specific muscle and subtracting the minimum reflex amplitude achieved across the movement cycle. The MI was used to determine if the absolute difference in the amount of modulation was different between the two groups and tasks. This method provides an indication of the amount of modulation occurring across the movement cycle. The Kruskall-Wallis ANOVA was used to determine if there were significant differences in MI between tasks (cycling and wheeling) and if there were differences between groups (AB and MWUs).

The r_S was used to determine if there was a correlation between the amount of modulation (MI) among the MWUs group and the years of manual wheeling experience.

3.3 Results

3.3.1 Subjects

The specific characteristics of the manual wheeling group and able-bodied group can be observed in Tables 3.1 and 3.2 respectively. The results of the WUSPI confirmed that the experienced MWUs did not experience shoulder pain.

Table 3.1 Manual wheelchair users participant data

Subject	Sex	Age (yr)	Injury level	years since injury	Dominant hand	Height (cm)	Weight (kg)
1	M	36	T3	18	R	188.0	88.5
2	M	48	T4	7	L	188.0	79.4
3	M	39	T4	20	R	182.9	68.0
4	F	34	T5	29	R	163.0	47.6
5	M	27	T5	6	R	177.8	59.0
6	F	46	T10	27	R	162.6	40.9
7	F	54	T7	19	R	160.0	65.3
8	F	52	T1	14	R	165.1	72.6
9	M	66	T2	2	R	190.5	90.7

Table 3.2 Able-bodied participant data

Subject	Sex	Age (yr)	Dominant hand	Height (cm)	Weight (kg)
1	F	26	R	175.3	63.5
2	F	28	R	172.7	61.2
3	M	25	R	179.1	77.6
4	F	28	R	172.7	65.8
5	F	24	R	160.0	65.8
6	M	28	R	183.0	69.3
7	F	25	R	154.9	50.8
8	M	27	R	180.0	62.0
9	F	25	R	157.5	56.7
10	F	24	R	175.3	64.9
11	M	25	R	180.3	72.6
12	F	26	R	170.0	54.0
13	F	24	R	167.6	70.3

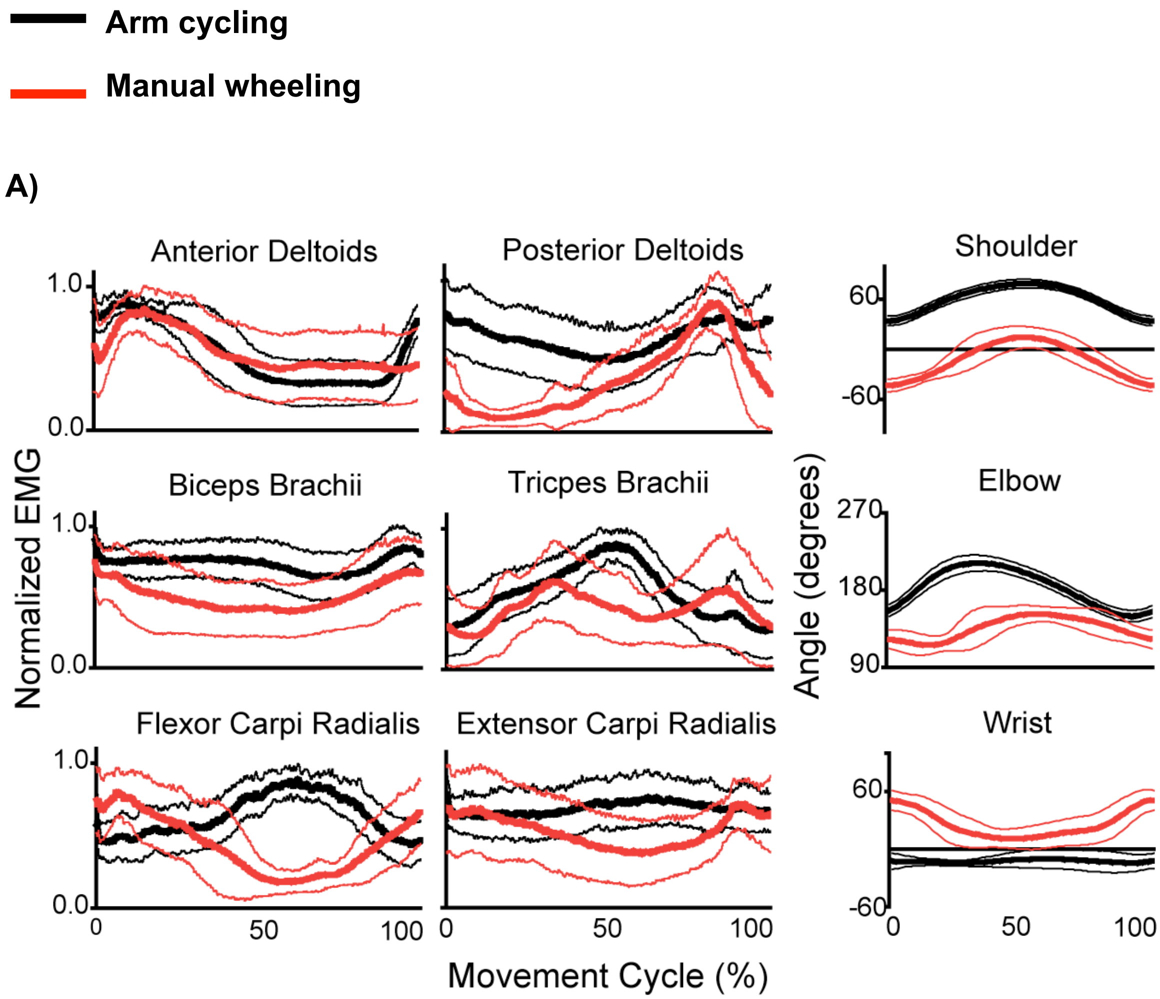
3.3.2 Background EMG and kinematics

Background EMG demonstrated similar timing and amplitudes of bursting activity for the AB and MWU groups for both the cycling and wheeling tasks (Figure 3.4). The AD and PD showed a strong reciprocal activation pattern for the wheeling trials in both groups. Although there are three pairs of antagonist muscles, the PD and AD were the only pair that showed a very clear antagonistic relationship during manual wheeling.

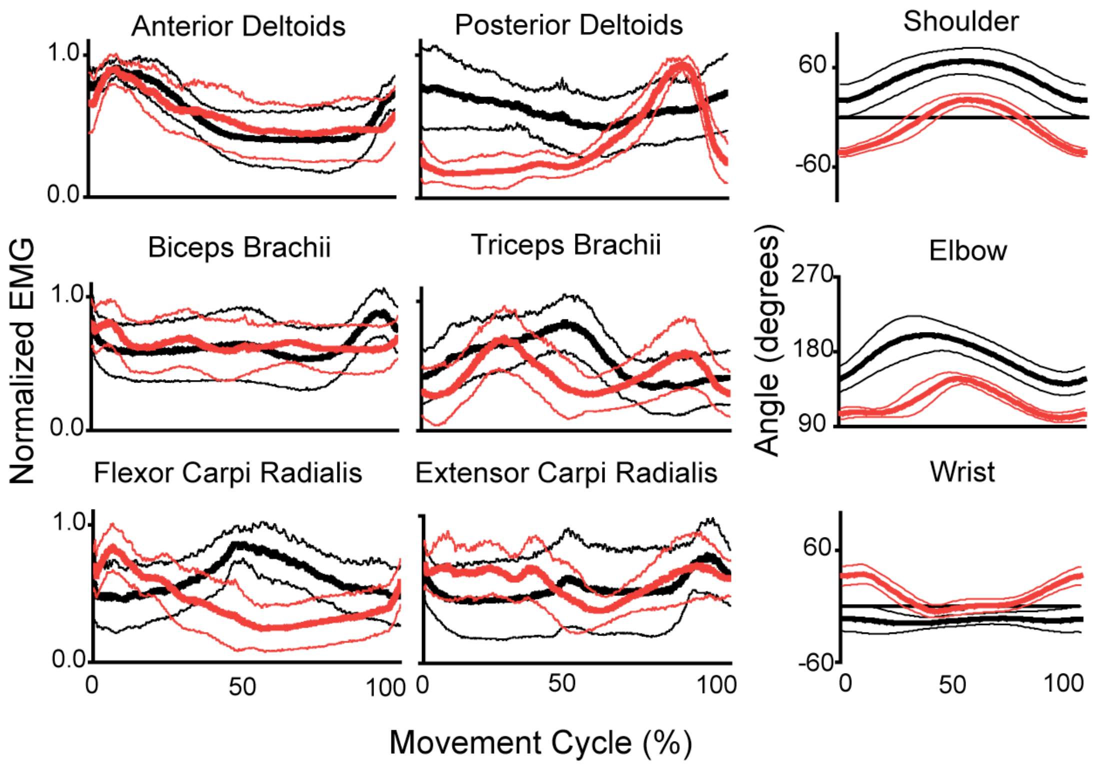
There appeared to be a few notable differences between the EMG profiles during wheeling and cycling. The AD, BB and ECR muscles demonstrated very similar muscle activation patterns including similar timing and amplitude of bursting activity between the two tasks, while the PD, TB, and FCR muscles showed different muscle activation patterns between the two tasks. During wheeling, the PD showed a strong burst in the latter half of the wheeling cycle, which corresponds with the recovery phase. This differs from the more constant muscle activity produced during arm cycling. The TB exhibited two distinct bursts corresponding with the propulsion and recovery phases during wheeling. During wheeling, the TB displayed a single peak during the cycle corresponding to elbow extension. Lastly, the FCR shows a peak and then drop during wheeling and a peak during the middle of the movement during cycling (Figure 3.4).

The joint angles also demonstrated similar timing and amplitudes between groups for each task (Figure 3.4A vs. B, right panels). However, there were differences in joint angles between the manual wheeling and arm cycling tasks (Figure 3.4).

Figure 3.4 Background EMG during manual wheeling and arm cycling in able-bodied subjects and experienced manual wheelchair users



B)



The graphs above represent the EMG activity and joint angles during wheeling and arm cycling. Figure A represents the data for MWUs while Figure B represents the data for able-bodied individuals. For the shoulder angle positive values represent flexion and for the elbow angle upward deflections represent extension. During manual wheeling upward deflections of the wrist angle represent radial deviation and for arm cycling negative values represent wrist flexion.

3.3.3 General reflexes and modulation during wheeling

Anterior deltoids and posterior deltoids

In the AD, the greatest reflex responses in both groups occurred during bins 1-3 corresponding with the initiation of propulsion and the propulsion phases (Figure 3.5A). There was an early latency inhibitory response as well as a middle latency excitatory response during this time.

PD demonstrated greatest reflex responses during bins 5-7 of the wheeling cycle in both groups (Figure 3.5B). This phase corresponds with the end of recovery and the initiation of handrim contact. The early latency reflex response was excitatory while the middle latency response was inhibitory. These trends are opposite that seen in the AD muscle.

Biceps brachii and triceps brachii

BB showed greatest reflex responses during bins 5-7 in the AB group and bins 6-8 in the MWUs group (Figure 3.6A). Those phases correspond with early and late recovery respectively. Early latency responses demonstrated an excitation while middle latency demonstrated a slight inhibitory response.

TB showed greatest reflex responses during the first couple of bins 1-3 in the AB group and bins 6-8 in the MWUs group (Figure 3.6B). There was an inhibitory response in the first few bins during early latency and an excitation response in the first few bins during the middle latency response.

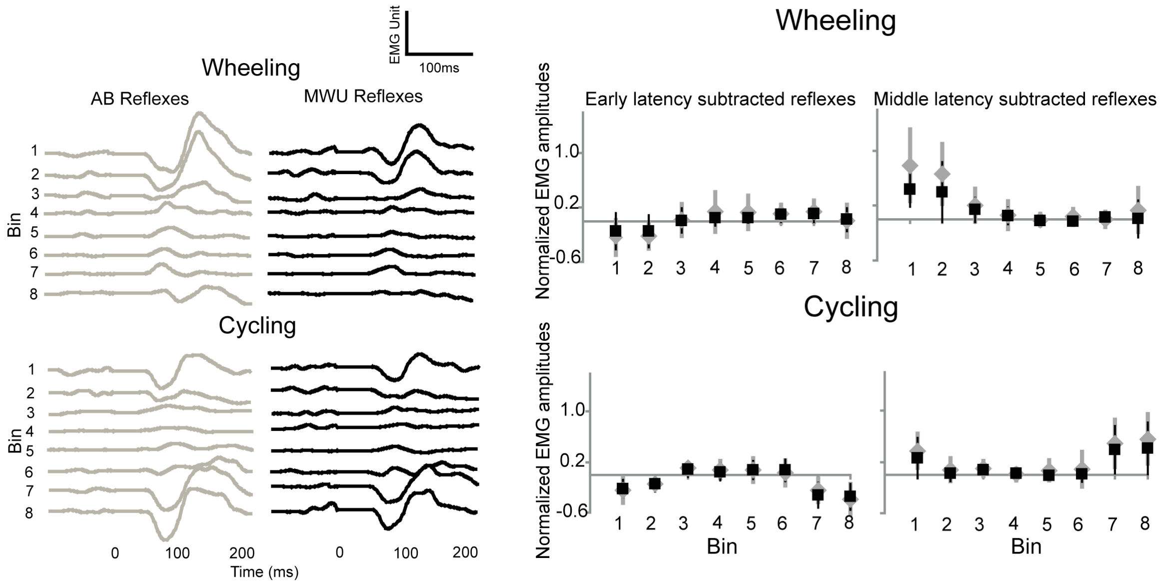
Flexor carpi radialis and extensor carpi radialis

FCR showed greatest reflex responses during bins 1-3 during wheeling (Figure 3.7A). The amplitude of the reflexes were much larger in the AB group. Early latency reflexes evoked a large facilitation response whereas the middle latency reflexes evoked a slight facilitation response.

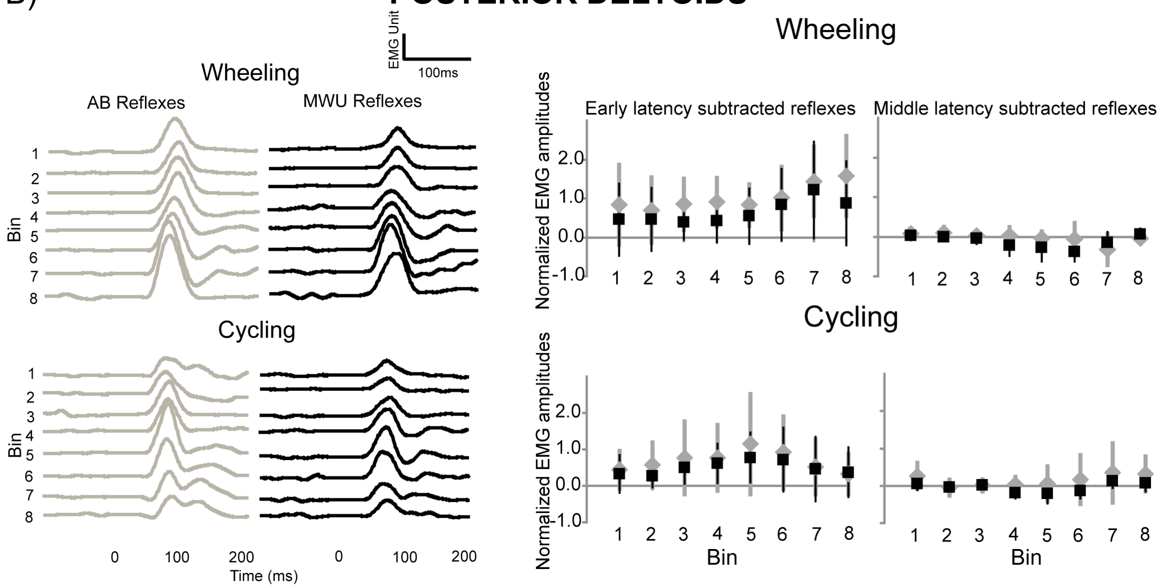
ECR demonstrated greatest reflex responses during bins 1-3 (Figure 3.7B). Like the FCR, the ECR reflex responses were larger in the AB group. Early and middle latency reflexes were excitatory

Figure 3.5 Anterior and posterior deltoids reflex modulation plots in able-bodied subjects and manual wheelchair users

A) **ANTERIOR DELTOIDS**



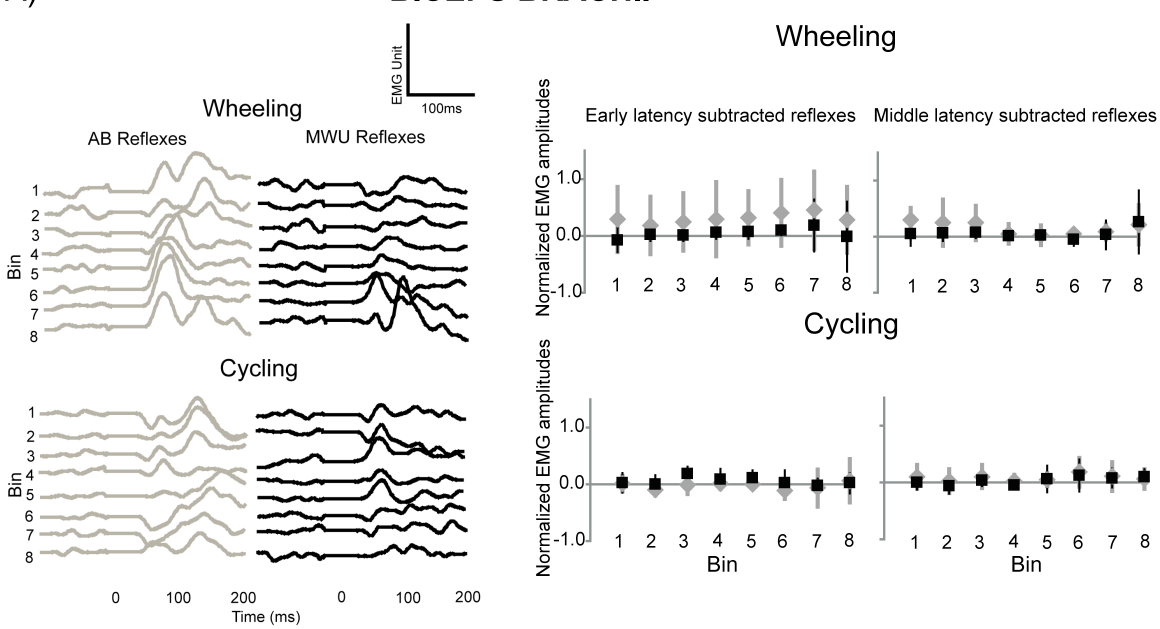
B) **POSTERIOR DELTOIDS**



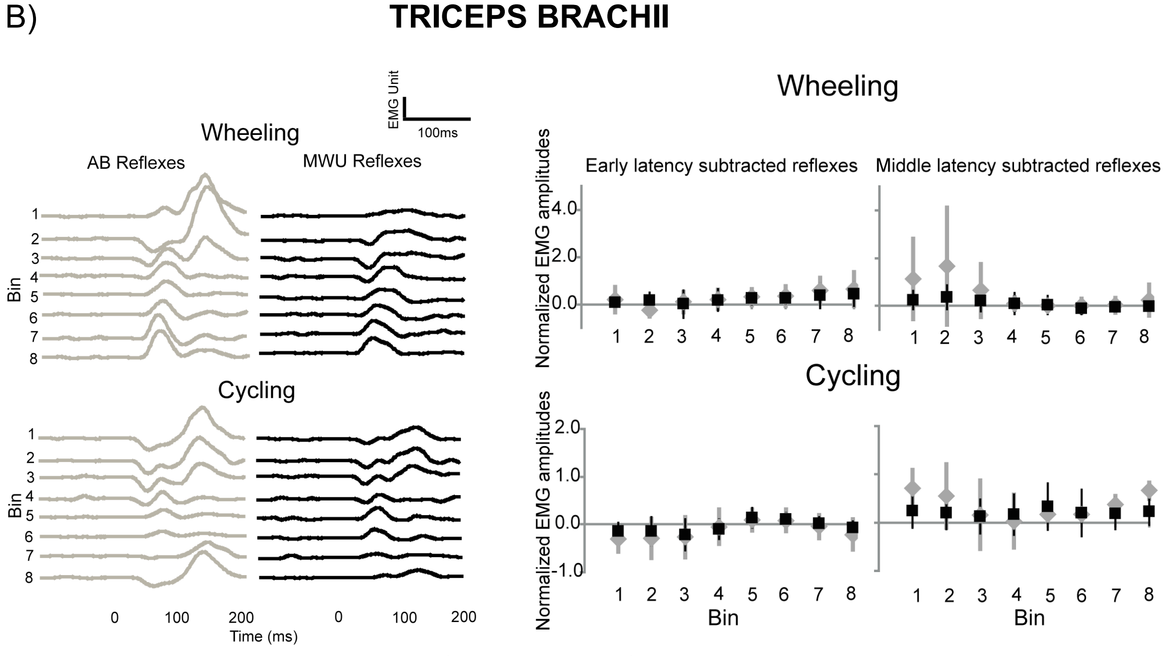
This diagram represents the reflexes and reflex modulation in the AD and PD muscles. The graphs to the left show the normalized averaged group reflexes across the eight bins for the AB group (grey) and MWU (black). The stimulation occurs at 0ms and the early and middle latency reflexes occur at 50-75ms and 75-120ms respectively. The black vertical line represents one arbitrary EMG unit. The plots to the right represent the reflex amplitude across the eight bins during early (left) and middle (right) latencies. The black squares represent the MWUs group and the grey diamonds represent the AB group.

Figure 3.6 Biceps and triceps brachii reflex modulation plots in able-bodied subjects and manual wheelchair users

A)

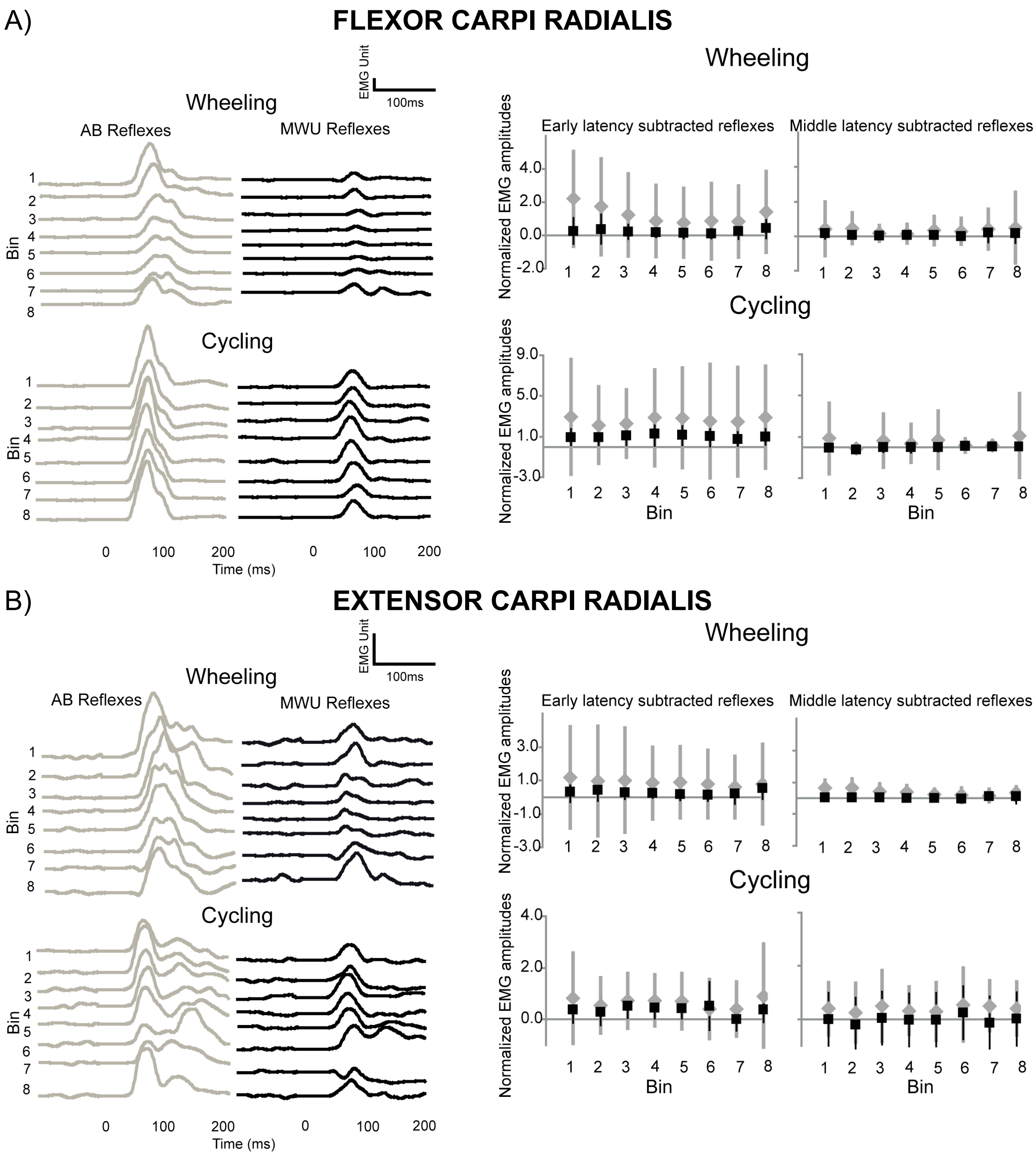


B)



This diagram represents the reflexes and reflex modulation in the BB and TB muscles. The graphs to the left show the normalized averaged group reflexes across the eight bins for the AB group (grey) and MWU (black). The stimulation occurs at 0ms and the early and middle latency reflexes occur at 50-75ms and 75-120ms respectively. The black vertical line represents one arbitrary EMG unit. The plots to the right represent the reflex amplitude across the eight bins during early (left) and middle (right) latencies. The black squares represent the MWUs group and the grey diamonds represent the AB group.

Figure 3.7 Flexor and extensor carpi radialis reflex modulation plots in able-bodied subjects and manual wheelchair users



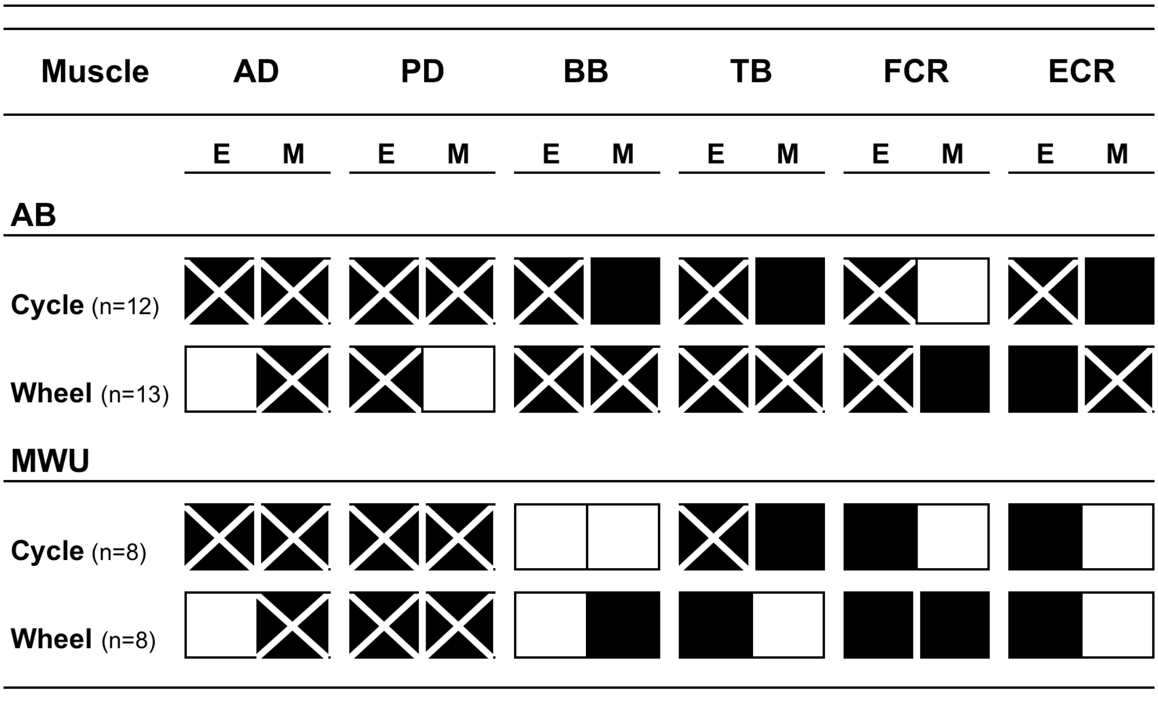
This diagram represents the reflexes and reflex modulation in the FCR and ECR muscles. The graphs to the left show the normalized averaged group reflexes across the eight bins for the AB group (grey) and MWU (black). The stimulation occurs at 0ms and the early and middle latency reflexes occur at 50-75ms and 75-120ms respectively. The black vertical line represents one arbitrary EMG unit. The plots to the right represent the reflex amplitude across the eight bins during early (left) and middle (right) latencies. The black squares represent the MWUs group and the grey diamonds represent the AB group.

3.3.4 Reflexes and modulation

The majority of muscles showed an early or middle latency reflex response (Figure 3.8). There were 2 latencies evaluated for each of the six muscles for a total of 12 combinations of muscle latencies. According to the Wilcoxon signed-ranks test, the AB group showed significant reflexes in 11/12 of muscle latencies during cycling and 10/12 during wheeling ($p<0.05$). This is contrasted to the 8/12 of significant reflex responses observed in the MWUs group during both cycling and wheeling ($p<0.05$).

Significant reflex modulation, according to the Friedman's test for multiple comparisons, occurred in 8/12 of the muscle latencies in the AB group during both wheeling and cycling ($p<0.05$, Figure 3.8). The MWUs group demonstrated significant modulation during cycling in 5/12 of the muscle latencies during cycling and 3/12 during wheeling. Figure 3.8 shows that the AB group showed more significant reflexes responses and reflex modulation for both cycling and wheeling compared to the MWUs group.

Figure 3.8 Muscles exhibiting significant early and middle latency reflexes and phase-dependent reflex modulation



The black boxes represent the muscles and latencies where significant reflex modulation occurred ($p<0.05$). The black boxes with white x's represent the muscles which also demonstrated significant reflex modulation ($p<0.05$). The white boxes represent muscles that did not show significant reflexes or reflex modulation.

3.3.5 Correlations between background EMG and reflexes

The majority of reflex responses were not significantly correlated with EMG_{BG} (Table 3.3). The only muscle that showed significant correlation between EMG_{SUB} and EMG_{BG} was the AD (early latency response $r_S = -0.70$; middle latency response $r_S = 0.59$) in the AB group during cycling ($p<0.05$).

Table 3.3 Spearman's rank correlation: EMG_{SUB} and EMG_{BG}

Muscle	AD		PD		BB		TB		FCR		ECR	
	E	M	E	M	E	M	E	M	E	M	E	M
AB												
Cycle (n=12)	**-0.70**	**0.59**	0.06	0.15	-0.40	0.24	-0.48	0.07	0.07	-0.27	0.22	-0.27
Wheel (n=11)	-0.45	0.48	0.37	-0.27	-0.44	0.08	-0.22	0.26	0.06	-0.13	-0.29	-0.10
SCI												
Cycle (n=8)	-0.71	0.34	-0.37	-0.19	-0.33	0.26	-0.31	-0.02	0.27	-0.30	0.10	-0.48
Wheel (n=8)	-0.35	0.35	0.25	-0.50	-0.14	0.02	-0.08	0.22	-0.27	0.13	-0.13	-0.39

Significant Spearman r-values at the level of $P<0.05$ are denoted in bold.

3.3.6 Task dependent reflex modulation

In the AB group, there were significant differences between tasks for the amount of reflex modulation in the middle latency reflex response in the PD (Kruskal Wallis H=4.275, 1 d.f., $p<0.05$) and the early latency response of the TB (Kruskal Wallis H=5.861, 1 d.f., $p<0.05$). Both the middle latency of the PD as well as the early latency of the TB demonstrated a greater MI during wheeling compared to arm cycling. In the MWUs group, only the modulation of the early latency of the TB muscle showed significant differences between wheeling and cycling (Kruskal Wallis H=4.206, 1 d.f., $p<0.05$). The early latency of the TB in the MWUs group demonstrated a greater MI during wheeling compared to arm cycling.

3.3.7 Experience and reflex modulation

There were no significant differences in the amount of reflex modulation that occurred between the AB and MWUs groups for either latency in any of the muscles. However within the MWUs group, there were significant correlations between years of experience and MI for some muscles (Table 3.4). There was a significant correlation between MI of the middle latency reflex response of the AD, observed in the cycling task, and years of manual wheeling experience (r_s=0.79, p<0.05). There was also a significant correlation between the MI of both tasks, during early latency of the TB, and years of manual wheeling experience (r_s=0.74 for both tasks, p<0.05).

Table 3.4 Spearman's rank correlation: years of manual wheeling experience and modulation index

Muscle	AD		PD		BB		TB		FCR		ECR	
	E	M	E	M	E	M	E	M	E	M	E	M
SCI												
Cycle (n=8)	0.60	**0.79**	0.02	-0.21	-0.36	0.21	**0.74**	0.36	0.50	0.14	-0.46	-0.07
Wheel (n=8)	-0.02	0.64	0.36	0.24	0.38	0.55	**0.74**	0.60	0.19	0.02	0.02	0.19

Significant Spearman r-values at the level of P<0.05 are denoted in bold.

3.4 Discussion

This study compared the cutaneous reflex response to stimulation of the superficial radial nerve during arm cycling and manual wheeling in able-bodied and spinal cord injured MWUs. In the majority of muscles studied, cutaneous reflex responses to stimulation of the superficial radial nerve during manual wheeling and arm cycling were modulated according to the phase of the movement cycle. Larger suppressive and facilitation responses were evoked in most muscles during the ‘transition to propulsion’ and ‘propulsion’ phases of the wheeling cycle.

The MI in most muscles did not differ between tasks. In the AB group, there was significantly more modulation during wheeling compared to arm cycling in the middle latency response of the PD and the early latency response of the TB. In the MWUs group, there was significantly more modulation during wheeling compared to arm cycling in the early latency of the TB. Lastly, the amount of reflex modulation did not differ between the AB and MWU groups, however there was a significant correlation between years of manual wheeling experience and the quantity of modulation during arm cycling during the middle latency of the AD and early latency of the TB and during wheeling in the early latency of the TB.

3.4.1 Phase-dependent modulation of reflexes during manual wheeling

Manual wheeling versus gait

The desired outcome of locomotion, whether by walking or manual wheeling is functional mobility. Both walking and manual wheeling have two distinct phases of the movement cycle including contact and no contact. Contact corresponds with propulsion in manual wheeling and stance in walking whereas no contact corresponds with recovery in manual wheeling and swing in walking. The major difference between these two tasks is that walking is an asymmetrical movement while manual wheeling is a symmetrical movement. However, since they are both rhythmic movements that serve the same functional purpose, it could be expected that the reflex patterns are similar between the two tasks.

When stimulating the superficial peroneal nerve during walking, there is a significant suppression of the tibialis anterior during swing, corresponding with a reduction in dorsiflexion, while the biceps femoris and vastus lateralis demonstrate an excitation during swing corresponding with knee flexion (Zehr, Komiyama et al. 1997). This response is known as the stumbling corrective response, which is very similar to that observed when encountering a mechanical perturbation and is an example of a phase-dependant reflex response (Forssberg 1979; Schillings, Van Wezel et al. 1996; Zehr, Komiyama et al. 1997). If we compare the leg to the arm, the knee extensor function of vastus lateralis is analogous to the elbow extensor function of the TB. Similarly, the biceps femoris as a knee flexor has its upper limb analogue in BB as an elbow flexor. When stimulating the superficial radial nerve, which is considered to be analogous to the superficial peroneal nerve, we see an increase in reflex response in the BB during recovery and an increase in reflex response in the TB during the end of recovery and transition to propulsion. This is somewhat analogous to what is observed in walking, thus an analog of the ‘stumbling corrective response’ could also be observed in the upper limbs. If the dorsal aspect of the hand is obstructed by a perturbation during recovery (swing), it appears that there is an excitation response towards elbow flexion occurring in the BB during early latency as well as shoulder extension occurring in the PD during early latency. It is of interest to note that the characteristics of an analogous ‘stumbling corrective response’ observed in the upper limbs are facilitated during early latency inferring that this response may be mediated at the level of the spinal cord. In addition, Zehr et al (1997) found smaller kinematic changes during stance and larger changes during swing which is generally what we observed in this study (Zehr, Komiyama et al. 1997). The critical phases exhibiting greater changes in reflex modulation appear to occur during the transition into propulsion occurring at approximately bins 1-2 and 7-8 of the wheeling cycle). This may have the functional importance of maintaining stability allowing the progression of movement.

Asymmetrical versus symmetrical rhythmic activity

Despite the functional relevance of the symmetrical rhythmic movement of manual wheeling, the majority of studies exploring phase dependent cutaneous reflex modulation in the upper limbs have examined arm cycling 180 degrees out of phase (asymmetrical movement) (Carroll, Zehr et al. 2005). One study examined the difference between in-phase arm cycling (symmetrical) and 180 degree out of phase arm cycling (asymmetrical) and found that the cutaneous reflex responses were quite similar (Carroll, Zehr et al. 2005). The reflex responses differed only in the early latency components in the posterior deltoids, extensor carpi radialis and the early and middle responses of the anterior deltoids (Carroll, Zehr et al. 2005). These results suggest that the relative phasing of contralateral limbs has minor effect on the sensory modulation of rhythmic movements.

Asymmetrical arm cycling has been more commonly used in studies examining rhythmic arm movement so that the results can be easily compared to walking, however, symmetrical movements are also commonly found in nature. Rabbits, kangaroos and frogs are only a few animals that exhibit the symmetrical movement of hopping. A study examining the effects of a SCI at the lower thoracic vertebrae in rabbits found that upon stimulation, an alternating (asymmetrical) walking movement occurred unless the animal was in the correct posture and the soles of the feet were in contact with the ground (Ten Cate 1963). They concluded that the spinal cord was capable of generating both symmetrical and asymmetrical (alternating) movement, but that supraspinal centers of the brain stem controlled the posture, which could not be achieved after the injury without assistance from the researchers. This study may allude to the idea that the most simplistic movement generated by the spinal cord may be that of asymmetrical (alternating) movement, however depending on the specific sensory feedback (e.g. from appropriate posturing), spinal cord locomotor circuits may be modulated to generate symmetrical movement.

Differences in background EMG

Both arm cycling and manual wheeling demonstrated distinct rhythmic EMG bursting. There were distinct bursting patterns between agonist and antagonist muscles and EMG activity was rhythmically modulated throughout the wheeling cycle. For both the arm cycling and the manual wheeling tasks, the AB and MWU groups showed very similar EMG profiles and kinematics for the specific task. It was expected that these variables would be similar between groups for cycling because arm cycling is a fixed movement and all individuals had intact function of their arms. Therefore all individuals performed the same movement. It was uncertain if the background EMG would look similar between the two groups for manual wheeling because the recovery phase of the movement is not fixed and therefore could lead to different EMG patterns or kinematics (see Chapter 2), however the overall EMG patterns and kinematics were also very similar between the two groups indicating that they did very similar movements. From this observation, we can infer that any differences in reflexes between the AB and MWU groups were not due to significant differences in movement within a task.

Phase dependent modulation and background EMG

Over the past decade researchers have explored cutaneous reflexes during asynchronous arm cycling (Zehr and Chua 2000; Zehr and Kido 2001; Zehr and Hundza 2005). An earlier study found that for some muscles, cutaneous reflexes were modulated with the movement cycle and were strongly correlated to the amplitude of the EMG_{BG} while more than half of the muscles studied showed phase-dependent reflex modulation (Zehr and Kido 2001). However, in the current study, it is notable that reflex responses were not correlated with background EMG for any of the muscles and latencies for either the arm cycling or manual wheeling tasks among subjects in the MWU group. However, subjects in the AB group showed a significant correlation between EMG_{SUB} and EMG_{BG} in the early and middle latency responses of the AD during arm cycling, which is consistent with previous findings (Zehr and Chua 2000) (Zehr and Kido 2001).

<u>3.4.2 Differences in reflex responses between wheeling and cycling</u>

Differences in kinematics and background EMG

There were differences in the kinematics between the two tasks. During manual wheeling, both groups achieved greater shoulder extension reaching behind the midline of the body, but did not go as far into shoulder flexion compared to the arm cycling task. During manual wheeling greater elbow flexion was achieved compared to arm cycling, however greater elbow extension was achieved during arm cycling compared to manual wheeling. The wrist angles were not comparable because during arm cycling the wrist undergoes flexion and extension because it is holding the handle, however during manual wheeling the wrist primarily undergoes ulnar and radial deviation. These differences in kinematics are likely due to the different postures between the two tasks. During arm cycling the arms must resist gravity and are held with the shoulder in approximately 90 degrees of flexion while during manual wheeling the arms hang beside the body and are at a more natural posture, similar to arm swing during gait.

There were differences in EMG profiles between tasks in some muscles. Rather than a single burst in the later phase of the movement cycle observed during manual wheeling, the PD shows a non-bursting pattern during arm cycling. This could be related to the observation that during manual wheeling, the shoulder moves from extension through flexion and back to extension, however, in arm cycling, the shoulder remains in a fixed posture. The TB exhibits a burst during both propulsion and recovery phases of the wheeling cycle corresponding with forearm extension during propulsion and shoulder extension during recovery. During arm cycling the TB exhibits one burst corresponding with elbow extension. Lastly the FCR exhibits an early burst and then a drop during manual wheeling corresponding with radial deviation and then ulnar deviation of the wrist. During arm cycling the EMG profile demonstrated a burst during the middle of the movement. The remainder of the muscles studied exhibited similar muscle activation between tasks. In addition, the EMG_{bg} muscle activation patterns during arm cycling were similar to those reported in previous studies (Zehr and Chua 2000; Zehr and Kido 2001).

Task-specific reflex modulation

Zehr and colleagues found that in the reflex response in the AD muscle during arm cycling, there was an early latency suppression followed by a middle latency facilitation and in the PD muscle there was an early latency facilitation (Zehr and Chua 2000). This same trend was observed in our study in both the arm cycling and manual wheeling condition. In some cases we also observed a small middle latency inhibitory response in the PD reflex response. It was also previously reported that the BB and TB demonstrated an early latency suppression and a middle latency facilitation (Zehr and Chua 2000). This pattern was also observed in our study for the arm cycling trials, but somewhat more sporadically during wheeling. The pattern of early-latency suppression and middle latency facilitation may suggest that the reflex responses in these muscles could be mediated by supraspinal centers rather than spinal control during both the arm cycling and manual wheeling tasks, but perhaps more so for arm cycling.

In the AB group, there were differences in the amount of reflex modulation between tasks in the middle latency reflex response of the PD (facilitative) and the early latency reflex response in the TB (suppressive). Both of these reflexes demonstrated a significantly larger MI during manual wheeling compared to arm cycling in the AB group. Interestingly, the TB and PD have similar function in extending the arm at the shoulder joint in both tasks. However, in the MWUs group, the only task-specific difference was in the early latency reflex response in the TB (suppressive). The early latency TB reflex response also demonstrated a significantly larger MI during manual wheeling compared to arm cycling in the MWUs group. The significant modulation of PD in the AB group but not in the MWUs group may point to the possibility that the PD may be used differently in individuals not experienced in manual wheeling although both groups have similar patterns for wheeling. This could indicate that there is a different level of neural drive to the PD that cannot be discerned by the basic EMG activation pattern. On the other hand, the TB appears to have a critical role in both experienced and inexperienced MWUs during manual wheeling compared to cycling.

3.4.3 Effect of manual wheeling experience on reflex modulation

There were no significant differences in the MI between the groups. However, upon further analysis in the MWUs group relating the years of wheeling experience versus MI, the MI of the middle latency response of the AD and the early latency response of the TB were significantly correlated with years of manual wheeling experience. The early latency response of the TB occurring during arm cycling was also significantly correlated with years of manual wheeling experience. Interestingly, there was also a difference in MI of the early latency response in the TB between wheeling and cycling in the MWUs group (see previous section). The muscles showing significant correlation with MI and years of manual wheeling experience were the muscles that demonstrated the early latency suppressive response and the middle latency facilitative response. The TB plays a unique role during wheeling in that there is approximately equal muscle activation during both the propulsion and recovery phases of the cycle. It appears that the TB plays a critical role in manual wheeling and perhaps that role evolves with manual wheeling experience.

To date, there has been very little research that has looked at the effect of training or experience on changes in cutaneous reflex response. The acute changes are well documented in many rhythmic tasks, however it is uncertain how cutaneous reflexes change with training or development of a skill such as manual wheeling. Previous literature has demonstrated differences in H-reflexes between trained versus untrained ballet dancers. Nielson and colleagues found that the Hmax/Mmax ratio was larger in moderately and well-trained dancers compared to untrained subjects, however the ratio was smaller in professional ballet dancers (Nielsen, Crone et al. 1993). It was suggested that these changes may be a result of changes in the excitability of the motoneuronal pool or the transmission across the Ia synapses. Earlier research proposed that the H-reflex decreased among highly anaerobically trained athletes (Rochcongar, Dassonville et al. 1979). This explanation does not fit for the results we see in this study because the muscles in the upper limbs are largely used for aerobic activities including wheeling. However, the reduction in H-reflexes among the ballet dancers could be a result of the large amount of time performing the skill compared to the less trained ballet dancers.

One study examined the effect of step training on cutaneous reflexes in cats. The cats were spinalized and then trained to perform stepping movements (Cote and Gossard 2004). Cote and colleagues found that the non-trained control group exhibited exaggerated reflexes with large amplitudes after spinalization however, the cats trained in stepping after spinalization showed a decrease in reflexes (Cote and Gossard 2004). This indicates that with training, normalization of cutaneous pathways may be associated with recovery of movement. Despite the lack of literature in this area, there is some compelling evidence to support the plasticity of cutaneous pathways and the ability to modify these reflexes to aid in movement.

3.5 Conclusion

Overall, the majority of muscles demonstrated significant reflex responses during both tasks. These reflexes were not generally correlated with background EMG. The TB and AD appear to play a unique role during manual wheeling. They exhibit an early latency suppressive response and a middle latency facilitative response. The early latency response of the TB differed between tasks and was also significantly correlated with the MI and years of manual wheeling experience. The TB seems to play a critical role during both propulsion and recovery phases of the manual wheeling cycle.

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4 GENERAL DISCUSSION

4.1 Summary

Currently very little literature has explored the multiple facets of the biomechanics and neural control of manual wheeling, despite the large incidence of overuse injuries and the need to better understand optimum wheeling strategies. Many studies have observed manual wheeling from a strictly kinematic and kinetic perspective, however specific roles of the muscles and the nervous system in producing this movement have been neglected. The purpose of this thesis was to provide a comprehensive examination of the biomechanics as well as the neural control of manual wheeling. To our knowledge, no previous research has examined the neural control of manual wheeling, however a great amount of effort has been made in understanding the neural control of walking and other rhythmic tasks.

In Chapter 2 of this thesis, the kinetics, kinematics and muscle activation patterns, as well as manual wheeling strategies, used by experienced manual wheelchair users (MWUs) and able-bodied (AB) subjects were examined. The overall goal of this study was to determine if there were differences in the biomechanical characteristics between the two groups. Many studies use AB subjects because they are easier to recruit, however it is uncertain if this is representative of experienced MWUs. The majority of biomechanical characteristics were similar between the two groups, however the characteristics that differed including the average forces, wrist ROM, push angle, and percent of the wheeling cycle spent in propulsion could play an important role in effective manual wheeling and aid in the prevention of injury.

In Chapter 3, the sensory modulation of muscle activity patterns during manual wheeling was studied by using the technique of cutaneous nerve stimulation. Reflex responses from cutaneous nerve stimulation during manual wheeling were compared to those elicited during arm cycling and differences in reflexes between MWUs and AB groups were evaluated. Interestingly, manual wheeling shared many of the same trends observed during gait as well as arm cycling in terms of reflex modulation.

4.2 Future directions

<u>4.2.1 Biomechanics of manual wheeling</u>

There has been an on-going debate about whether inexperienced AB subjects should be used when exploring manual wheeling strategies and whether those results can be applied to the manual wheelchair user population. This thesis examined the difference between AB subjects who had not previously used a manual wheelchair, and experienced MWUs with spinal cord injuries at T1 and below.

Wheeling strategies: MWUs versus AB

Using a chi-squared test of independence, wheeling strategy was found to be dependent on manual wheeling experience ($p<0.01$). This indicates that AB individuals and experienced MWUs will use different wheeling strategies. Therefore, in studies analyzing wheeling strategy, subject recruitment should involve experienced MWUs so that the results are externally valid. This study only examined experienced manual wheelchair users with spinal cord injuries at T1 and below, consequently it is uncertain if these results can be applied to other populations who use manual wheelchairs (e.g. tetraplegics). Future studies should compare across multiple populations who use wheelchairs for locomotion, as well as AB individuals who are experienced MWUs.

There is no consensus in the literature as to which manual wheeling strategy is most efficient and which should be taught by therapists in rehabilitation centers (Consortium for Spinal Cord Medicine, 2005). The most efficient wheeling strategy likely depends on the functional mobility of the MWU and therefore may change according to lesion level or reason for using a manual wheelchair. Previous studies have provided recommendations for wheeling strategies (Shimada, Robertson et al. 1998; Boninger, Souza et al. 2002), however these studies have provided conflicting results.

Future studies should also examine wheeling strategies at different times post injury. It is unknown if wheeling strategies and biomechanical characteristics change over time with different amounts of experience. An inclusion criterion for this study was that all subjects must be a minimum of one-year post injury; in our study the individuals who participated in this study ranged from 2-29 years post injury. Therefore it is possible that some of the subjects may change their wheeling strategies with additional experience or skills training. However, it is likely that these factors would occur on an individual basis and may depend on personal experience.

Kinematic and kinetic differences in manual wheeling

In this study, significant differences were found in average resultant force and average tangential force as well as percentage of the wheeling cycle spent in propulsion and push angle between the two groups ($p<0.05$). The experienced MWUs group demonstrated larger average forces as well as more time spent in propulsion and a greater push angle. It is intuitive that the average forces were higher in the MWUs group because they had spent a greater percentage of the wheeling cycling in propulsion rather than the recovery phase and had been in contact with a greater amount of the pushrim resulting in a larger push angle. Greater percentage of time spent in propulsion phase is thought to increase mechanical efficiency because more work is done to propel the wheelchair forward per cycle and the cadence is often reduced (Boninger, Souza et al. 2002). From these results we can infer that spinal cord injured MWUs with lesions at T1 and below would display similar characteristics, however we cannot make the assumption that this is true for all experienced manual wheelchair users.

There were some striking differences in joint kinematics. Significant differences were found in wrist range of motion (ROM), peak wrist radial deviation, and peak elbow flexion between AB and MWU groups ($p<0.05$). The increased range of motion of the wrist is likely developed with manual wheeling experience and improvements in overall wheeling technique including strategy. Increased wrist ROM has been shown to be associated with better nerve conduction, less force applied to the pushrim and fewer strokes required to propel a wheelchair a specific distance (Boninger, Impink et al. 2004). Therefore there appears to be a high level of functional importance in having an increased range of motion of the wrist joint. Impaired wrist range of motion would greatly decrease push angle unless other joints compensated for the lack of wrist movement. Consequently, it is very important that nerve health is maintained so that debilitating conditions such as carpel tunnel syndrome can be avoided.

There appear to be some very important differences that should be taken into consideration when selecting an AB versus manual wheeling population. The differences in this study may be key in understanding efficient wheeling strategies. The wrist range of motion likely has great functional importance and may be a limiting factor in wheeling efficiency. Future research should focus on the impact of wrist ROM on manual wheeling efficiency.

The results of this study cannot necessarily be applied to other manual wheeling populations who do not have full function of their upper limbs and therefore further research should examine wheeling among different populations. An effort has been made to establish normative data for a larger population of MWUs in the SmartWheel Users' Group database (Cowan, Boninger et al. 2008). As the database develops and increases in size, it will likely provide normative manual wheeling data for individuals with different lesion levels. The information in the database is collected through the use of a SmartWheel (Three Rivers Holdings, Mesa, AZ, USA) which is an expensive device not commonly used in Canada. Therefore it may be difficult for many people who do not use a SmartWheel to find normative values relating to manual wheeling biomechanics.

Clinicians and therapists can collect important variables through observation, these variables can include an estimation of push frequency, percentage of wheeling cycle spent in propulsion, push angle, velocity, and wheeling strategy.

4.2.2 Neural control of manual wheeling

Previous research has examined cutaneous reflex modulation during walking and arm cycling, however to our knowledge no previous research has explored cutaneous reflex modulation during manual wheeling. This thesis examined the cutaneous reflex response during arm cycling and manual wheeling in AB and spinal cord injured MWUs. The goal was to explore cutaneous reflex modulation during manual wheeling and to examine potential differences between manual wheeling and upper limb cycling tasks as well as potential differences between experienced MWUs and a novice control group.

Reflex modulation during manual wheeling

In the majority of muscles studied, cutaneous reflex responses to stimulation of the superficial radial nerve were modulated according to the phase of the movement cycle during both arm cycling and manual wheeling. Larger reflex responses, including suppressive and facilitative responses, were evoked in most muscles during the ‘transition to propulsion’ phases of the wheeling cycle. This is similar to what is observed in walking (Zehr, Komiyama et al. 1997). This study only observed reflex responses on the side of the body ipsilateral to the stimulation. Therefore it is unknown how neural control is provided to the contralateral arm during manual wheeling. It is possible that MWUs develop strong interlimb coordination through the continuous use of their upper limbs for symmetrical movement and that strong interlimb reflex responses could be expected.

Task specificity

In the AB group, there was more reflex modulation during wheeling compared to arm cycling in the middle latency response of the posterior deltoids (PD) and the early latency response of the triceps brachii (TB). In the MWU group, there was more modulation during wheeling compared to arm cycling only in the early latency component of the TB reflex response. It has been proposed that manual wheeling may elicit a larger amount of reflex modulation because of the functional importance of manual wheeling to MWUs. However, considering the theoretical framework of this project, all upper limb rhythmic movement should have the same general level of neural control (Zehr, 2005).

Although arm cycling may be considered to be a less functionally important task compared to wheeling, it is capable of producing locomotion. Murkherjee and colleagues found that asymmetrical arm cycling propelled devices are significantly less physiologically demanding compared to manual wheelichairs (Mukherjee and Samanta 2001). Arm cycle mobility devices can be either symmetrical or asymmetrical, however it has been argued that asymmetrical mobility devices are more efficient than symmetrical mobility devices (Glaser, Sawka et al. 1980). However, it appears that many commercially available arm cycling mobility devices in Canada require symmetrical cycling. Future research should explore differences in neural control between individuals skilled in asymmetrical arm cycling and asymmetrical arm cycling mobility devices. Understanding the neural control of these movements may help in the development of future technologies.

Reflexes and manual wheeling experience

The quantity of reflex modulation did not differ between the AB and MWU groups, however there was a significant correlation between years of manual wheeling experience and the quantity of modulation during arm cycling during the middle latency of the AD and during wheeling in the early and middle latencies of the TB. It is uncertain if any differences or lack of differences between groups is a result of manual wheeling experience or if it is a result of the changes in the nervous system due to the SCI. Future studies should include a third group involving AB subjects who are experienced MWUs (e.g. AB wheelchair sport athletes). There is no current human literature on the effect of training on the cutaneous reflex response. A study examined the effect of step training in cats. Cote and colleagues found that the non-trained control group exhibited exaggerated reflexes with large amplitudes after SCI however, the cats trained in stepping after their SCI showed a decrease in reflexes (Cote and Gossard 2004). MWUs could be considered highly trained in manual wheeling. There are many aspects of training that may influence reflex responses during a rhythmic movement. In Cote and colleagues' study it is unknown whether it was the stepping movement or if it was the sensory feedback provided from touching the ground which influenced the cutaneous reflex response. Ideally a third 'air stepping' group could have been used to explore the effect of training without the sensory feedback from the paws hitting the ground.

This indicates that with training, normalization of cutaneous pathways may be required to recover the movement of stepping after SCI. Despite the lack of literature in this area, there is compelling evidence to support the plasticity of cutaneous reflexes and the ability to modify these reflexes to aid in movement. Although we attempted to look for an effect of experience, we could not explicitly look for an effect of training. Future research could explore changes in cutaneous reflexes in the upper limbs post injury and one year after using a manual wheelchair or by providing a period of manual wheelchair training to a group of AB individuals.

There are many areas surrounding manual wheeling that need to be further studied to provide a more in-depth understanding of manual wheeling. Over half of people who use manual wheelchairs will develop shoulder pain and potentially shoulder injury throughout their lifetimes (Lal 1998; Curtis, Drysdale et al. 1999; Dalyan, Cardenas et al. 1999). Therefore efforts should be made in understanding efficient manual wheeling strategies and how the nervous system helps to regulate and control the cyclical movement of manual wheeling.

4.3 References

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APPENDIX 1

Means and standard deviations for averaged variables for each group.

Variable	Able-bodied Group	Manual wheelchair users
Kinematics (degrees)		
Wrist ROM*	**43.0 (7.2)**	**57.0 (11.1)**
Elbow ROM	51.3 (7.2)	47.2 (14.5)
Shoulder ROM	64.4 (6.0)	64.1 (11.9)
Wrist peak radial deviation*	**35.0 (10.3)**	**50.6 (10.8)**
Elbow peak flexion*	**99.5 (4.5)**	**112.7 (10.5)**
Shoulder peak flexion	21.8 (7.6)	17.9 (8.9)
Wrist peak ulnar deviation	-8.1 (6.7)	-6.3 (6.3)
Elbow peak extension	150.8 (6.9)	158.7 (8.1)
Shoulder peak extension	-42.6 (5.3)	-46.3 (7.6)
Kinetics		
Push Angle* (degrees)	**56.8 (8.8)**	**78.4 (15.3)**
Percent time spent prop.*	**42.0 (6.1)**	**50.5 (9.9)**
Cadence (cycles/sec)	0.7 (0.0)	0.7 (0.0)
Velocity (m/s)	0.75 (0.1)	0.82 (0.3)
Force resultant max (N)	45.2 (12.3)	34.3 (9.9)
Force tangential max (N)	34.3 (9.9)	34.1 (15.8)
Force resultant avg.* (N)	**11.3 (1.3)**	**14.9 (4.2)**
Force tangential avg.* (N)	**7.6 (1.3)**	**9.7 (4.1)**
EMG (ms)		
Time to peak AD	14.5 (13.5)	15.1 (5.1)
Time to peak PD	84.6 (3.3)	81.9 (8.8)
Time to peak BB	18.0 (17.9)	39.1 (45.1)
Time to peak TB	36.1 (24.0)	58.0 (32.0)
Time to peak FCR	13.3 (13.3)	22.1 (31.5)
Time to peak ECR	33.5 (28.5)	39.2 (43.6)