METABOLIC DEMANDS AND EMG PROFILES OF BWS TREADMILL WALKING IN PERSONS WITH SCI

INFLUENCE OF BWSTT FOR INDIVIDUALS WITH INCOMPLETE SCI: METABOLIC DEMANDS AND EMG PROFILES

By

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ABSTRACT

Body weight supported treadmill training (BWSTT) is being promoted as an effective means of restoring ambulatory abilities among individuals with incomplete spinal cord injuries. The emphasis of this thesis is on the description of the metabolic demands and the EMG profiles of able-bodied persons and individuals with a spinal cord injury (SCI) while walking under the identical conditions on a body weight support (BWS) treadmill. The secondary purpose was to contrast the metabolic and muscular responses between the two groups. Two separate chapters describing the metabolic demands and EMG profiles respectively follow the review of the literature. The metabolic results indicate that raising the speed and/or decreasing the amount of BWS increase the intensity of BWS treadmill walking, with speed having a more profound effect. The SCI group was less efficient and they had greater metabolic rates of oxygen consumption than the controls for all conditions examined. This led to the conclusion that walking on the treadmill, for the SCI group can provide an effective aerobic exercise stimulus. The EMG profiles suggest that speed and BWS affect the phasic characteristics of the muscular activity while walking for both groups. Furthermore, abnormalities, omissions and inappropriate levels of activity were observed in the SCI group when compared to the controls. These irregularities suggest that the SCI participants have adopted altered motor strategies while walking, relative to the control group. Nonetheless, the SCI participants showed evidence of appropriate modulations in their EMG activity to meet the demands of the task as they changed from one condition to the next.

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LIST OF ABBREVIATIONS

ACE	Arm Crank Ergometry
ANOVA	Analysis of Variance
ASIA	American Spinal Injury Association
BWS	Body Weight Support
BWSTT	Body Weight Support Treadmill Training
EA	Ensemble Average
EMG	Electromyography
FES	Functional Electrical Stimulation
FWB	Full Weight Bearing
FWR	Full Wave Rectification
GA	Gastrocnemius
GEA	Grand Ensemble Average
KAFO	Knee Ankle Foot Orthosis
KE	Kinetic Energy
LCE	Leg Crank Ergometry
LE	Linear Envelope
MG	Medial Gastrocnemius
PE	Potential Energy
RGO	Reciprocating Gait Orthosis
RHC	Right Heel Contact
RMG	Right Medial Gastrocnemius
RPE	Rating of Perceived Exertion
RTA	Right Tibialis Anterior
SCI	Spinal Cord Injury
TA	Tibialis Anterior
VO2	Metabolic Rate of Oxygen Consumption

1.0 REVIEW OF LITERATURE

1.1 Introduction

Walking is by far one of the most fundamental forms of human locomotion and when performed abnormally or inefficiently, the impact on activities of daily living can be devastating. The ability to walk requires a healthy cardiovascular system in order to oxygenate and supply blood to the working muscles, an intact nervous system to effectively recruit and coordinate the locomotor musculature, as well as good peripheral muscular strength capable of supporting the body's weight and performing the mechanical work of walking. Healthy humans are relatively efficient at bipedal locomotion, with up to 65% of the mechanical energy required for locomotion being recovered from stride to stride through the transformation of potential to kinetic energy and vice versa (Saiben, 1990). A different picture emerges however, when studying the locomotor patterns of individuals with gait abnormalities. Extraneous movements in the vertical and lateral planes, poor coordination and timing of movements all result in the inability to conserve energy, thus, increasing the muscular and metabolic demands of locomotion (Fisher & Gullickson, 1978; Waters & Mulroy, 1999).

Due to the importance of walking for day-to-day living, gait has been studied extensively to determine what the causes are of various gait abnormalities, whether a treatment has produced and/or restored normal functioning and/or what alternative strategies an individual has adopted to achieve locomotion (Grieve, 1969). Age, disease states, orthopaedic injuries, acute trauma and various neuropathies can all result in the loss of ambulatory function. Consequently, much research has been conducted on developing effective rehabilitation techniques, orthotics and/or assistive devices to help restore functional and economical ambulation for those who have lost the ability to walk efficiently (Bunc and Dlouha, 1977).

Spinal cord injuries (SCI) are one of the most catastrophic forms of acute neurological trauma. Although the severity of these injuries varies greatly from person to person, one of the most devastating outcomes is usually the loss of the ability to walk due to paralysis of the lower extremity musculature. One of the newest forms of rehabilitation employed to restore locomotor function in individuals with SCI, is body weight support treadmill training (BWSTT), which is being researched all over the world (Barbeau & Rossignol, 1994).

This relatively novel form of gait rehabilitation involves suspending an individual above a moving treadmill in a modified parachute harness in order to support a proportion of their bodyweight, while still allowing full range of motion of the lower extremities for walking. In general, a BWSTT regime is interactive and progressive, similar to any exercise program, where increases in intensity and duration are implemented in a gradual manner. More specifically, it has been presumed that increasing the intensity of a BWS training program can be accomplished by either reducing body weight support (BWS) and/or increasing the speed of the treadmill (Gardner, Holden, Leikauskas, & Richard, 1998; Protas et al., 2001 & Wernig & Muler, 1992). Most patients will start a BWSTT program at very slow speeds (less than 1km/hr) with fifty or more percent BWS. As the training progresses, the speed is gradually increased and BWS is progressively reduced, in an attempt to wean the individual off the BWS altogether and restore independent walking over ground (Wernig, Muller, Nanassy, & Cagol, 1995; Wernig, Nanassy, Muller, & Cagol, 1998). In general, during the initial stages of training, individuals with SCI require physical assistance from therapists to move their legs through a normal gait pattern. It has also been reported that with sufficient training, some patients with incomplete SCI regain the ability to take independent steps (i.e. no physical assistance from a therapist to move their legs) on the treadmill, while fewer have exhibited the ability to ambulate over ground without the use of any assistive devices (Wernig et al., 1995 & Wernig et al., 1998).

Based on work done to date, it has been argued that BWSTT provides an ideal stimulus for gait rehabilitation in individuals with SCI (Barbeau & Rossignol, 1994; Schindle, Forstner, Kern & Hesse, 2000; Wernig & Muller, 1992; Wernig et al., 1995, & 1998). However, the scientific literature is limited in its presentation of the activity patterns of the locomotor musculature while individuals with SCI ambulate on a BWS treadmill. Furthermore, no study has systematically examined the muscular activity patterns of able-bodied individuals ambulating on a BWS treadmill in order to make an objective comparison between the two groups. Consequently, it is impossible to quantify and/or describe the normalcy of the dynamic EMG profiles of individuals with incomplete SCI who have received extensive BWSTT. Therefore, the scientific literature is in need of a study that describes and compares various metabolic and muscular activity patterns of individuals with SCI who have received extensive BWSTT to that of a control group comprised of able-bodied individuals, who are free from gait abnormalities.

1.2 Metabolic Demands and Cost of Walking in Able-bodied Individuals

In order for metabolic energy expenditures to have any inherent value, the data must be normalized in order to make comparisons and draw conclusions regarding the energy expenditure of an activity between subjects and/or conditions. Therefore, metabolic rate refers to the oxygen consumption per unit time, which is often further normalized to bodyweight in kilograms to allow for comparisons between individuals (ml/kg/min). Although rates of oxygen consumption are useful for comparing the metabolic demand of one task to another, this metric does not allow for determining the efficiency at which an activity is being performed. Thus, the metabolic cost of walking is generally used as the standard for drawing conclusions regarding the efficiency of gait in both healthy individuals and those with gait abnormalities (Bunc & Dlouha, 1997; & Mattson, 1989). Metabolic cost is determined by normalizing the rate of oxygen consumption to the amount of external work performed by taking the rate of oxygen consumption (ml/kg/min) and dividing by speed (m/min) to get an oxygen cost value expressed in milliliters oxygen per kilogram bodyweight per meter walked (ml/kg/m). Consequently, as the metabolic cost of an activity increases, greater inefficiencies are indicated (Waters & Mulroy, 1999).

1.2.1 Rate of oxygen consumption – speed relationship

It has been well established that the rate of oxygen consumption increases in a curvilinear fashion as speed increases (Blessey, Hislop, Waters, & Antonelli 1976; Blessey, 1978; Fisher and Gullickson, 1978; Mattson, 1989; Waters and Mulroy, 1999). Consequently, many second order regression equations have been developed to predict the metabolic rate of walking for any given speed (equations 1-3, in Table 1.1). Upon examination of the oxygen consumption – speed relationship, it is clear that rates of ambulation are accompanied by very subtle and gradual increases in oxygen consumption over the comfortable range of locomotion 60 – 100 m/min (Waters, Lunsford, Perry, & Byrd, 1988). These equations were developed using the range of customary walking speeds and are not intended for extremely slow speeds (i.e. < 40 m/min). Nevertheless, at very slow speeds, the metabolic demands tend to level off with an average rate of oxygen consumption around 5.7 ml/kg/min (Waters & Mulroy, 1999).

Table 1.1: Equations for predicting the metabolic rate of oxygen for walking at various speeds. Where "S" is the measure of speed in m/min and the estimated rates are in ml/kg/min. (adapted from Waters & Mulroy, 1999)

Equation	Author	
	Second Order Regression Equations	
1	Ralston (1971)	O_2 Rate = $0.0011\ S^2 + 5.9$
2	Corcoran & Gelmann (1970)	O_2 Rate = $0.001\ S^2 + 6.2$
3	Molen & Roxendal (1967)	O_2 Rate = $0.00105\ S^2 + 7.1$

1.2.2 Oxygen cost – speed relationship

When individuals freely select their speed of locomotion, they tend to naturally find the most efficient speed, by means of reducing the metabolic demands per meter walked (Fisher and Gullickson, 1978; & Waters and Mulroy, 1999). In order to examine the effect of speed on oxygen cost, researchers have imposed unnatural speeds of locomotion on their subjects. Controlled speeds of locomotion have been shown to affect the efficiency of gait because participants are forced to walk using unnatural stride frequencies, stride lengths and/or any combination thereof (Bunc & Dlouha, 1997; Holt, Hamill, & Andres, 1991 & Waters & Mulroy, 1999). The oxygen cost – speed relationship while walking over ground is best described as slightly "U" shaped, with the greatest efficiency occurring near 80 m/min (4.8 km/hr), which is generally consistent with most peoples' self-selected comfortable walking speed (Waters & Mulroy, 1999). Thus, speeds less and greater than this result in poorer efficiency. Bunc and Dlouha (1997), on the other hand, reported that the most efficient speed of walking on a treadmill occurs at 66 m/min (~ 4 km/hr), with the metabolic cost also increasing exponentially as speed is either increased or decreased.

Walking at higher speeds result in inefficiency because the rate of oxygen uptake increases disproportionately to increases in speed. At the other end of the spectrum, unnaturally slow gait patterns, especially those less than 40 m/min, create instability, resulting in a loss of equilibrium (Holt et al., 1991). Workman and Armstrong (1986) also emphasized the maintenance of balance as having a major impact on the metabolic cost of slow walking. At slow speeds, the normally smooth mechanics of locomotion are disrupted and there is an increase in muscle co-activation and inefficient movements, which could account, at least in part, for the increase in metabolic cost. Longer stride lengths have also been observed during slow gait (Bunc & Dlouha, 1997), causing greater vertical displacement of the body's centre of mass, thus increasing the mechanical work performed and driving up the metabolic cost (Minetti, Capelli, Zamparo, di Prampero, & Saibene., 1995). Beillot et al. (1996) argue that during slow walking, a greater proportion of the gait cycle is spent in metabolically demanding phases of gait (i.e. stance and double limb support), which may also help account for the higher metabolic cost of slow walking speeds.

1.2.3 Comfortable walking speeds

In general, when asked to walk at a comfortable speed, able-bodied adults will ambulate somewhere between 60 and 100 m/min with the average being 80 m/min (Waters et al., 1988), with the metabolic cost ranging between 0.14 – 0.16 ml/kg/m (Waters & Mulroy, 1999). Natural walking speed has also been found to vary significantly between the genders (Fisher & Gullickson, 1978 & Waters and Mulroy 1999); with men walking faster than women. However, Blessey et al., (1976) and Blessey (1978) did not find a significant difference between the naturally selected walking speeds between men and women.

Table 1.2 summarizes the results from several studies examining the naturally self-selected speeds of adult locomotion (20-59 year old men and women combined) and their associated oxygen rates and costs. From this, it can be readily determined that the naturally selected speed of locomotion varies slightly, but approximates 77.2 m/min (4.6 km/hr). More importantly however is that the oxygen cost remains relatively constant from one study to the next, despite the variations in speed of locomotion. Thus, although the comfortable walking speed of healthy adults may vary, the rate of oxygen consumption fluctuates appropriately with changes in speed, keeping the metabolic cost relatively constant and in the order of magnitude of 0.16 ml/kg/m.

Table 1.2: Summary of self-selected comfortable walking speeds among healthy adults and their associated oxygen rate and metabolic cost.

Author	**Speed (m/min)**	**O_2 Rate (ml / kg / min)**	**O_2 Cost (ml / kg / m)**
Petrofsky & Smith (1991)	67	10.9	0.16
Ralston (1971)	74	11.95	0.16
Mattson (1989)	76.8	12.29	0.16
Watters et al., (1988)	80	12.1	0.15
Blessey, (1978)	82	12.8	0.16
Fisher & Gullickson (1978)	83	13.0	0.16
Mean	77.2	12.2	0.16

1.2.4 Effect of BWS on oxygen demands of walking in able-bodied persons

It has been determined that body weight is directly related to the rate of oxygen consumption while walking, hence the need to normalize energy demands to body weight (Fisher & Gullickson, 1978 & Waters and Mulroy, 1999). For example, obese individuals have been observed to have significantly higher metabolic demands than that of leaner individuals when controlling for speed, otherwise their self selected speed is significantly less than normal, presumably to maximize efficiency (Waters & Mulroy, 1999).

Artificial loads, such as a 20 kg weight carried over the trunk of a male subject did not appreciably increase the energy demand of walking nor did the self-selected speed of walking decrease (Waters and Mulroy, 1999). However, Pandolf et al., (1978) discovered that carrying a load equivalent to 40% body weight resulted in an equivalent increase in the rate of oxygen consumption. Additionally, oxygen uptake has also been observed to increase by as much as 30% when small loads (i.e. 2 kg) are fastened around the ankles. The impressive increases in oxygen consumption observed due to small peripheral loads as compared to nominal changes when bigger loads are carried on the torso have been attributed to the larger acceleration and decelerations of the limbs (Waters & Mulroy, 1999).

Most of the research examining the effects of unloading the body on metabolic demands comes from studies investigating the oxygen cost during exercise under simulated reduced gravity in order to draw conclusions about exercise and space travel. In a study conducted by Fox, Bartels, Chaloupka, Klinzing, and Hoche (1975), sub-gravity was accomplished via horizontal centrifugation and by means of an inclined plane. The horizontal centrifugation technique required that the subject's long axis be parallel to the floor, while their feet were placed on the wall by suspending the subject from the ceiling in slings. Effectively, in this position the force of gravity on the feet is zero. By varying the speed of the rotating room, an unlimited number of sub-gravity conditions could be created.

The speed versus metabolic demand versus gravity relationship was examined extensively in Fox et al.'s (1975), study and it was determined that sub-gravities ranging from 0.5G down to 0.25G resulted in significant decreases in oxygen consumption while walking at all speeds examined (16.7 – 133 m/min). For example, while walking at 66.7 m/min, the metabolic cost for walking at 1G using the centrifugation technique was 0.20

ml/kg/m, whereas at 0.5G and 0.25G the metabolic cost of walking at the same rate of speed were 0.13 ml/kg/m and 0.09 ml/kg/m respectively, both of which were significantly less than the metabolic cost at 1G.

Fox et al.'s (1975) study indicates that, as the body is un-weighted, the metabolic demand of walking decreases. An interaction effect between speed and un-weighting was also observed with the metabolic cost of walking increasing with speed, but more at 0.5G than at 0.25G. Fox et al., rationalized the significant reduction in oxygen consumption to the smaller loads borne by the lower extremities when un-weighted. Griffin et al., (1999) expanded upon Fox et al.'s explanation, indicating that reduced body weight was associated with proportionately reduced ground reaction and muscle forces.

In a study examining the efficacy of harness-suported treadmill walking for knee rehabilitation, Colby, Kirkendall, and Bruzga (1999) determined that supporting a proportion of an individual's bodyweight at a controlled speed, significantly reduced the rate of oxygen consumption. More specifically, the metabolic costs of walking on the treadmill at 80 m/min with 20 and 40% BWS were 0.17, and 0.16 ml/kg/m, respectively, both of which were significantly less than the full weight bearing condition (0.18 ml/kg/m). Danielsson and Sunnerhagen, 2000, also observed a significant reduction in VO2 in able-bodied and hemiparetic stroke patients while walking at a self-selected speed with 30% BWS.

From the evidence above it can be concluded that the metabolic demands of walking are affected by a person's weight, and the amount of loading. Reduced body weight via simulated reduced gravity or BWS has been shown to significantly decrease VO2. It appears that the most plausible explanation for the observed decreases in metabolic demands during unloading is the reduction of forces, but controlled studies investigating the mechanism have yet to be completed.

1.3 Metabolic Demands and Cost of Walking in Persons With a Spinal Cord Injury

Similar to the able-bodied population, individuals with gait abnormalities tend to self-select speeds of ambulation that minimize the metabolic cost of walking, which are often much slower than that of the healthy population (Mattson, 1989; Fisher and Gullickson1978; & Waters and Mulroy, 1999). However, for many individuals with a pathological gait, the metabolic cost of locomotion still remains very high, resulting in quick exhaustion and/or non-functional speeds of walking. In fact, walking activities performed by individuals with pathological gait can take several minutes longer than that for a normal individual at metabolic rates several times higher than that of the healthy population (Waters & Mulroy, 1999).

Pathological gait abnormalities are extremely varied in the SCI population. While some individuals with SCI can walk independently with no assistance others may require ambulatory aids or orthotics for locomotion. Further, many people with SCI are more catastrophically impaired and may be rendered wheelchair dependent for the remainder of their lives. To date extensive amounts of research have examined the energetics of walking in individuals with SCI before and after conventional therapies, and with various types and combinations of orthotics and assistive devices. On the other hand, far fewer studies have examined the metabolic demands for those receiving or who have received BWSTT.

Typically individuals with complete paraplegia will require the use of bilateral knee-ankle-foot orthoses (KAFO's) in order to achieve any degree of functional walking. The KAFO's serve to stabilize the individual's lower extremities while the individual uses forearm-supported crutches to bear the weight of the body with their upper extremities while swinging the lower extremities forward. This form of gait is commonly referred to as a "swing-through gait" (Waters & Mulroy, 1999) and is typically very metabolically demanding. Fatigue occurs quite quickly because the upper limbs produce the majority of the locomotor function.

Individuals with SCI using KAFO's can achieve speeds of 29 m/min with oxygen rates as high as 16.3 ml/kg/min. The ensuing cost can be determined to be 2.5 times greater than that for a able-bodied individual walking at the same speed (0.56 ml/kg/m Vs 0.22 ml/kg/m) (Waters & Mulroy, 1999). After calculating and comparing the metabolic cost of this activity to the energy demands of wheelchair use (0.18 ml/kg/m at 75 m/min), it should come as no surprise that most individuals with a SCI resort to using a wheelchair for locomotion in lieu of using their orthotics and aids (Cerny, 1978 & Rosman & Spira, 1974).

In a study by Cerny in 1978, energy cost comparisons were made between customary SCI walkers and customary wheelchair users. During a free-velocity walking test, the customary walkers achieved higher speeds of ambulation than the non-customary walkers with metabolic costs of 0.36 ml/kg/m compared to 1.28 ml/kg/m for the wheelchair group. In a follow up study conducted by Cerny, Waters, Hislop, and Perry (1980), similar conclusions were drawn. Individuals using KAFO's were found to have a significantly higher metabolic cost than that found during wheelchair locomotion, yet those who customarily walked were still found to be more efficient than those who did not. Furthermore, Cerny et al., concluded that wheelchair propulsion is an efficient mode of transportation for paraplegics because the metabolic cost and corresponding speed of locomotion are approximately equal to that for walking in able-bodied individuals.

Clinkingbeard, Gersten, and Hoehn (1964) conducted an extensive study examining the metabolic cost of ambulation in nineteen people with traumatic paraplegia using KAFO's, and attempted to compare energy expenditure to the duration of training and the level of the lesion. Interestingly, a similar speed – oxygen uptake relationship existed for the SCI group as was observed in able-bodied individuals. However, the maximum speed achieved by one subject was 29 m/min (1.7 km/hr) whereas the group average velocity was 11.3 m/min (0.66 km/hr), both of which are much less than that found in the normal population and that cited above for wheelchair use. In addition to their pathologically slow gait, the average oxygen cost for walking in this population was calculated to be 1.27 ml/kg/m, also suggesting a very inefficient form of locomotion.

One of the more interesting outcomes from Clinkingbeard et al.'s (1964) study was that significant increases in speed and improvements in both metabolic rate and cost are possible within three weeks of ambulatory training. Anecdotally, Clinkingbeard et al. (1964), attributed the changes in efficiency to increases in skill and coordination achieved during walking training. Unfortunately, once the training stopped, most patients discontinued their regular ambulatory training bouts and returned to their pre-exercise state.

A more recent and now more common walking device used by individuals with SCI, is the reciprocating gait orthosis (RGO's) (Beillot, et al., 1996 & Bowker, Messenger, Ogilvie, & Rowley, 1992). Unlike the swing through gait used when wearing KAFO's, a reciprocating gait orthosis is a mechanical assistive device, which allows for a more natural reciprocal gait pattern to be achieved (Bowker et al., 1992).

Bowker et al. (1992) examined the energetics of paraplegic walking using RGO's using the Physiological Cost Index (PCI = walking HR – Resting HR / walking speed). The mean velocity achieved by the traumatic paraplegics examined using RGO's was 16.1 $\pm$ 10.7 m/min with a PCI of 5.6 $\pm$ 3.6 beats/m. From this, Bowker et al (1992) concluded that the RGO did not provided an efficient or functional means of ambulation. In fact, comparisons to able-bodied individuals revealed that the RGO was only 1/25 as efficient as walking and 1/7 as efficient as wheelchair use

In a more comprehensive study conducted by Beillot et al., (1996), the metabolic cost of walking using an RGO with and without the use of functional electrical stimulation (FES) was conducted on 14 individuals with spastic, complete paraplegia with lesion levels varying between T2 –T12. After having completed 12 weeks of training, the average speed of walking was 13.8 m/min and the metabolic cost of walking was found to be 1.97, 0.99, and 0.90 ml/kg/m while walking at 6, 12, and 18 m/min respectively. When the FES was used, the metabolic cost was found to be 2.05, 1.16 and 1.03 ml/kg/m while walking at the identical speeds. Therefore, energy cost of ambulation decreased as speed increased in both conditions, yet the use of FES appeared to have a negative effect on walking efficiency.

Protas and associates (2001) are the only researchers to use oxygen uptake measures (ml/kg/min) in individuals with incomplete SCI while walking over ground pre and post BWSTT. This pilot study included three participants with incomplete SCI (2 ASIA D, 1 ASIA C) who completed three months of training, consisting of five, 20-minute training sessions per week. Testing was conducted before training began and every three weeks thereafter. Metabolic measures were taken while the subjects ambulated as quickly as possible for five-minutes back and forth on a 5 meter walkway while using any of their customary assistive devices (e.g. walker, cane, crutches, etc.) and/or orthotics.

During the baseline measurements, the average speed of walking was calculated to be 4.06 m/min (0.24 km/hr) while the metabolic demand and cost were 7.96 ml/kg/min and 1.96 ml/kg/m respectively. After 12 weeks of training, the subjects were able to ambulate faster, with an average speed to 12.7 m/min (1.15 km/hr) with the metabolic demands increasing to 17.15 ml/kg/min yet the metabolic cost was significantly reduced to 1.35 ml/kg/m. From these results it is apparent that the SCI participants became more efficient with training and Protas et al. concluded that BWSTT is an effective rehabilitation strategy for improving gait efficiency in individuals with incomplete SCI with only three months of therapy.

From all of the above results, it is apparent that the metabolic demands of individuals with SCI, while ambulating with orthoses, are much greater than those of able-bodied individuals. Ambulation with assistive devices can be achieved only at slower speeds and at higher rates of oxygen consumption than that identified for wheelchair use. However, there is some evidence to suggest that improvements in metabolic cost are possible with training and that BWSTT can also provide these individuals with the appropriate stimulus to achieve faster speeds of ambulation and better metabolic efficiency.

1.4 Phasic Muscular Activity Patterns While Walking in Able-bodied Individuals

While walking, the musculature of the lower limbs is cyclically active in order to produce rhythmical moments of force about the hip, knee and ankle joints. The moments of force must be appropriately sized and timed in order for the gait pattern to be fluid, coordinated and efficient. Thus, any pathology, or injury affecting the appropriate phasic and/or activation of the primary gait or postural control musculature will have a dramatic effect on the efficiency of gait. Gait analysis is essential in order to determine the cause of gait abnormalities, develop appropriate treatments and/or evaluate the effectiveness of assistive devices and rehabilitation programs (Grieve, 1969).

Among the many tools used by gait researches, electromyography (EMG) is a common choice because it provides a means to examine and quantify the dynamic activity of skeletal muscle. This rather unique and sophisticated tool provides a graphical display of the electrical activity occurring within a muscle, which is both time and amplitude sensitive and provides valuable information unavailable to simple observation. The graphical representation of muscle activity, while walking for example, can be used to determine what muscles are doing, or not doing, to contribute to the gait deviations observed and/or measured with other instrumentation (i.e. footswitches, motion, etc). In fact, EMG recordings are commonly used for making surgical decisions to correct gait abnormalities (Delisa, 1998).

When recording EMG activity during gait, it is of fundamental importance to have some way of determining the phases of the gait cycle, otherwise the EMG signal is meaningless (Delisa, 1998). Phasing of gait allows researchers to determine the onset and cessation of muscle activity, as well as times of peak activation with regards to the gait cycle. Therefore, properly collected EMG data that has been accurately cycled allows researchers to precisely define the muscle actions that control joint movements, and interpret their functional role.

For the purposes of this literature review, the movement, function and EMG activity of the musculature across the ankle joint while walking will be discussed. More specifically, the focus will be isolated to the gastrocnemius (GA) and tibialis anterior (TA).

The GA is situated in the posterior compartment of the leg and is comprised of two halves, which are anatomically described as medial (inner) and lateral (outer). In general, their functions are identical, serving to plantar flex (extend) the ankle joint. While walking, the GA exhibits major activity during mid-stance leading into toe-off. The activity generally starts at approximately 15-20% of the gait cycle and terminates at approximately 50–60% of the stride (Arsenault, Winter, & Marteniuk, 1986a; Inman, Ralston, & Todd, 1981; Murray, Mollinger, Gardner, & Sepic 1984; Nilsson,

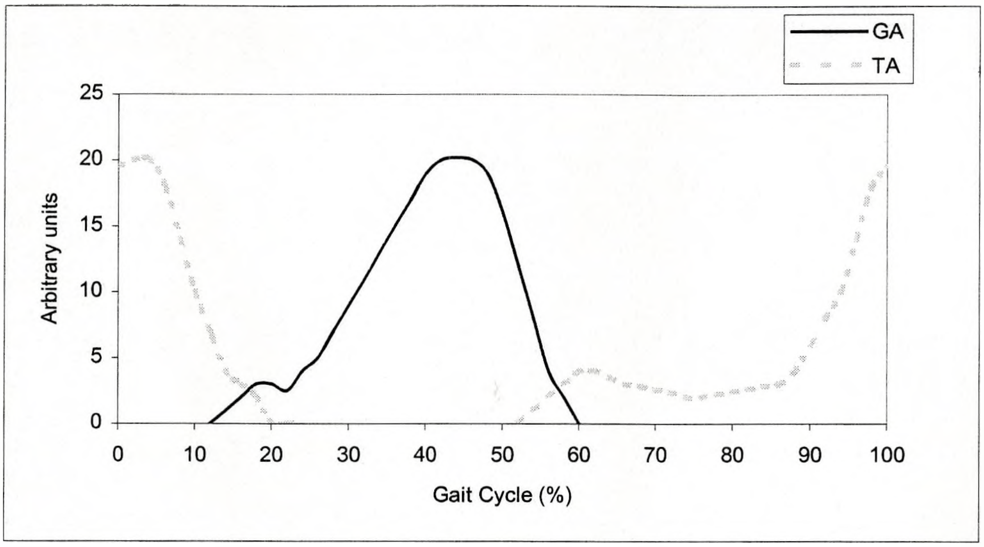

Figure 1 1 Schematic diagram of the phasic muscular activity of the gastrocnemius (GA, solid line) and tibialis anterior (TA, broken line), over an entire gait cylce. 0% corresponds with heal strike and toe-off occurs at ~60% of the gait cycle. Adapted from Inman et al., (1981).

Thorstensson & Halbertsma, 1985 & Winter & Yack, 1987). The muscular activity that occurs initially during mid-stance is the result of an eccentric contraction due to the GA restraining/controlling the amount of ankle dorsiflexion as the leg and body rotates over the fixed foot. Later in stance, the contraction of the GA eventually becomes concentric in nature at the transition point where the ankle begins to plantarflex and the heel is raised off the ground (Inman et al., 1981, Kameyama, Ogawa, R., Okamoto, T., & Kumamoto, 1990 & Murray et al., 1966) (See Figure 1 1). Ultimately, the GA is responsible for shifting weight to the fore part of the foot and assisting with forward and upward propulsion of the body, and is said to be a major power generator during the gait cycle (Murray, Guten, Sepic, Gardner, & Baldwin, 1978). Consequently, the GA is commonly examined during EMG gait analysis due to its readily observed phasic activity and important functional role while walking (Murray et al., 1978).

The TA is situated in the anterior compartment of the leg, just lateral to the crest of the tibia and serves to dorsiflex (flex) the ankle and is antagonistic to the GA. The normal activity of the TA while walking is biphasic (See Figure 1.1). The first peak of activity occurs during weight acceptance, at heel strike (0 – ~20% of the gait cycle), whereas the second burst generally occurs in early swing (~60% of the gait cycle), during and/or just after toe-off and remains moderately active throughout swing while rising up again to a peak near the end of the gait cycle (100%) (Inman et al., 1981). The first burst is primarily due to the eccentric activity of the TA as it decelerates and lowers the foot to the ground to prevent foot drop and absorb the shock of heel strike (Kameyama et al., 1990). During the stance phase of the gait cycle the TA is generally silent while the second EMG burst begins eccentrically as the heel is raised (i.e. the ankle is plantar flexed). The TA activity then becomes concentric in order to dorsiflex the ankle during the initial portion of the swing phase to provide toe-clearance and prevent toe-drag. Finally, the activity of the TA reaches its second peak of activity at the end of the gait cycle in preparation for the subsequent heel strike (Inman et al., 1981; Murray et al., 1966; Murray et al., 1978).

It should be noted that both the phase and amplitude of the ambulatory musculature are almost identical with those seen at comparable speeds of walking on a treadmill and walkway. Moreover, the stride-to-stride variability was found to be smaller while walking on a treadmill than over ground (Arsenault, Winter & Marteniuk, 1986b). Similarly, the results from Murray, Spurr, Sepic, Gardner, & Mollinger (1985) indicate that treadmill walking kinematics and EMG profiles are not markedly different from those observed while walking over ground. Therefore, treadmills do not greatly affect gait patterns or muscular outputs and consequently have been readily used for performing gait analysis in both able-bodied individuals and those with gait abnormalities.

1.4.1 Effect of speed on the EMG patterns in able-bodied individuals

Changes in the speed of walking have been shown to cause significant changes in various temporal and distance gait parameters such as stride length, duration, cadence, and gait kinematics (Murray et al., 1966; Murray et al., 1984; & Nilsson et al., 1985). Slower speeds are often associated with shorter strides, over a longer duration, whereas faster speeds are associated with an increase in stride frequency, longer stride lengths and shorter stride durations (Murray et al. 1966). Shorter stride lengths are associated with less hip flexion of the leading limb during heel contact, and less hip extension accompanied by less ankle plantar-flexion of the trailing limb, compared with longer strides (Murray et al. 1984). In addition, Detrembleur, Willems, and Plaghki (1997) indicated that faster speeds were associated with increases in the proportion of time spent in swing phase, and subsequently reduced time spent in stance. This result has been found to be consistent across ages and gender.

Despite speed related kinematic and temporal changes to the stride, the peaks and valleys of the EMG patterns remain tightly coupled to stride events such as heel strike and toe-off (Yang and Winter, 1985). Other researchers agree with Yang and Winter that the phasic activity of the TA and soleus remains consistent across all speeds (Finch et al., 1991; Kameyama et al. 1990; Milner, Basamajian, & Quanbury, 1971 & Winter 1983).

On the other hand, many non-classical EMG patterns have been observed to occur while able-bodied individuals walk at different speeds (Shiavi, Champion, Freeman, & Griffin, 1981). Thus, some researchers argue that walking EMG patterns do change as the speed of locomotion increases (Detrembleur et al., 1997; Murray et al. 1984; & Wooten, Kadaba, & Cochran, 1990 & Shiavi, Bugle & Limbird, 1987). For example, Murray et al. (1984) reported that the phasic activity of the calf musculature activity was prolonged toward the end of stance while walking faster. This fact was argued to be functionally necessary in order to help propel the body and swinging limb forward with greater force. Additionally, Shiavi et al. (1981) reported that TA activity became monophasic while walking slowly (1.55 km/hr) and multiphasic at faster speeds (5 km/hr).

Unlike the controversy over the effects of speed on the phasic nature of EMG while walking, it has been well established that changes in speed have a more profound effect on EMG amplitude. More specifically, EMG amplitude increases as the speed increases (Brandell, 1977; Kameyama et al. 1990; Milner et al. 1971; Murray et al., 1984; Yang & Winter 1985 & Winter 1983). Greater EMG amplitudes at higher speeds are most likely due to the increase in the muscle forces required to accelerate and decelerate the limbs in shorter periods of time (Yang & Winter 1985). Another concern with slow walking speeds is that greater EMG variability is observed between subjects due to diverse coping strategies (e.g. different cadences, stride lengths, etc.) employed by different people (Shiavi et al. 1981 & Yang & Winter, 1995).

In conclusion, it would appear that changes in speed have more consistent effects on EMG amplitudes than on the phasic characteristics of EMG across the stride.

Therefore, if comparing the phasic nature of EMG between individuals, comparable speeds should be used in order to eliminate phasic dissimilarities caused by speed.

1.4.2 Effect of BWS on the EMG patterns in able-bodied individuals

The unloading effects, via body weight support, on the phasic and/or amplitude of locomotor EMG patterns have not been as readily examined as the effects of speed. However, the rationale for such studies can be easily justified, given the extent of studies examining the effect of BWS on individuals with pathological gait abnormalities, especially those with SCI. Furthermore, it is realistic to assume that unloading the body would have an effect on EMG output especially for the plantar flexors. Given that the calf musculature is primarily used for propulsion during push-off and must overcome body weight in order to accelerate the body upward and forward, one would expect that an increase in BWS should reduce the amplitude of the EMG activity of this muscle group due to reductions in the ground reaction forces.

Finch et al. (1991) conducted a comprehensive study examining the effects of both speed and BWS on EMG characteristics. Their study involved walking on a treadmill at progressively slower speeds as BWS was increased. Their justification for reducing the speed of ambulation as BWS increased came from their pilot study that indicated that comfortable walking speeds decreased as the level of BWS increased for able-bodied persons. The dependent measures during each of the BWS conditions were compared to that of a full weight-bearing (FWB) condition while walking at the identical speeds. This comparison allowed for the independent effect of BWS to be determined.

Furthermore, the design of this study also allowed for an examination of the effects of speed on various gait parameters while walking at full weight. However, comparisons between BWS conditions were not possible because speed and BWS were varied together. In other words, the difference in EMG or gait patterns between 30 and 50% BWS could not be reasonably compared because both speed and BWS had changed. Thus, the interaction effect, if any, between speed and BWS could not be determined.

The levels of BWS used were 0 (i.e. FWB), 30, 50 and 70%, while the speed of walking varied from 90 m/min (5.4 km/hr) for the 0% BWS condition to 39 m/min (2.34 km/hr) when 70% BWS was employed. EMG activity was collected from the vastus lateralis, medial hamstrings, tibialis anterior, and the medial gastrocnemius. Determination of various temporal gait characteristics was achieved by placing 3 footswitches on the sole of the foot. Joint angle data was also determined via videotape (60 frames/sec).

Increasing speed was discovered to significantly reduce cycle time, whereas increases in BWS were not observed to have a main effect on cycle time. Hip and knee angle displacement patterns and/or amplitudes were not observed to be greatly affected by walking speed. On the other hand BWS was observed to significantly decrease the amplitude of both hip and knee maximum swing angles. It was speculated that the observed decrease in angular displacement during mid-stance was due to the harness constraining vertical movement, thus preventing their bodies from descending during stance. The increases in BWS were also speculated to reduce the magnitude of push-off and limit the momentum and displacement of the swinging limb.

The on-off timing of the medial gastrocnemius (MG) and tibialis anterior (TA) activity was not significantly affected by speed across the full weight-bearing conditions, whereas BWS moderately affected the timing of both. In general the MG activity was turned on sooner during stance yet terminated prematurely. Likewise, the TA activity was also observed to activate sooner but also terminated later during swing, thus causing an overall increase in the TA burst durations.

Unlike other studies (Brandell, 1977; Kameyama et al. 1990; Milner et al. 1971; Murray et al., 1984; Yang & Winter 1985) that found an increase in EMG amplitude as speed increased, the mean amplitude of the TA and MG were not significantly affected by decreases in speed in Finch et al.'s study (1990). The comparisons between the FWB and BWS conditions revealed that TA activity was significantly higher as BWS increased while MG activity was significantly reduced. Colby et al. (1999) also reported reductions in gastrocnemius, hamstrings and quadriceps activity as BWS was increased. The reductions in MG activity is readily explained because less weight was borne during stance when the level of BWS was higher, subsequently reducing the forces required to raise the heel and body, in preparation for push-off. The increase in TA activity during swing was not rationalized.

Probably the most important conclusion drawn from the studies conducted by Finch et al. (1991) and Colby et al. (1999) was that BWS does not significantly alter, or impair gait muscle activation patterns, yet some irregularities were observed. Moreover, these results suggest that ambulation on a BWS treadmill should allow individuals with gait pathologies to progress more easily from stance to swing. These conclusions are of great importance for those intending to use BWS treadmills for therapeutic applications. Consequently, both groups of researchers indicate that BWS treadmill walking would be a valuable aid in the rehabilitation for those with gait abnormalities. However, these results may also suggest that comparisons of EMG patterns between individuals with gait disturbances to those of control subjects should be conducted under similar, if not identical levels of BWS and speed.

1.5 Phasic Muscular Activity Patterns While Walking in Persons with SCI

Very few studies have been able to examine the voluntary EMG activity of the lower limbs of individuals with SCI while walking due to the level of paralysis suffered from this type of injury. Nevertheless, some literature has examined upper body musculature activity via EMG while ambulating with crutches (Chantraine and Onkelinx, 1975), while other more recent studies have examined lower extremity EMG patterns of individuals with SCI while walking with the use of assistive devices, while wearing orthoses and while ambulating on body weight support treadmills (Dietz et al 1994; Dietz, 1995; Dietz et al., 1997; Dietz, Nakazawa, Wirz, & Ernie 1999; & McKay, Metman, Dimitrijevic, Sherwood, & Dimitrijevic, 1993).

Lower extremity surface EMG was recorded bilaterally from the quadriceps, hamstrings, tibialis anterior, and the triceps surae on sixteen subjects with incomplete SCI, all with varying walking abilities by McKay et al. (1993). The subjects were evaluated while ambulating on a 5m path, while footswitches were used to synchronize the EMG to the gait cycle. The subjects walked at a self-selected, comfortable speed, while using their usual assistive devices (i.e. walker, cane, crutches, orthotics or any combination thereof). When compared to a group of five controls the discrepancies in self-selected speed of walking become apparent. The controls had a mean stride time of 1.1 seconds, compared to a mean of 1.4 seconds for the three fastest SCI ambulators whereas, the moderately-slow (6 subjects) and very-slow ambulators (7 subjects) had stride times ranging from 2.0-3.2 & 3.4-8.0 seconds respectively. Furthermore, large discrepancies in gait symmetry were observed among all SCI participants, as well as substantially shorter swing phases, longer stance and longer double limb support phases compared with the control group.

In terms of locomotor EMG patterns for the triceps surae, numerous abnormal phasic patterns were observed, more so for the slowest ambulators. A large proportion of the subjects exhibited inappropriate activity of the calf at heel strike, whereas very few had inappropriate activity during the swing phase. Other abnormal EMG patterns observed in the SCI group for the triceps surae include: premature peak of activity during stance; absence of a peak during stance; clonus; late peak during stance; and/or continuous activity throughout the gait cycle. Again the slowest walking subjects exhibited greater discrepancies with respect to that of the control group for the tibialis anterior musculature, showing merged and/or continuously low amplitudes of activity throughout swing and stance without any of the typically observed peaks of activity at heel strike and early-swing in able-bodied gait. The TA was also observed to be inappropriately active or suppressed during stance. Some participants exhibited a singularly timed peak of activity during early swing or at heel strike yet the second burst was absent.

In general, the EMG activity observed in the SCI group from McKay et al's study (1993) could be characterized as having small amplitudes with limited modulation, and not well synchronized to the gait cycle. Furthermore, the interchange between flexors and extensors were impaired, resulting in unwanted levels of coactivation.

In a study by Dietz et al. (1999), the similarity of TA and MG muscle activity was compared between 18 persons with ASIA "A" and "B" complete or nearly complete SCI to that of 16 able-bodied subjects during an acute bout of BWS treadmill walking. Stepping movements could only be elicited in the SCI subjects while walking at approximately 1.5 km/hr with $57\pm12\%$ BWS and while assistance was provided by therapists to move their legs. The phasic similarity of the EMG profiles between groups was determined using a variation ratio and the amplitude relationship was determined using the root mean square (RMS) of the signal's energy.

The results from this study suggest that the EMG amplitude was lower in those with SCI compared with controls for the TA and MG during swing and stance respectively. The EMG amplitude was also observed to be much smaller in those with lower-level injuries (e.g. T9) than that in persons with higher-level injuries (e.g. C6). Likewise, the similarity of the phasic EMG patterns followed this trend, allowing Dietz et al. to conclude that individuals with cervical injuries had better EMG amplitudes and phasic activity than those with thoracic level injuries. Nevertheless, when compared to the control group, the phasic patterns of both TA and MG for all subjects were observed to have pathological modulations, and omissions as well as inappropriate activity during swing in the MG. Furthermore, the EMG amplitudes for both muscles in the SCI group were significantly smaller than that of the control group.

Despite the encouraging results from Dietz et al.'s (1999) study, Stewart, Barbeau, & Gauthier (1991), found that the passive stretching of leg musculature can elicit EMG activities that are rhythmical and similar to those produced when moving the legs actively. Consequently, the passive stretching of the musculature by the therapist can create the illusion of appropriately timed EMG activity (Rossignol & Barbeau, 1994). Thus, it becomes very difficult to partition the observed EMG signal into the components produced by passive stretching of the leg musculature by the therapist and that by the subject via active leg movements. Therefore, the results from Dietz et al. (1999) should be interpreted with caution and studies that have examined the EMG activity in persons with SCI while ambulating independent of therapists assistance are perhaps more meaningful.

Wernig and Muller's 1992 BWSTT study was one of the few studies to record voluntary electromyography (EMG) from four individuals with incomplete SCI while lying supine and while walking over ground. Voluntary movements of the more affected lower limb while lying supine were not possible, whereas EMG activity was observed from some musculature on the affected limb during voluntary movements, yet they were not always time-locked to the verbal commands (i.e. activate & relax). Of greater interest however, were the cyclical flexor and extensor movements and their associated phasic EMG burst patterns from the more severely affected limb while the participants ambulated over ground. Although not depicted in graphical form, the patterns of EMG reportedly varied between subjects with respect to the amount of antagonistic activity, clonus, and duration. Nonetheless, the patterns of EMG are reportedly phasic over several gait cycles. Unfortunately, no interpretation or discussion regarding the appropriateness of the patterns, relative to the gait cycle, was made. Furthermore, no comparisons of the recorded EMG patterns were made to that of able-bodied control subjects walking over ground at similarly slow speeds.

In a case study by Calancie et al. (1994), rhythmical EMG burst patterns were collected from their SCI subject while walking between parallel bars at self selected slow, and fast speeds as well as while suspended over a treadmill at 3, 12 and 18 m/min with 40% BWS. While walking with the assistance of the parallel bars, prolonged extensor activity was observed, especially at slower speeds as well as poorly timed quadriceps activity. Clonus was also observed in the soleus, in both legs during voluntary walking with the parallel bars and while on the treadmill. Increases in speed on the treadmill resulted in corresponding increases in the rate of muscular discharges and for some muscles the burst patterns became more distinct in comparison with the continuous, low-level discharge observed at slower speeds. Leroux, Fung and Barbeau (1999) examined the lower-limb movements and muscle activity in 7 ambulatory individuals with ASIA "D" incomplete SCI while walking on a treadmill at four different levels of incline (0, 5, 10 and 15% grade). Essentially, the duration, timing and EMG amplitude for all muscle activity examined in the SCI group showed little to no adaptation while walking on an incline compared to considerable modulations in all EMG characteristics for the control group.

Leroux et al. (1999), concluded that the SCI group was able to adapt to up-hill walking but employed different strategies than those observed in the control group. The abnormal timing of the plantar flexors was speculated to be caused by stretch activation, whereas co-contraction across the ankle joint during stance may have been necessary to stabilize the joint and compensate for muscle weakness. Another possible explanation for the discrepancies observed in Leroux et al.'s study between groups was that speed was not controlled for. The SCI subjects were allowed to walk at their comfortable walking speed, which ranged from 18 – 48 m/min (1.08 – 2.88 km/hr), whereas the control group walked under two set-speed conditions, 24 and 60 m/min (1.44 & 3.6 km/hr). Therefore, the comparisons made between the two groups might not be valid, especially when these authors admittedly cited other research indicating that adaptations to external demands (e.g. speed), are generally accomplished by changing the pattern of limb motion (Murray et al., 1966 & Nilsson et al., 1985) and varying motor recruitment patterns (Nilsson et al., 1985). Consequently, comparing the motor recruitment and limb motion between the two groups while ambulating at different speeds may be flawed. Furthermore, toe-off for the control group consistently occurred at approximately 60% of the gait cycle whereas for the SCI group, it occurred closer to 80%, indicating phasic/temporal differences between the two groups.

In the study by Fung and Barbeau (1989), surface EMG was measured from the TA and MG for the purposes of developing a "Spastic Locmotor Disorder Index" to quantify disordered muscle activation patterns in spastic paretic subjects. Eight subjects with incomplete SCI performed an acute bout of walking on a treadmill at their maximum comfortable speed (speeds ranged from 16-26 m/min or 0.96-1.56 km/hr). The SCI group was further subdivided into two groups. Four of the eight SCI participants were deemed to be "more" functionally impaired due to their inability to walk faster than 16 m/min (0.96 km/hr). Three of these required manual assistance to advance their limbs on the treadmill, and two of these also required 40% BWS in order to alleviate some of their spasticity. Five control subjects were used for comparison purposes yet they also walked at their comfortable walking speed, which was substantially faster than the SCI groups' average speed. Surface EMG was collected from the TA and MG for all subjects.

Upon comparison of the SCI ensemble averages to that of the control group, the less affected SCI group's TA and MG profiles were characterized by early stretch activation, resulting in inappropriate levels of activity when the muscles should have been silent. In addition to early stretch activation, the "more" functionally impaired group also exhibited clonus (rapidly alternating muscular contractions and relaxations) and elevated tonic activation (sustained partial contractions), resulting in somewhat of a phase shift where the muscles were maximally active when they should have been quiescent and vice versa. Further inspection of the SCI ensemble averages, revealed great variability between subjects, whereas inter-subject variability for the control group was deemed to be much smaller. Furthermore, the patterns produced by the SCI participants were vastly different than those of the control group with periods of premature and delayed activation, co-contraction and many other deviations from "normal".

In a comparative pre-post BWSTT study conducted by Dietz et al. in 1995, EMG was used to determine to what extent muscle activity can be elicited and trained in 10 individuals with complete and 3 with incomplete SCI. All 13 subjects participated in daily locomotor BWS treadmill training for five months, and initially all participants required assistance from therapists to elicit stepping movements by lifting their legs to initiate swing, and provide knee extension during stance. In general all SCI participants in Dietz et al's (1995) study were characterized with poorly modulated gastrocnemius activity, while those with complete injuries also had considerably less EMG activity altogether. The primary concern with the TA was its high levels of co-contraction with the gastrocnemius during stance. Nonetheless, training resulted in a significant increase in MG amplitude whereas the effect of training on the amplitude of TA activity was not as distinct. Training was also found to improve the phasic nature of both muscles to be more like that observed in the able-bodied controls.

The increase in MG amplitude with training observed by Dietz et al. (1995) has functionally meaningful ramifications during gait, especially during stance, and may be directly associated with the SCI participants' ability to bear more weight after training or conversely that with more weight (i.e. increased forces), the magnitude of MG activity was increased. However, Dietz et al. (1995) found little correlation between MG amplitude increases and unloading, thus favoring the assumption that locomotor training improved extensor activity, allowing for greater loads to be born by the SCI participants during stance.

The literature indicates clearly that individuals with complete and incomplete SCI are capable of rhythmical walking over ground (Calancie et al, 1994; Wernig & Muller 1992; Wernig et al., 1995 & Wernig et al., 1998) as well as on treadmills (Calancie et al., 1994; Fung and Barbeau 1989; Leroux et al., 1999 & Wernig & Muller 1992; Wernig et al., 1995 & Wernig et al., 1998). The muscular activities associated with their walking, however, have been observed to be irregular and inappropriately timed relative to the patterns observed in able-bodied controls. Moreover, the between subject variability is generally much higher in the SCI population than that observed in able-bodied controls. The use of BWSTT however, reportedly decreases the stretch of the gastrocnemius during stance, which may facilitate proper activation of this muscle in those with gait abnormalities (Finch et al., 1991), allowing for a more fluid rocker-gait. The results from Dietz et al's (1995) BWSTT study clearly demonstrate improvements in phasic EMG patterns and amplitude modulation after five months of therapy. Thus, despite the irregularities cited in the EMG amplitude and phase during SCI gait, improvements in both dimensions appear to be possible with training. Proper timing of muscle patterns during gait is relatively important for smooth walking patterns. BWSTT facilitates the recovery of locomotion by allowing muscle output patterns to be similar to those observed in normals, which may enhance the probability of regaining full weight bearing activities (Barbeau and Rossignol, 1994).

Due to the complex nature of spinal cord injuries and the difficulties with EMG gait analysis in this special population, many studies have had to compare the muscular activity of controls to that of the SCI population while walking under different conditions such as speed, BWS and /or assistance from a therapist, assistive device or orthotics (Leroux, Fung and Barbeau 1999; Fung and Barbeau, 1989). Consequently, some of the discrepancies observed between the two groups may be in part due to the dissimilarities between the walking conditions in which the measures were taken. Thus, future research should conduct direct comparisons between the gait of spinal cord injured persons to that of able-bodied persons walking under identical conditions.

1.6 Conclusion

From the literature reviewed, it is obvious that most individuals living with traumatic SCI are not capable of functional and/or efficient forms of walking after receiving conventional forms of rehabilitation training. Even with the use of various orthotics, FES and assistive devices, spinal cord ambulation remains inefficient and exhausting (Fisher and Gullickson, 1978 & Waters & Mulroy, 1999). As a direct outgrowth of the high metabolic demand of walking in the SCI population, research indicates that the adherence to walking exercise regimes are low (Rosman & Spira, 1974) and most individuals return to using wheelchairs as their primary means of mobility due to higher metabolic efficiency and greater speeds of locomotion associated with the use of this device (Cerny, 1978).

With the introduction and development of BWSTT, and its reported effectiveness at restoring gait in individuals with SCI (Wernig et al., 1995 & Wernig et al., 1998), it would now appear that functional and efficient ambulation in this population might be a reality given the appropriate training stimulus. However, there is no indication from the literature at this time with regards to the efficiency or metabolic demands while individuals with SCI ambulate on BWS treadmills.

Furthermore, no research has specifically examined the effects of varying BWS and/or speed on the intensity of the exercise stimulus while person with SCI ambulate on a BWS treadmill. Consequently, researchers and therapists alike are blindly prescribing exercise for participants without empirical data to guide and ensure that the desired training stimulus is achieved. Data of this nature would be invaluable for both therapists and researchers when developing experiments, prescribing exercise, evaluating the success of training and/or comparing between studies.

Intuitively, as the mechanical demands of the walking task change so too must the locomotor movements. Many studies have investigated the effects of various experimental conditions on the amplitude and phasic nature of muscular activity during able-bodied locomotion. However, due to the limited ambulatory abilities of most individuals with SCI, few studies have examined the EMG activity within this population. More recently, studies have begun to examine the ambulatory EMG patterns of those with SCI while ambulating over ground and while on a treadmill with BWS (Dietz et al., 1994; Dietz et al., 1995; Dietz et al., 1999; Fung & Barbeau. 1989 & Leroux, et al. 1999). In general, the studies that have examined the dynamic muscular activity during gait in those with SCI indicate improvements in muscular activity with training yet variations and irregularities compared to able-bodied individuals still persist. However, these comparisons may be flawed due to the reported effects of both speed and BWS on dynamic EMG patterns in able-bodied individuals. Consequently, better comparisons between the muscular activity in individuals with SCI to that of able-bodied persons while walking under identical conditions (i.e. speed and assistance) are needed.

In conclusion, the overwhelming potential of BWSTT cannot be overstated, however, as with any treatment modality, there is always a need for improvement and better understanding regarding the variables influencing its success and/or its shortcomings. Further investigations into the energetics and dynamic EMG profiles of those receiving and completing BWSTT programs are needed. More empirical data is required for researchers and therapists to develop, prescribe, evaluate and improve BWSTT programs.

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2.0 Influence of BWSTT for Individuals With Incomplete Spinal Cord Injury: Metabolic Demands

Abstract

Body weight supported treadmill training (BWSTT) is being promoted as an effective means of restoring ambulatory abilities among individuals with incomplete spinal cord injuries (Wernig et al., 1995; & Werning et al., 1998). A lesser-publicized benefit of BWSTT however, is that it may provide a significant cardiovascular and muscular exercise stimulus (Gardner et al., 1998 & Wernig & Muller, 1992). To date, very little is known about the metabolic demands of persons with incomplete spinal cord injury (SCI) while ambulating independently on a body weight support (BWS) treadmill. Oxygen consumption data (VO2) was collected on ten control subjects (mean age = 25.6 ± 2.6 years), free from any gait abnormalities and four individuals with ASIA "C" incomplete SCI (mean age = 28.0 ± 5.2 years) who had completed ten months (122.8 ± 0.96 sessions) of BWSTT. All participants walked independently on the treadmill under nine different test conditions (3 levels of BWS: 20, 40 & 60% X 3 speeds: 0.5, 1.0 & 1.5 km/hr). Oxygen consumption analysis revealed a significant "Group X BWS", $F(2,24) = 6.28$; $p \leq 0.0064$ and "Group X speed" interaction effect, $F(2,24) = 15.92$; $p \leq 0.00004$. Increasing BWS was observed to significantly reduce VO2 for the SCI group but not for the controls, whereas increases in speed were observed to increase VO2 for both groups, with increases in speed having a greater effect on the SCI participants. Despite the lack

of a significant three-way interaction, it was evident that the SCI group's oxygen consumption was more than 1.5 times that of the control group for all nine conditions tested with the exception of one. Upon examining the metabolic demands per unit walked (i.e. metabolic cost = ml/kg/m), a significant three-way interaction effect (Group X Speed X BWS) was observed, $F(4,48) = 4.73$; $p \leq 0.0027$, with significantly higher levels of inefficiency being found in the persons with SCI. Due to the significantly higher metabolic demands of walking in the SCI group, BWSTT has the potential for providing an effective aerobic exercise stimulus in this population.

2.1 Introduction

Body weight supported treadmill training (BWSTT) has been shown to be an effective means of restoring ambulatory abilities among individuals with incomplete spinal cord injuries (Wernig, Muller, Nanassy, & Cagol, 1995; & Wernig, Nanssy, Muller & Cagol 1998). However, the gait achieved by these individuals is generally awkward, and can be achieved only over a reduced range of speeds (McKay, Metman, Dimitrijevic, Sherwood & Dimitrijevic, 1993 & Leroux, Fung & Barbeau, 1999). Oxygen uptake measures have been taken while individuals with spinal cord injury (SCI) ambulate over ground with assistive devices (canes, crutches, walkers, etc.) and/or while wearing orthoses. Those results have shown metabolic rates as much as six times higher than that in able-bodied persons (Clinkingbeard, Gersten & Hoehn, 1964; Fisher & Gullickson 1978, & Waters & Mulroy, 1999). Consequently, after receiving conventional therapy, many persons with SCI return to using their wheelchairs as their primary means of locomotion (Rosman & Spira, 1974) because their energy expenditure while walking far exceeds that of wheelchair use (Cerny, 1978 & Cerny, Waters, Hislop, & Perry, 1980).

The speed of locomotion affects the metabolic demands of walking in able-bodied individuals in a curvilinear fashion, where increases in speed result in higher metabolic rates of oxygen consumption (Corcoran & Gelmann, 1970; Fisher & Gullickson, 1978; Molen & Roxendal, 1967; Ralston, 1971; & Waters and Mulroy, 1999). A similar speed – oxygen uptake relationship was also reported by Clinkingbeard et al. (1964) for individuals with SCI ambulating with knee-ankle-foot orthoses (KAFO's).

Increases in BWS have also been shown to cause a reduction in the metabolic demands while walking in able-bodied individuals (Colby, Kirkendall, and Bruzga, 1999 & Danielsson & Sunnerhagen, 2000). To date, no study has examined the effect of BWS on metabolic demands in persons with SCI.

Comfortable walking speeds in the able-bodied population reportedly range between 60 and 100 m/min (Waters, Lunsford, Perry & Byrd, 1988), whereas the speed of ambulation in persons with SCI, while using braces and orthoses, is pathologically slow (Beillot, et al., 1996; Fisher & Gullickson, 1978 & Waters & Mulroy, 1999). Compared with an average speed of 80 m/min in able-bodied individuals (waters & Mulroy, 1999), the average speed of ambulation in persons with SCI with KAFO's was 11.3 m/min (Clinkingbeard et al., 1964), whereas those using reciprocating-gait orthoses (RGO) was 16.1 m/min (Bowker, Messenger, Ogilvie, & Rowley, 1992). In addition to their non-functional speeds of ambulation, individuals with SCI have correspondingly high metabolic costs, ranging from 0.56 ml/kg/m (Waters & Mulroy, 1999) up to 1.97 ml/kg/m (Beillot et al., 1996). Compared to the metabolic cost of the able-bodied population walking at comfortable speeds (0.14 ml/kg/m – 0.16 ml/kg/m), it becomes evident that the remaining ambulatory function in individuals with SCI is greatly limited (Waters & Mulroy, 1999). On the other hand wheelchair locomotion in persons with SCI is much more efficient (0.18 ml/kg/m), and can attain more functional speeds of locomotion (75 m/min) (Cerny, 1978).

Although better ambulatory abilities are reported post BWSTT in persons with incomplete SCI, as compared to conventional therapy (Wernig et al., 1995 & Wernig et al., 1998), there is no evidence to suggest that these individuals achieve the ability to ambulate effectively (i.e. > 60 m/min with a metabolic cost ~ 0.16 ml/kg/m) in the community. In fact, Wernig et al.'s (1995 & 1998) studies indicate only that some subjects were capable of taking 5 consecutive steps after training, but no measures of speed or endurance were given.

To date, only one study has examined the energy demand of ambulation in persons with SCI before and after twelve weeks of BWSTT (Protas, Holmes, Qureshy, Johnson, Lee, & Sherwood, 2001). The results from this study indicate that after training, the subjects were capable of greater speeds of locomotion, (4 m/min pre Vs 19 m/min post), and achieved significantly better metabolic efficiency (1.96 ml/kg/m pre Vs 1.33 ml/kg/m post). However, the metabolic demands and speeds of ambulation reported in this study were taken while the subjects ambulated as quickly as possible, back and forth on a 5 m walkway while using their customary assistive devices and/or orthotics. Although these results may seem promising, neither the post-training speeds or metabolic efficiencies come close to that reported for able-bodied individuals, suggesting that despite significant gains, individuals with SCI cannot achieve functional speeds of ambulation after BWSTT.

A lesser-publicized benefit of BWSTT however, is that it may provide a significant cardiovascular and muscular exercise stimulus (Gardner et al., 1998 & Wernig & Muller, 1992). Unlike conventional therapies, where the exercise stimulus might be limited to the upper extremities, BWSTT provides the opportunity for progressive, and upright, lower extremity exercise in a safe and controlled environment. Unfortunately, very little is known about the achievable range of exercise intensities and/or the combined effects of BWS and speed on the metabolic demands of walking in either able-bodied individuals, or those with incomplete SCI. Since both BWS and speed have been found to have independent effects on the metabolic demands of walking in able-bodied individuals, both of these factors need to be controlled for in studies attempting to compare the metabolic demands of walking between able-bodied and SCI individuals. A study of this nature will provide objective information for researchers and therapist to develop future training studies, evaluate the progress of their patients and/or effectively prescribe exercise in this population.

2.1.1 Purpose

Very little is known about the metabolic demands (i.e. intensity) of walking in persons with incomplete SCI, nor are the independent and/or interaction effects of BWS and speed clearly understood. The purpose of this paper is two-fold. First, to develop an index of the metabolic demands of able-bodied individuals while walking on a body weight support treadmill under varying combinations of speed and BWS. Secondly, to compare and contrast the metabolic demands of the able-bodied control group to that of individuals with incomplete SCI who have received extensive BWSTT.

2.2 Methods

2.2.1 Subjects

Ten able-bodied male and female control participants (five male, five female) volunteered to take part in the present study. Their ages ranged from 22 to 30 (mean 25.6 ± 2.6) years, their heights ranged from 152 – 180 (mean 167.8 ± 8.6) cm and their weights ranged from 50.0 to 88.6 (mean 71.1 ± 13.5) kilograms. None of the control subjects had any musculoskeletal problems and/or gait abnormalities.

Four individuals with ASIA "C" incomplete SCI who had completed ten months (122.8 ± 0.96 sessions) of thrice-weekly BWSTT were recruited from a pool of participants in a 12-month BWSTT study at McMaster University, Ontario (See Table 2.1 for the detailed SCI participant information). In order to be included in the study, the SCI participants had to be able to walk independently (i.e. move both their legs free from the influence, guidance and/or assistance from external sources) on the treadmill for twelve consecutive minutes at 1.5 km/hr with 20% BWS. Although some of the SCI participants were able to ambulate on the treadmills with no BWS, this was not a criterion for inclusion.

This study received ethical clearance from the McMaster Research Ethics board. All participants read and gave their written consent to participate after reading the information letter outlining the procedure, risks and benefits of the study.

Table 2.1: SCI Subject information. All numerical values are the mean ± one standard deviation

Subject	Age	Weight (kg)	Height (cm)	Session Tested	Gender	Cause of Injury	Lesion Level	ASIA Score	Yrs. Post Injury
SCI 01	33	50	157.5	124	F	MVA	C5-6	C	6
SCI 02	23	64.1	165.0	122	M	Fall	T12-L1	C	2.5
SCI 03	24	103.2	190.5	123	M	Birth	C5-6	C	24
SCI 04	32	107.7	185.4	122	M	Knife	C4	C	2.5
Mean ± SD	**28.0 ±5.2**	**81.3 ±28.6**	**174.6 ±15.8**	**122.8 ±0.96**					**8.8 ±10.3**

2.2.2 Experimental Design and Statistical Analysis

The present study was a three-factor (3 levels of BWS X 3 speeds X 2 groups) mixed design (2 within, 1 between), with fixed effects (See Table 2.2 for the design). The three levels of BWS used were 20, 40 and 60% and the three speeds employed were 0.5, 1.0 & 1.5 km/hr (8.3, 16.7, & 25 m/min, respectively). The levels of BWS and speeds used were selected because they are conditions commonly used during BWSTT in individuals with incomplete SCI and all of the subjects with SCI had previous experience walking under these conditions. A three-way, mixed ANOVA, was used to analyze the data while Tukey's HSD was used for all post-hoc analyses ($P < 0.05$).

2.2.3 VO2 Data Collection System

The metabolic data (VO2) was collected using a metabolic cart equipped with a barometer, ambient temperature thermometer, mixing box, flow turbine, CO2 analyzer (Hewlett Packard, 47210A), O2 analyzer (Beckman, OM-11), and a generic computer running Turbofit metabolic software (Vacumed, V4.047). After input of the participant's, weight, height and gender into the software, the expired gas samples were collected by placing a sterilized Hans Rudolph valve in the participant's mouth and placing a cushioned clip over their nose. A lightweight headpiece was worn by all subjects to help support the weight of the valve. The participants inhaled ambient air while the expired gases were directed towards a mixing chamber via a 1.5" diameter clear bore tubing where the mixed gases were analyzed for CO2 and O2. At the opposite end of the mixing chamber was a low resistance flow turbine to determine the expired gas volumes (VE). The data from the gas analyzers and other instruments were output to the computer running Turbofit which in turn outputted oxygen consumption values (VO2, ml/kg/min), minute ventilation (VE, l/min), carbon dioxide production (VCO2, ml/kg/min) and respiratory exchange ratios every thirty seconds. All data were stored on the local computer hard-drive and printed after each test. The system was calibrated prior to data collection and recalibrated after every bout of exercise using room air and a sample of known gas concentrations (5.08% carbon dioxide, 11.99% oxygen and 82.93% nitrogen).

Table 2.2: Summary table of the three-factor, fixed effects, mixed design (3 Speeds X 3 BWS X 2 Groups). All measures were repeated over the nine conditions.

	Speed 1 (0.5 km/hr)			Speed 2 (1.0 km/hr)			Speed 3 (1.5 km/hr)		
	20% BWS	40% BWS	60% BWS	20% BWS	40% BWS	60% BWS	20% BWS	40% BWS	60% BWS
Controls N = 10									
SCI N = 4									

2.2.4 Procedure

Oxygen consumption data (VO2) was collected continuously on all participants while walking independently on a body weight support treadmill (Woodway Lokosystem), under nine different test conditions (3 levels of body weight support X 3 speeds). The treadmill is equipped with an overhead hoist system, which works with counter weights of varying mass to provide numerous levels of body weight support and has a control panel, with digital display, to vary speed. The body weight support cables were fastened to a harness donned by all participants. In order to ensure that the prescribed level of BWS was maintained throughout each condition, the participants were encouraged not to use the handrails for support (yet were allowed to use them for balance only).

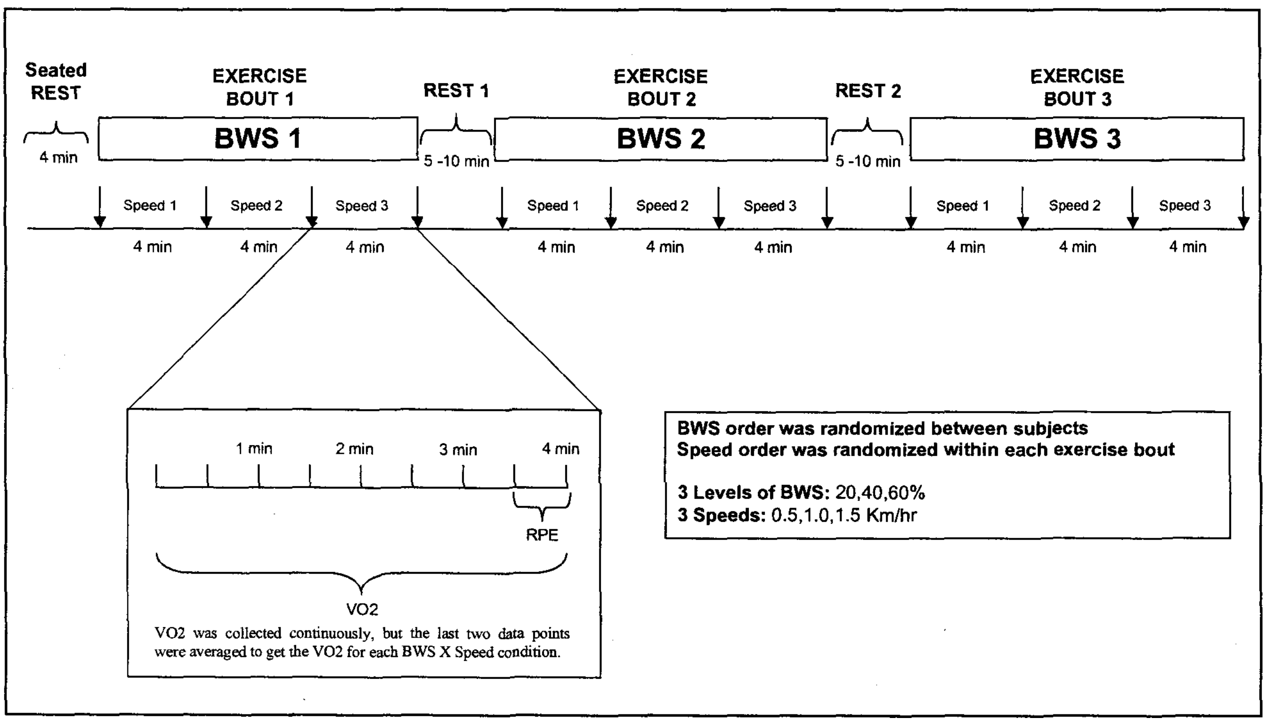

Figure 2.1: Schematic illustration of data collection time-line and procedure. See text for description.

The experimental session consisted of three, twelve-minute bouts of exercise, separated by five to ten minutes of rest to prevent fatigue. Each session started with a four minute seated rest period in order for the participants to acclimatize to breathing through the system. After the rest period was over, the first exercise bout commenced. Each bout of exercise occurred at a randomly different, yet constant level of BWS, while the speed of the treadmill was randomly changed every four minutes within each exercise bout, allowing for steady-state to be achieved for each condition (See Figure 2.1). Thus, BWS was randomized between subjects while speed was randomized within each bout of exercise and between subjects. The last two VO2 values outputted by the system (i.e. 3:30 and 4 minutes) were averaged in order to ensure more stable and precise values. The metabolic cost was derived by taking the metabolic rate (ml/kg/min) and dividing by speed (m/min).

2.2.5 Heart Rate and Rating of Perceived Exertion

All participants were equipped with a wireless Polar™ heart rate monitor in order to record heart rate during the last 30 seconds of each exercise condition. Unfortunately, for most subjects, due to movement of the body weight support harness, the chest sensor moved and a consistent heart rate could not be recorded. The participant's rating of perceived exertion (RPE) for how hard they perceived their breathing to be was monitored using the modified ten point Borg Scale (See Appendix A). Participants were asked "How hard are you breathing" during the last 30 seconds of each condition, while the scale was held in front of them. The participants would either point to the scale or nod their heads when the experimenter pointed to the correct score on the chart as they progressively worked their way up the scale from zero (0).

2.3 Results

2.3.1 Metabolic Demands

Table 2.3 presents the mean metabolic values, plus-minus one standard deviation (metabolic rate ml/kg/min & metabolic cost ml/kg/m) for both groups across all nine conditions tested. Although the 3-way interaction for the metabolic rate data was not found to be significant ($p \geq 0.19$), the SCI group's oxygen consumption was on average 1.7 times higher than that for the control group across all nine conditions. In fact, all VO2 values were at least 1.5 times higher in the SCI group, with the exception of one condition (0.5 km/hr @ 40% BWS), which was still 41% greater. However, two of the SCI conditions had metabolic rates nearly twice that of the control group (1.0 km/hr @ 20% BWS and 1.5 km/hr @ 20% BWS). In general, the metabolic rate for both groups increased as speed increased within each level of BWS, but more so for the SCI group than the controls. On the other hand, increasing BWS tended to reduce the metabolic rate for the SCI group only.

2.3.2 Effect of BWS

The results from the 3-way ANOVA on the metabolic rate data (ml/kg/min) revealed that there was a significant "Group X BWS" interaction, $F(2,24) = 6.28$; $p \leq 0.0064$. Figure

2.2 presents the "Group X BWS" interaction effect for the metabolic rate data. The VO2 for the SCI group was significantly higher than that for the control group for all three levels of BWS (indicated by the asterisk "*" between the lines). Increases in BWS from 20% to 40% or from 20% to 60% BWS were found to significantly reduce oxygen consumption for the SCI group whereas increasing BWS did not affect the metabolic demands of the control group, nor was there any apparent trend.

Table 2.3: Group mean metabolic rates (ml/kg/min) and costs (ml/kg/m) for all nine conditions tested (Controls N = 10, SCI N = 4).

		Metabolic Rate (ml/kg/min)		Metabolic Cost (ml/kg/m)	
BWS (%)	Speed (km/hr)	Control	SCI	Control	SCI
	0.5	5.78 ± 1.06	10.12 ± 1.73	0.69 ± 0.13	1.22 ± 0.21
20	1.0	5.99 ± 0.59	11.72 ± 1.48	0.36 ± 0.04	0.70 ± 0.09
	1.5	6.69 ± 0.91	12.75 ± 1.93	0.27 ± 0.04	0.51 ± 0.08
	0.5	5.99 ± 1.24	8.43 ± 1.08	0.72 ± 0.15	1.01 ± 0.13
40	1.0	6.38 ± 1.14	10.47 ± 1.33	0.38 ± 0.07	0.63 ± 0.08
	1.5	6.64 ± 1.25	11.64 ± 1.74	0.27 ± 0.05	0.47 ± 0.07
	0.5	5.17 ± 0.83	7.77 ± 0.70	0.62 ± 0.10	0.93 ± 0.08
60	1.0	5.62 ± 0.92	10.56 ± 2.33	0.34 ± 0.06	0.63 ± 0.14
	1.5	6.17 ± 1.03	10.42 ± 0.85	0.25 ± 0.04	0.42 ± 0.03

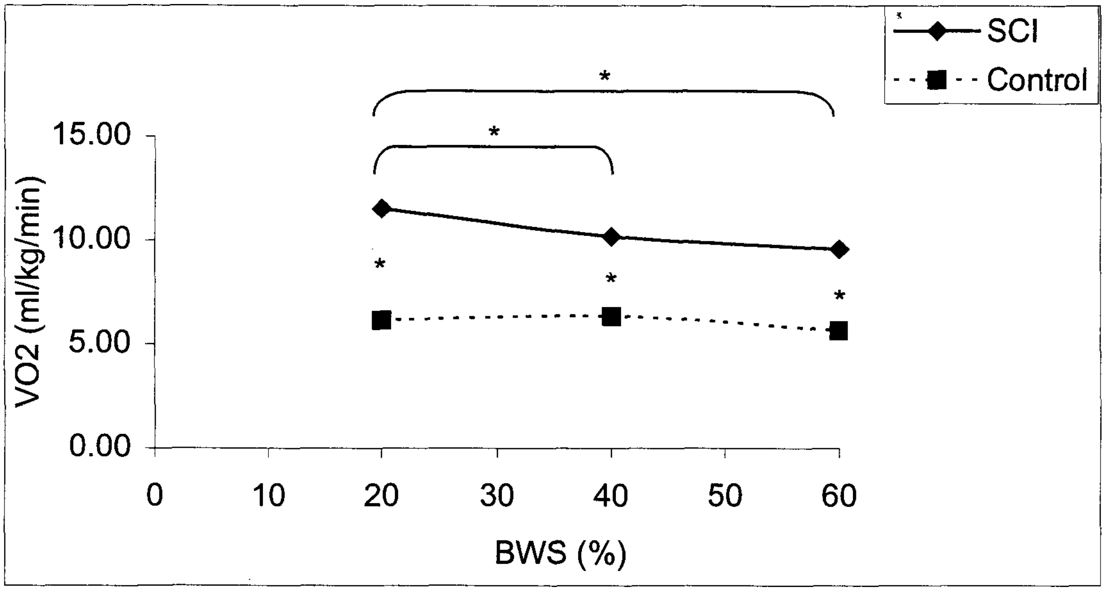

Figure 2.2: Group X BWS interaction: "*" Between lines indicates a significant between group difference. The"*" above bracket indicates a significant difference between the two conditions enclosed by the brackets. Solid line represents the SCI group whereas the broken line represents the controls.

2.3.3 Effect of Speed

A significant "Group X speed" interaction effect, F (2,24) = 15.92; $p \leq 0.00004$ was also found and is illustrated in Figure 2.3. Again, the VO2 values for the SCI group, for all three speeds, were found to be significantly higher than those for the control group (indicated by the asterisk "*" between the lines). The post-hoc analysis revealed that increasing the speed from 0.5 to 1.0 km/hr significantly increased the VO2 for the SCI group but not for the controls. Tripling the speed (0.5 – 1.5 km/hr) significantly increased the metabolic demands for both groups, whereas increasing the speed from 1.0 to 1.5 km/hr did not significantly increase oxygen consumption for either group, yet an upward trend was still visible. In general, increasing speed had a greater effect on VO2 in the SCI group versus the controls.

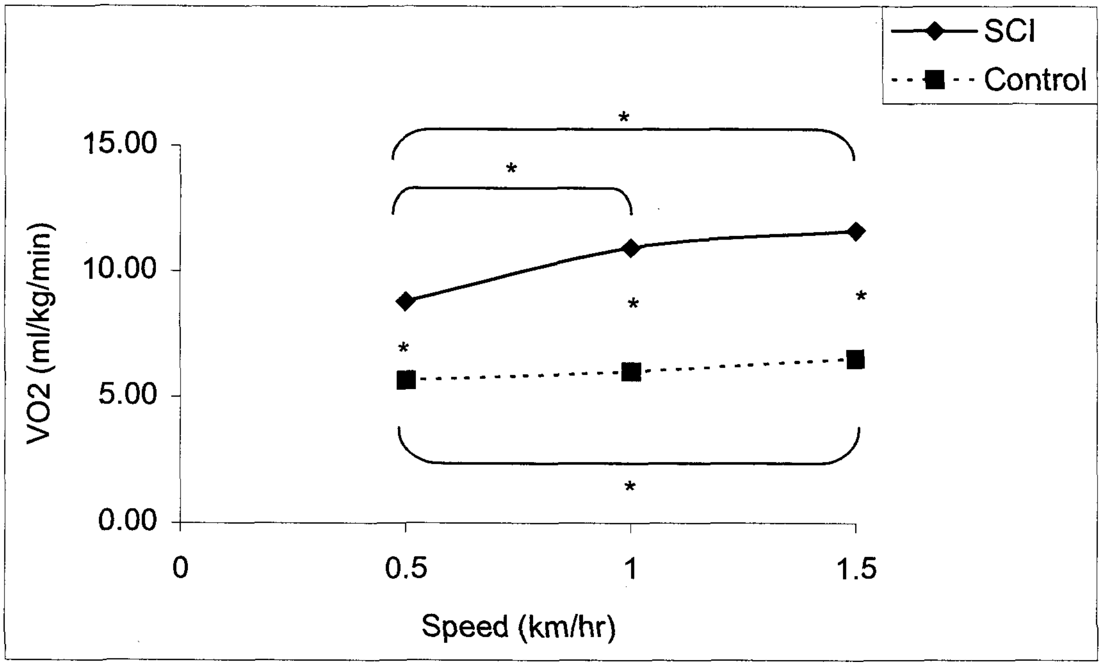

Figure 2.3: Group X Speed interaction: "*" Between lines indicates a significant between group difference, "*" above bracket indicates a significant difference between the two conditions enclosed by the brackets. Solid line represents the SCI group whereas the broken line represents the controls.

2.3.4 Ratings of Perceived Exertion

A three-way interaction was found for the RPE data, $F(4,48) = 4.79$; $p \leq 0.002$ and are presented in Table 2.4. The RPE's were found to be significantly higher in the SCI group for only three of the nine conditions. These significant between group differences occurred while walking at 1.0 and 1.5 km/hr both with 20% BWS and while walking at 1.5 km/hr with 60% BWS (indicated by "*"). Additionally, the breathing effort while walking at 1.5 km/hr with 20% BWS was perceived to be significantly harder than any of the other eight conditions within the SCI group as well as that for any of the control conditions (indicated by "†"). The SCI RPE while walking at 1.0 km/hr with 20% BWS (RPE = 1.5), and that for 1.5 km/hr with 60% BWS (RPE = 1.63) were also perceived to be significantly harder than all other conditions in the control group. As for the control group, none of the conditions were perceived to be significantly different from one another.

Table 2.4: Ratings of perceived exertion (RPE) for breathing effort. The asterisk "*" denotes a significant higher RPE for the SCI group relative to the controls. The "†" denotes a significantly higher RPE from all other conditions, SCI and control alike.

		RPE Breathing	
BWS (%)	**Speed (km/hr)**	**Control**	**SCI**
	0.5	0.25 ± 0.42	0.88 ± 0.85
20	1.0	0.40 ± 0.66	1.50 ± 1.73 *
	1.5	0.30 ± 0.42	2.50 ± 1.73 * †
	0.5	0.30 ± 0.42	1.00 ± 1.41
40	1.0	0.35 ± 0.47	1.00 ± 1.41
	1.5	0.40 ± 0.46	1.13 ± 1.44
	0.5	0.35 ± 0.67	1.00 ± 1.35
60	1.0	0.50 ± 0.94	0.75 ± 0.96
	1.5	0.55 ± 0.80	1.63 ± 1.60 *

2.3.5 Metabolic Cost Three-way Interaction

The ANOVA performed on the metabolic cost data (ml/kg/m) revealed a significant three-way "Group X BWS X Speed" interaction effect, F (4,48) = 4.73; $p \leq 0.0027$. These data are presented in Table 2.3 and depicted graphically in Figure 2.4. On average, the metabolic cost across all nine conditions for the SCI group was 72% higher than that of the controls. In fact, the SCI group's metabolic costs were significantly higher for each of the nine conditions examined (as indicated by the asterisks "*" above the brackets in Figure 2.4). Furthermore, the SCI metabolic cost while walking at 0.5 km/hr with 20% BWS (i.e. 1.21 ml/kg/m) was significantly higher than all other SCI and control values respectively

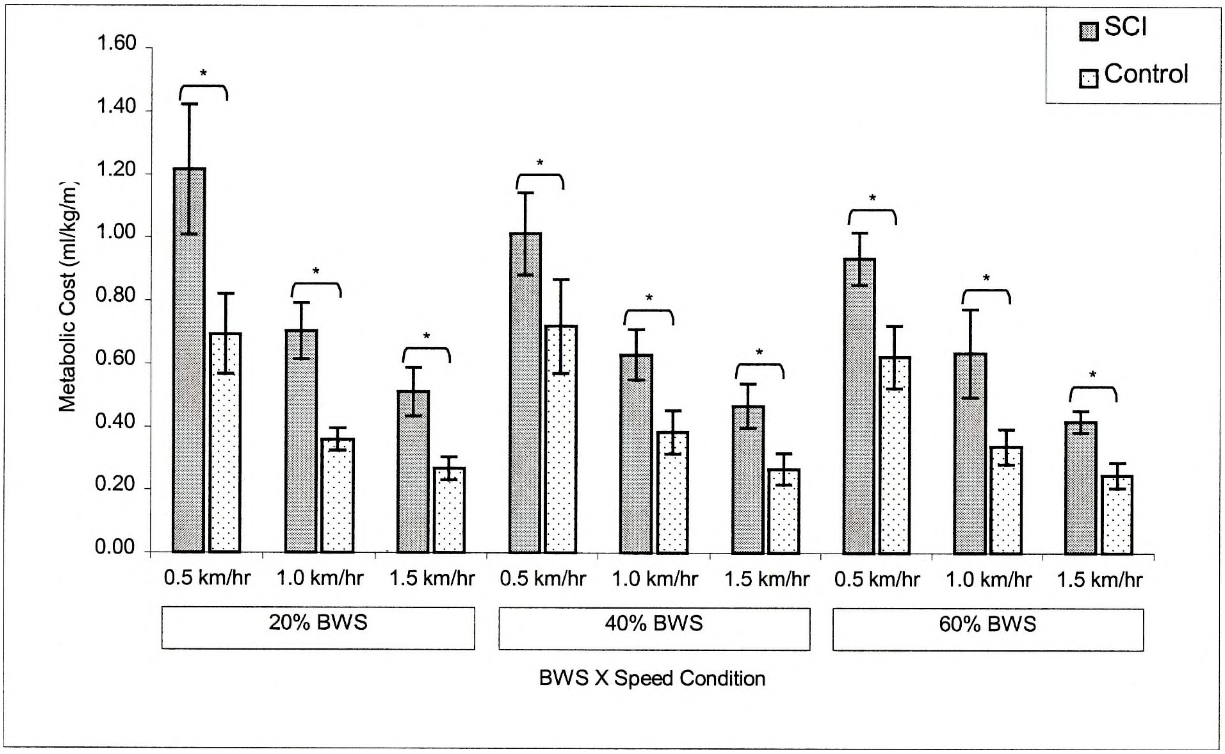

Figure 2.4: Metabolic cost (ml/kg/m) "Group X Speed X BWS" 3-way interaction. "*" Above bracket indicates a significant difference between groups for that "BWS X Speed" condition. Dotted bars are control group, solid bars are SCI group.

Any increase in BWS (i.e. 20-40, 40-60 or 20-60%) while walking at 0.5 km/hr was found to significantly decrease the metabolic cost for the SCI group; whereas, only increases from 20-60% or from 40-60% BWS significantly decreased the metabolic cost for the control group while walking at 0.5 km/hr The only other significant improvement in metabolic cost associated with an increase in BWS was found for the control group when the BWS was increased from 40-60% while walking at 1.0 km/hr. All other increases in BWS did not significantly improve the efficiency for either group. On the other hand any increase in speed (i.e. 0.5-1.0, 1.0-1.5, or 0.5-1.5), regardless of the level of BWS, resulted in a significant reduction in metabolic cost for both groups.

Equivalent metabolic costs between the two groups occurred only when the speed of ambulation for the SCI group was faster than that of the control group. For example, the metabolic cost for the SCI group while walking at 1.0 km/hr with 20% BWS (0.70 ml/kg/m) was similar to that for the control group while walking at 0.5 km/hr (0.69 ml/kg/m) with the same level of BWS. Similarly, the metabolic costs between the two groups were equivalent when the SCI group walked at 1.0 km/hr and the control group walked at 0.5 km/hr and while the SCI group walked at 1.5 km/hr and the control group at 1.0 km/hr with 40% BWS. Likewise, at 60% BWS, the SCI group had equivalent metabolic costs to that of the controls when walking one speed faster. Unlike the effects of speed, no manipulations in BWS resulted in equivalent metabolic costs between the two groups.

2.4 Discussion

The present study was conducted in order to describe and compare the metabolic demands of walking under varying levels of BWS and speed between able-bodied individuals and persons with incomplete SCI who had completed ten months of BWSTT. Comfortable walking speeds for able-bodied individuals may range anywhere from 4.0

km/hr (Petrofski and Smith, 1991) to 5.0 km/hr (Fisher and Gullickson, 1978). However, in this study, the control subjects were required to walk at unusually slow speeds, ranging between 0.5 km/hr and 1.5 km/hr (increased by ½ km/hr intervals), while a proportion of their body weight was supported by a harness. These artificial walking conditions were deemed necessary in order to make direct comparisons between the control subjects and that of four individuals with SCI who had received extensive (122.8 ± 0.96 sesions) BWSTT and were capable of walking independently on the treadmill. This experimental manipulation was completed in order to compare, describe and determine more precisely the effects of both speed and BWS on the metabolic demands of walking in both groups.

2.4.1 Metabolic Rate

Despite the lack of a three-way interaction for the metabolic rate data, it was clear that the SCI group's VO2's were higher than that for the controls for all nine conditions tested. The VO2 values for the SCI group were at least 1.4 times higher with the average metabolic rate being 72% greater for all nine conditions, five of which had metabolic rates ≥75% that of the controls. The highest VO2 recorded for both groups occurred while walking at the highest speed (i.e. 1.5 km/hr), with the least amount of BWS (i.e. 20%), which was hypothesized to be the hardest of the nine conditions. The RPE data also supports the notion that this condition was indeed the hardest of all those tested. On the other hand, the lowest VO2 corresponded with the condition with the slowest speed with the most BWS, again, inline with our hypothesis that this would be the easiest condition. The RPE was perceived to be significantly lower for this condition than while walking at 1.5 km/hr with 20% and 60% BWS respectively, and was not perceived to be any different than any of the other six conditions for the SCI group. None of the conditions were perceived to be different from one another by the control group.

Based upon the above findings, it would seem reasonable to conclude that increasing BWS and reducing walking speed causes a perceivable reduction in metabolic demands, whereas decreasing BWS and increasing speed has the reciprocal effect for persons with SCI. Our results also suggest that manipulating speed and BWS seem to have a more profound effect on persons with SCI than on able-bodied individuals.

One method of getting a better appreciation of the intensity of the work performed by the SCI group while walking on the treadmill is to estimate an equivalent walking speed of an able-bodied person. By substituting the SCI group's metabolic rates into Ralston's (1971) metabolic predictive equation (i.e. $VO2 = 0.0011V^2 + 5.9$, where V is in m/min) and solving for velocity, the resultant walking speed of an able-bodied person while walking over level ground can be estimated. For example, the VO2 for the SCI group while walking at 0.5 km/hr with 60% BWS was 7.77 ml/kg/min yet would be the equivalent to an able-bodied person walking at 2.5 km/hr with no BWS, over level ground. Similarly, the highest rate of oxygen uptake for the SCI group occurred while waking at 1.5 km/hr with 20% BWS (12.75 ml/kg/min), which is equivalent to that of an able-bodied person walking at 4.7 km/hr (again using Ralstons' equation and solving for velocity).

Hence, from this simple calculation, it can be argued that the SCI participants achieved an exercise stimulus equivalent to that of able-bodied persons walking at much greater speeds on level ground, with no BWS. Furthermore, all of the SCI participants in this study completed a total of 36 minutes of walking during the testing session. Consequently, if individuals with SCI could take part in a BWSTT program consisting of thirty to sixty minutes of walking, under any of the conditions tested in this study (0.5 km/hr @ 60% BWS up to 1.5 km/hr @ 20% BWS), four days a week, they would meet the minimum exercise recommendations in Health Canada's physical activity guide (2002). An exercise stimulus of this nature should help improve their fitness level and help guard against many health risks of inactivity such as heart disease, obesity, depression, etc., if accompanied with a healthy lifestyle.

A more conventional method of determining the intensity of the nine walking conditions would be to compare the SCI participants' sub-maximal VO2's to VO2peak values determined using a graded exercise test. However, no such test was conducted on our participants in this study. Nevertheless, it is still reasonable to compare the metabolic demands of the hardest (1.5 km/hr @ 20% BWS) and easiest (0.5 km/hr @ 60% BWS) conditions in this study to VO2peak values reported in the literature for graded exercise tests conducted on individuals with SCI. These comparisons should provide some insight into the intensity of the exercise achieved while walking on the treadmill.

Beillot et al.'s (1996) VO2peak for their graded arm-crank ergometry (ACE) test on 24 paraplegics with T2 – T12 lesions, after 12-weeks of reciprocating gait orthoses (RGO) ambulatory training, was 20.0 ± 4.5 ml/kg/min. Thus, expressing our VO2's as a percent of the mean VO2peak in Beillot et al.'s study indicates that our subjects may have been working at 64% VO2peak for the hardest condition or 39% VO2peak for the easiest. Given that our subjects were exercising with their legs, it may however, be more appropriate to express their VO2's as a percentage of a leg-cycling ergometry, LCE-VO2peak. This comparison reveals a 66% and 40% VO2peak respectively for the hardest and easiest conditions when using the VO2peak value reported by Hooker, Scremin, Mutton, Kunkel & Cagle's 1995 training study (i.e. 19.4 ± 7.5 ml/kg/min post 27 weeks of progressive neuromuscular-electrical-stimulation LCE training).

The only concern with the above two comparisons however, might be that a proportion of the VO2 values presented in our paper may be due to the work performed by the upper limbs of our SCI participants on the treadmill handrails due to their various ambulatory difficulties. Although instructed not to bear any weight with their arms, the load, if any, applied to the handrails was not monitored by means of any instrumentation (e.g. strain gauge). Even small loads applied to an assistive device by the upper arms (i.e. 2-3% body weight) such as a cane, have been found to increase the oxygen uptake rate by as much as 38% of that observed in normals (Waters, Yakura, Adkins, & Barnes, 1989). Consequently, it is possible that a proportion of the VO2's observed in the SCI group, especially at low levels of BWS, may have been due to loads borne by their arms.

Thus, it may also be instructive to express our values as a percentage of the VO2peak values reported by Mutton et al. (1997) who trained eight ASIA "A" SCI participants over 52 weeks (i.e. 93.6 ± 28.1 30 minute exercise sessions) using a hybrid program which involved functional-electrical-stimulation leg-cycle ergometry along with arm-crank ergometry, (FES-LCE-ACE). Using Mutton et al.'s (1997) data (i.e. VO2peak

= 23.2 ± 5.9 ml/kg/min post training) our subjects were walking at 55% and 30% VO2peak for our hardest and easiest conditions respectively.

From the above three comparisons, it is evident that depending on the VO2peak value used, and which walking condition used from our study, the intensity of walking on the BWS treadmill may range between 30% (using the FES-LCE-ACE-VO2peak) for the easiest condition, upwards to 66% (using LCE-VO2peak) while walking at 1.5 km/hr with 20% BWS. Although these comparisons are not ideal, it is a reasonable assumption that our subjects would have had similar VO2peaks to those reported in the literature given the similarity in injuries and extent of training received. Thus, these values provide a reasonable estimate of the lower intensity limit of exercise achieved while our SCI participants walked on the treadmill at the speeds and levels of BWS tested. Consequently, the metabolic rate data from the present study for the SCI group while walking independently on the BWS treadmills suggests that an effective exercise stimulus is possible, especially when increasing the treadmill speed to 1.5 km/hr and reducing the BWS to 20%, which is in line with previous recommendations made by Colby et al., 1999 & Protas et al., 2001 (i.e. increasing the speed of the treadmill and/or decreasing BWS can increase the intensity of BWS treadmill training).

2.4.2 Effect of BWS on Metabolic Rate

By collapsing across all three speeds for each of the BWS conditions we realized a significant "Group X BWS" interaction effect. This finding provides some insight into the relationship between BWS and rates of oxygen consumption, and perhaps more importantly that BWS may affect oxygen consumption differently, depending on whether or not you have a gait abnormality.

The SCI group's metabolic rates were all significantly higher than those of the control group (see Figure 2.2). Furthermore, increasing the BWS from 20 to 40% significantly reduced the VO2 from 11.53 to 10.18 ml/kg/min (13% reduction in VO2) and an increase from 20% to 60% also reduced the VO2 by 20% in the SCI group (11.53 – 9.58 ml/kg/min). Conversely, no significant changes in VO2 with changing levels of BWS were observed in the control group. Therefore, while no apparent trend was observed in the control group, BWS appears to provide some form of metabolic relief for SCI participants.

The different effects of BWS on oxygen consumption between the two groups may result from the fact that all of the SCI participants had visually detectable gait abnormalities. These gait abnormalities presumably cause exaggerations in the motion of the body's centre of mass in all planes, thus interfering with the smooth and natural conversion of potential (i.e. $PE = mgh$) to kinetic energy (i.e. $KE = \frac{1}{2}mv^2$) and vice versa, throughout the gait cycle, subsequently driving up the metabolic demands of walking (Griffin, Tolani & Kram, 1999). Based upon our observations it is also assumed that the BWS harness may have improved the conservation of energy for the SCI group by smoothing out and/or eliminating some of the irregularities in displacement patterns of the body's centre of mass in the vertical, medial-lateral and/or anterior-posterior directions. Additionally, the harness may have reduced the absolute vertical displacement (i.e. "h" in the PE equation) of the body's centre of mass as reported by

Finch et al., (1991) in their examination of the effects of BWS on able-bodied gait. This too would help reduce oxygen consumption, through a reduction in amount of PE required within the system.

The reduction in ground reaction forces (Griffin et al., 1999) and body mass provided by the BWS harness may have also reduced some of the muscular forces (i.e. F = ma) required to accelerate the body's centre of mass, upwards (recall; PE = mgh i.e. both "m" and "h" are presumably reduced) and forwards (recall; KE = ½ mv^2 , where "m" has been reduced), all of which would help reduce the metabolic rate of oxygen consumption when walking with BWS.

Interestingly, increasing the BWS from 20-40% in the SCI group reduced the metabolic rate by 13% whereas an equivalent 20% increase in BWS for the same group of subjects (i.e. 40-60% BWS) only provided a 6% reduction in metabolic rate. Thus, it would appear that the principle of diminishing returns applies to the benefits of BWS in reducing metabolic rates of oxygen consumption in persons with SCI. This result may have significantly meaningful metabolic ramifications when trying to wean SCI participants off BWS during a progressive BWSTT program. For example, initial reductions in BWS, from 60-40% may be readily achieved without causing dramatic increases in VO2, whereas later reductions (e.g. from 40-20% or perhaps from 20 –0%), may be associated with prohibitively high VO2's, thus limiting the participant's ability to exercise at this level of BWS. For training purposes then, initially large and quick reductions in BWS may be possible but later on in training, reductions from 40% and below may have to be small, with greater time between reductions to allow for a training effect to take place.

The lack of a significant effect of BWS on oxygen consumption in the control group came as a surprise given the significant reductions in VO2 caused by 20 and 40% BWS previously reported by Colby et al. (1999). Mechanical differences between the unloading system of the present study and that by Colby et al., (1999) may explain the apparent discrepancies. However, the more likely explanation for the discrepancy in results could be that our maximum speed was dramatically less than that used by Colby and colleagues (1.5 Vs 4.82 km/hr). Thus, from this, it could be argued that BWS may provide greater relief at higher speeds of locomotion for able-bodied individuals than while walking very slowly (i.e. less than 1.5 km/hr). Parenthetically, the differences between the two studies suggest an interaction effect between speed and BWS on metabolic rates, yet this interaction was not found from our data perhaps because the range of speeds examined was so small.

2.4.3 Effect of Speed on Metabolic Rate

By collapsing across all three levels of BWS for each speed we realized a significant "Group X Speed" interaction effect (See Figure 2.3). Increases in speed from 0.5 to 1.5 km/hr were paralleled by a significant 13% and 24% increase in VO2 for the control and SCI groups alike, although increasing the speed from 1.0 – 1.5 km/hr was not observed to increase oxygen consumption significantly in either group (8% and 6% increase for the control and SCI groups respectively). The only other significant increase in oxygen consumption related to speed was observed in the SCI group when the speed was increased from 0.5 to 1.0 km/hr (i.e. a 20% increase). These findings are consistent with the overwhelming amounts of literature relating increases in speed to increases in the metabolic rate of oxygen consumption while walking.

The large discrepancies in oxygen consumption between the two groups, across all three speeds revealed that the SCI group was working much harder at each speed than the control subjects were. In fact, the VO2 for the SCI group while walking at 0.5 km/hr was 35% higher ($p < 0.05$) than that for the control group walking three times faster (i.e. 1.5 km/hr). The abnormally high VO2's for the SCI group can be attributed to their gait abnormalities as discussed earlier.

2.4.4 Metabolic Cost

It is instructive to compare energy expenditures per unit distance between able-bodied individuals and those with gait abnormalities in order to get a better appreciation of the inefficiencies between the two groups (Inman, Ralston, & Todd, 1981). Our results clearly indicate that the SCI group was much less efficient than the control group for all nine conditions, suggesting that the SCI group has a very large imbalance between the amount of oxygen consumed and the quantity of work performed.

Improvements in metabolic cost were possible through increases in speed and/or BWS, yet increasing speed was more effective at improving efficiency. For example, the metabolic cost for the SCI group while walking at 0.5 km/hr with 20% BWS was 1.22 ml/kg/m. Doubling the speed improved the efficiency by 74%, whereas doubling the amount of BWS (20 to 40%) only improved the efficiency by 21% while tripling BWS only improved the efficiency by an additional 10% (i.e. 20 –60%). Also noteworthy was the fact that the control group's metabolic cost was never observed to get below 0.25 ml/kg/m (1.5 km/hr @ 60% BWS). Thus, if we were to assume a metabolic efficiency of 0.16 ml/kg/m for an able-bodied person walking at 4.6 km/hr as reported by Waters & Mulroy (1999), it becomes evident that walking at 1.5 km/hr with 60% BWS is 56% more costly to walk one meter for an able-bodied individual. Furthermore, had we compared the metabolic cost of the SCI group while walking at 1.5 km/hr with 60% BWS (i.e. 0.42 ml/kg/m) to that of the able-bodied population walking at their customary speed (i.e. 0.16 ml/kg/m while walking at 4.8 km/hr; Waters et al., 1988), it would have been found that the SCI group's metabolic cost was 163% or 2.6 times greater than the able-bodied population. Obviously this methodology is flawed and grossly overestimates the differences between the two groups. Thus, previous research using this paradigm should be evaluated carefully before judgments are made. Consequently, it is more appropriate to use identical walking conditions, if possible, when comparing the metabolic demands between two groups because slower speeds are inherently inefficient, even for healthy, able-bodied individuals as demonstrated in this study. Controlling for speed minimizes the discrepancies between the two groups, allowing for a more objective examination of the metabolic costs associated with the specific pathology of the patients and not that caused by the speed at which they are walking.

Unlike the modest effects of increasing BWS on improving efficiency, any increase in speed was observed to significantly improve the metabolic cost of walking.

This effect is most likely due to tight relationship between speed and metabolic cost as well as the fact that the speed – metabolic cost relationship is "U" shaped. Recall that metabolic cost is the ratio of oxygen consumed to speed (metabolic cost = VO2 / speed) and that slow (< 3.6 km/hr) and fast (i.e. > 6.0 km/hr) speeds of ambulation are inefficient because the oxygen consumed outweighs the work being done, whereas at mid-range speeds of locomotion, somewhere between 3.6 and 5.4 km/hr, an optimal metabolic cost is achieved whereby the oxygen consumed is well matched to the amount of work being done.

Using this same logic, the much higher metabolic cost of walking observed in the SCI group is a clear indication that these individuals are consuming abnormally high amounts of oxygen for the amount of work being done. However, in light of the fact that the very slow speeds also increased the metabolic costs for the control subjects compared with comfortable walking speeds, a proportion of the inefficiency observed in the SCI group may be attributed to the speed at which they were walking with the additional metabolic costs owing to their gait difficulties.

2.5 Conclusion

Unlike other studies, which have compared the metabolic demands of able-bodied individuals walking at their customary walking speeds to that of persons with incomplete SCI ambulating with assistive devices and/or at slower speeds, we have accounted for the inefficiency caused by slow speeds by performing a direct and systematic comparison between these two populations under 9 different conditions (3 levels of BWS and 3 speeds). The end result leaves us with a true representation of the metabolic demands of walking on the treadmills for individuals with incomplete SCI who have trained for ten months on the BWS treadmill.

Based upon the results of the present study, it can be concluded that increases in speed have a more profound effect on VO2 than do reductions in BWS for the spinal cord injured group. Increases in speed, accompanied by reductions in BWS were observed to cause an upward trend in VO2 for both groups. Thus, in order to increase the exercise intensity on a BWS treadmill, the speed can be increased or the level of BWS can be reduced, with greater gains coming from increases in speed. Presumably, a combination of the two would also be effective at modifying the exercise stimulus; the fact that a statistically significant interaction effect was not found in the present study was likely due to the small number of subjects in the SCI group. The metabolic cost data from the control group also provides evidence that pathologically slow speeds of ambulation are highly inefficient and that BWS can only slightly offset the inefficiency caused at these speeds.

Our results demonstrate that the metabolic demands are unquestionably higher for the SCI group, but at no time were they twice as high as the controls let alone several times higher. This result suggests that previous studies using different conditions for comparison have overestimated the discrepancies between these two populations. Nevertheless, the end result is consistent with other claims that the SCI population has greater metabolic demands while walking, but the magnitude of difference is somewhat smaller than previously published (Fisher & Gullickson, 1978 & Waters & Mulroy, 1999).

Speaking from an exercise standpoint, the metabolic rates achieved by the SCI participants while walking at 1.5 km/hr with 20% BWS would be equivalent to that of an able-bodied person walking at a brisk pace on level ground with no BWS. This may lead us to conclude that the metabolic demands of the SCI group while walking on the treadmills under these artificial conditions can provide an adequate exercise stimulus, and if accompanied by a healthy lifestyle, can guard against various health risks such as heart disease, diabetes, obesity, etc. Thus, our results are the first empirical evidence to support the notions made by Gardner et al., (1989) and Wernig and Muller (1992), that this form of rehabilitation can provide an effective cardiovascular stimulus for this population.

In conclusion, our results suggest several things. One, comparisons between able-bodied individuals and those with SCI should be done, if possible, under similar conditions in order to control for other sources of variance. Second, the SCI population is much less efficient than able-bodied individuals due to their various gait difficulties, however their inefficiency while walking on the treadmill may allow them to achieve an effective cardiovascular stimulus, equivalent to that of able-bodied persons walking at much greater speeds. Consequently, although the speeds used during BWSTT may appear to be very slow and would not have cardiovascular benefits for the able-bodied population, it may provide a protective effect against cardiovascular disease, diabetes, obesity and other health risks associated with a sedentary life style that are common in the SCI population.

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3.0 Influence of BWSTT For Individuals With Incomplete Spinal Cord Injury: EMG Profiles

Abstract

To date no study has made direct comparisons between the EMG patterns of able-bodied persons and individuals with an incomplete spinal cord injury (SCI) while walking using identical conditions i.e. speed, body weight support (BWS), with assistance, etc. Yet, it has been shown that the phasic nature of dynamic electromyography (EMG) while walking can be affected by BWS (Finch et al., 1990) and speed (Detrembleur et al., 1997; Murray et al., 1984; Shiavi et al., 1981; Shiavi et al., 1987 & Wooten et al., 1990) in able-bodied persons. Surface EMG from the right medial gastrocnemius (RMG) and right tibialis anterior (RTA) was collected on ten control subjects, free from any gait abnormalities and three individuals with ASIA "C" incomplete SCI, who had completed ten months (122 ± 1.0 sessions) of BWS treadmill training. All participants walked independently on a BWS treadmill under nine different test conditions (3 levels of body weight support: 20, 40 & 60% X 3 speeds: 0.5, 1.0 & 1.5 km/hr). The EMG signals from all subjects were digitized at 1024 Hz and stored for off-line data processing. The digitized EMG data was transformed into linear envelopes and as many gait cycles as possible were manually selected from each trial by means of a footswitch signal indicating heel-strike in order to generate individual ensemble averages (EA). The EA's were normalized to time by setting the stride period to 100% and normalized in amplitude by setting the peak amplitude for each stride to 100%. A grand

ensemble average (GEA) was generated by averaging across all the individual EA's from the control group. The similarity of each SCI participant's EA for both muscles and all nine conditions, was compared to that of the GEA, using a Pearson cross correlation. Only one subject had high correlations (i.e. ≥ 0.80), four of which were found for the RMG and only one for the RTA. The other two SCI participants did not have any high correlations but correlations for the RMG and RTA did appear to improve as BWS increased for two of the participants, whereas the activity of the third SCI participant showed very poor correlations for both muscles. A qualitative comparison between the GEA and that of each SCI participant's EMG profiles is discussed.

3.1 Introduction

Given the devastating neurological damage suffered by individuals with spinal cord injuries (SCI), it should come as no surprise that these individuals have been observed to adopt altered motor strategies (Leroux, Fung & Barbeau 1999 & Lewko, 1996) while walking at pathologically slow speeds relative to the able-bodied population (Dietz, Columbo & Jensen, 1994; Dietz, Columbo, Jensen & Baumgartner, 1995; Dietz, Nakazawa, Wirz, & Ernie 1999; Fung & Barbeau, 1989; Leroux et al., 1999 & Protas et al., 2001). Numerous studies have identified the viability of body weight support treadmill training (BWSTT) for partially restoring and/or improving the ambulatory gait patterns of individuals with incomplete SCI (Dietz et al., 1995; Wernig & Muller, 1992; Wernig, Muller, Nanassy, & Cagol, 1995; and Wernig, Nanassy, Muller, & Cagol, 1998) while walking over ground (Calancie et al, 1994; Wernig & Muller 1992; Wernig et al., 1995 & Wernig et al., 1998) as well as on treadmills (Calancie et al., 1994; Fung and Barbeau 1989; Harkema et al., 1997; Leroux et al., 1999 & Wernig & Muller 1992; Wernig et al., 1995 & Wernig et al., 1998). In addition, cyclical EMG burst patterns have been recorded from individuals with incomplete SCI while walking over ground (Calancie et al, 1994; Wernig & Muller 1992) and on treadmills with (Ditz et al., 1994; Dietz et al., 1995; Dietz et al., 1999; Fung & Barbeau, 1989 & Leroux et al., 1999) and without body weight support (BWS) (Dietz et al., 1994; Dietz et al., 1995 & Fung & Barbeau, 1989).

However, EMG profiles from the tibialis anterior (TA) and gastrocnemius (GA) of persons with SCI while walking on or off the treadmills, and/or with or without BWS, have numerous phasic and/or amplitude abnormalities relative to those from able-bodied individuals, including but not limited to: premature, late, prolonged and/or continuous muscle activation relative to critical gait events (Calancie et al., 1994; Dietz et al., 1995; Dietz et al., 1999; Fung & Barbeau, 1989; Leroux et al., 1999; McKay, Metman, Dimitrijevic, Sherwood & Dimitrijevic, 1983); absence of activity (McKay et al., 1983); poorly modulated and/or low amplitudes (Dietz et al., 1995; Dietz et al., 1999; Leroux et al., 1999; McKay et al., 1983); clonus (Fung & Barbeau, 1989; McKay et al., 1983); and/or coactivation (Fung & Barbeau, 1989; Leroux et al., 1999; McKay et al., 1983). Moreover, greater between-subject variability has also been reported for the SCI population (Fung & Barbeau, 1989) relative to able-bodied controls.

All of the above comparisons however, were conducted using different conditions (i.e. speed, BWS & assistance from a therapist) between the SCI group and the able-bodied controls. Yet, the phasic nature (Detrembleur, Willems, & Plaghki, 1997; Murray, Mollinger, Gardner, & Sepic,1984; Shiavi, Champion, Freeman & Griffin, 1981; Shiavi, Bugle, & Limbird, 1987 & Wooten, Kadaba, & Cochran, 1990) and amplitude (Brandell, 1977; Kameyama, Ogawa, Okamoto, & Kumamoto, 1990; Milner, Basamajian, & Quanbury 1971; Murray et al., 1984; Yang & Winter 1985 & Winter 1983) of dynamic EMG recordings in able-bodied individuals reportedly vary as walking speed changes. The EMG amplitude (Colby, Kirkendall & Bruzga, 1999 & Finch, Barbeau & Arsenault, 1990) and pattern (Finch et al., 1990) is also affected by BWS. Thus, since both speed and BWS have been shown to independently affect both EMG amplitude and temporal pattern, it is important to control for these potentially significant sources of variance. It is also impossible to isolate the EMG caused by the passive stretch of the ambulatory musculature by a therapist from that produced by voluntary contractions (Rossignol & Barbeau, 1995 & Stewart, Barbeau & Gauthier, 1991). Furthermore, slower speeds are also associated with greater stride-to-stride and between subject variability (Shiavi et al., 1981 & Yang & Winter, 1985), which may account for a proportion of the variability observed in the SCI groups' EMG recordings. Given all of the above sources of variance that were not controlled for in previous studies when comparing between the SCI population and able-bodied controls, great caution must be employed when interpreting these results.

It has been shown that the irregularities in the EMG amplitude and phase during SCI gait, can be reduced with BWSTT (Dietz et al., 1995). BWS reportedly decreases the stretch of the gastrocnemius during stance, which may facilitate proper activation of this muscle in persons with gait abnormalities (Finch et al., 1991), allowing for a more fluid rocker-gait. Proper timing of muscle patterns during gait is also very important in order to produce a fluid, coordinated and efficient walking pattern. BWSTT allegedly facilitates the recovery of locomotion by allowing muscle output patterns to be similar to those observed in able-bodied individuals, which may enhance the probability of regaining full weight bearing activities (Barbeau & Rossignol, 1994).

To date, no studies have made direct comparisons of the dynamic EMG of the lower leg musculature of able-bodied persons to that of individuals with incomplete SCI who have received extensive BWSTT, while walking independently under identical combinations of BWS and speed. Therefore, given our present understanding of the effects that BWS and speed have on the temporal and amplitude characteristics of EMG, future studies attempting to determine the "normalcy" of gait EMG in SCI persons should use identical conditions for the control group as the SCI participants.

3.1.1 Purpose

The purpose of this paper is to describe and compare the dynamic EMG profiles of individuals with incomplete SCI who have completed ten months of BWSTT, to that of able-bodied individuals under varying combinations of BWS and speed while walking on a treadmill. Moreover, it was our intent to examine the motor patterns of the SCI subjects to determine if they exhibit a cyclical gait pattern, which is consistent and/or similar to that of the control group.

3.2 Methods

3.2.1 Subjects

Three individuals with ASIA "C" incomplete SCI, who had completed ten months (122 ± 1.0 sessions) of thrice-weekly BWSTT were recruited from a pool of participants in a 12-month BWSTT study at McMaster University, Ontario (See Table 3.1 for the detailed SCI participant information). In order to be included in the study, the SCI participants had to be able to walk independently (i.e. move both their legs free from the influence, guidance or assistance from external sources) on the treadmill for twelve consecutive minutes at 1.5 km/hr with 40% BWS. Although some of the SCI participants were able to ambulate on the treadmills with no BWS, this was not a criterion for inclusion.

Ten able-bodied male and female control participants (five male and five female) volunteered to take part in the present study. Their ages ranged from 22 to 30 (mean 25.6 ± 2.6) years, their heights ranged from 152 – 180 (mean 167.8 ± 8.6) cm and their weights ranged from 50.0 to 88.6 (mean 71.1 ± 13.5) kilograms. None of the control subjects had any musculoskeletal problems and/or gait abnormalities.

This study received ethical clearance from the McMaster Research Ethics board. All participants read and gave their written consent to participate after reading the information letter outlining the procedure, risks and benefits of the study.

Table 3.1: SCI Subject information. All numerical values are the mean ± one standard deviation.

Subject	Age (yrs)	Weight (kg)	Height (cm)	Session Tested	Gender	Cause of Injury	Lesion Level	ASIA Score	Yrs. Post Injury
SCI 01	24	103.2	190.5	123	M	Birth	C5-6	C	24
SCI 02	32	107.7	185.4	122	M	Knife	C4	C	2.5
SCI 03	35	88.6	190.5	121	M	MVA	C4-6	C	21
Mean ±SD	**30.3 ±5.7**	**99.8 ±10.0**	**188.8 ±2.9**	**122 ±1.0**					**15.8 ±11.6**

3.2.2 Experimental Design and Procedure

The present study was a three-factor (3 levels of BWS X 3 speeds X 2 groups) mixed design (2 within, 1 between), with fixed effects (See Table 3.2 for the design). The three levels of BWS used were 20, 40 and 60% and the three speeds employed were 0.5, 10. & 1.5 km/hr (8.3, 16.7 & 25 m/min, respectively). These levels of BWS and speeds were selected because they are representative of the conditions commonly used during BWSTT in individuals with incomplete SCI and all of the subjects with SCI had previously experienced walking under these conditions.

Table 3.2: Summary table of the three-factor, fixed effects, mixed design (3 Speeds X 3 BWS X 2 Groups)

	Speed 1 (0.5 km/hr)			Speed 2 (1.0 km/hr)			Speed 3 (1.5 km/hr)		
	20% BWS	40% BWS	60% BWS	20% BWS	40% BWS	60% BWS	20% BWS	40% BWS	60% BWS
Controls N = 10									
SCI N = 3									

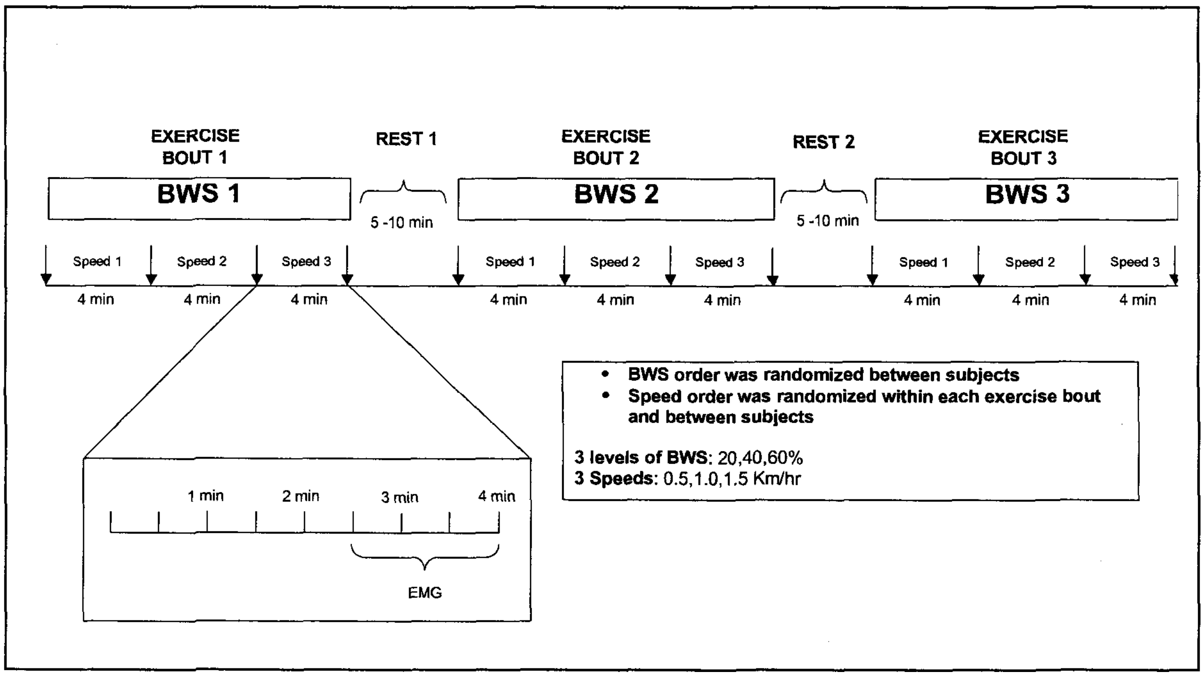

Figure 3.1: Schematic illustration of data collection procedure and time line.

Electromyography (EMG) was collected on all participants while walking independently on a BWS treadmill (Woodway Lokosystem), under nine different test conditions (3 levels of body weight support X 3 speeds). The treadmills are equipped with an overhead hoist system, which works with counter weights of varying mass to provide numerous levels of BWS and has a control panel, with digital display, to vary speed. The body weight support cables were fastened to a harness donned by all participants. In order to ensure that the prescribed level of BWS was maintained throughout each condition, the participants were encouraged not to use the handrails for support (yet were allowed to use them for balance).

Each experimental session consisted of three, twelve-minute bouts of exercise, separated by five to ten minutes of rest to prevent fatigue. Each bout of exercise occurred at a randomly different, yet constant level of BWS, while the speed of the treadmill was randomly changed every four minutes within each exercise bout (See Figure 3.1). Thus, BWS was randomized between subjects while speed was randomized within each bout of exercise and between subjects. The speed was held constant for four minutes in order for the participants to establish a consistent gait pattern.

3.2.3 EMG Data Collection

Raw EMG was recorded simultaneously from the right medial gastrocnemius (RMG) and right tibialis anterior (RTA) using a Bagnoli-8 EMG System (Delsys Inc., DS-80) during the last 90 seconds of each of the nine walking conditions (See Figure 3.1). These muscles were selected because of their functional role during gait and ease of accessibility. The electrode for the RMG was placed over the area of greatest muscle bulk on the medial calf, whereas the electrode for the RTA was placed over the area of greatest muscle bulk just lateral to the crest of the tibia on the proximal half of the leg. The orientation of the electrodes was such that they were placed along the axis of the muscle fibers similar to that used by Winter & Yack (1987). The reference electrode was positioned over a neutral and inactive site away from the EMG muscle sources. Prior to electrode placement, the skin was prepared using conventional techniques by lightly abrading and wiping the skin clean with isopropyl alcohol to lower skin impedance, improve electrode adhesion and ensure the proper recording of the muscle action potentials. In certain instances, it was necessary to remove excess hair using a disposable razor prior to electrode placement.

Each channel of EMG was differentially amplified and had an output impedance of 15 kΩ. Each active differential surface electrode is internally shielded to reject ambient electrical noise and is comprised of two, 1mm diameter by 10mm long silver bars, spaced 1 cm apart for optimal signal capture (see Delisa, 1998; pg 37 Figure 5a). Each signal was differentially pre-amplified by a factor of ten at the electrode site (CMRR of 92 dB). Each signal was bandpass filtered (20Hz –450 Hz) and independently amplified again by an additional factor of 10, 100 or 1000 at the bio-amplifier. The gain setting on the bio-amplifier was selected in order to provide the greatest signal gain and fidelity possible prior to digitization. Signal clipping was prevented by means of a warning buzzer emitted by the system when any channel was over-amplified (i.e. greater than ± 4.8 volts), thus prompting the experimenter to reduce the gain of any affected channel(s).

An additional channel was available to collect the signal from a pressure sensitive switch placed under the right heel which, served to disclose the gait cycle defined as the event occurring in time from right-heel contact to right-heel contact (RHC – RHC). The footswitch was 1mm thick (MA-153 standard footswitch by Motion Lab Systems Inc.) and was taped to the sole of the right heel of each participant using Transpore® surgical tape. The footswitch consisted of a 25mm diameter membrane with a 15mm sensor area and responded to 50 grams of pressure anywhere in the sensor area. After the switch was fastened to the sole of the foot, the subjects put their shoes and socks back on. Both of the EMG signals and the footswitch signal were digitized in real-time at a rate of 1024 Hz (16 bits A/D) using a generic Pentium computer. The digitized data was stored on the local hard drive for future off-line signal processing.

3.2.4 EMG data processing

The digitized EMG signals were further processed using proprietary software written at McMaster University. Each channel of EMG was full wave rectified (FWR) and low-pass filtered (fc = 3.0 Hz) to produce a linear envelope using an absolute value function and a second order, Butterworth filter respectively. The low-pass filtering was performed forwards and backwards in time to prevent any time lags in the signal. The data set was then further reduced by a factor of 25 (40.96 Hz) in order to facilitate data storage, manipulation and management.

Individual strides were manually located and selected using another proprietary software program using the right heel contact footswitch signal across the full 90-second record. Thus, the muscular activity from the RMG and RTA were "cycled" based upon the manually located foot switch data. Certain strides were not selected because of occasional footswitch malfunction or irregular footswitch patterns. The software outputted various temporal gait parameters such as the average stride length and duration, whereas the cadence (strides/min) was calculated using the mean stride duration.

The "cycled" stride-to-stride EMG signals were then rubber banded using a cubic spline function in order to ensure an equal number of frames in each stride. After which, each stride period (RHC - RHC) was normalized to time by setting the stride period to 100%, which allowed for pooling of data across strides and subjects. Each stride was then averaged into 2% intervals over the entire stride and for all cases right heel contact occurred at 0% of the stride, with the subsequent heel strike occurring at 100%. In order to average the EMG data across strides and perform phasic EMG comparisons between subjects, each time normalized stride was then normalized in amplitude by setting the peak EMG value of each stride to 100%. (See Figure 3.2 for a flow chart of the EMG data collection and processing steps). The Intra-subject or individual ensemble-averages were generated by averaging across all of the selected strides and provide a good representation of muscular activity across the stride period (Arsenault, Winter & Marteniuk, 1986a) (See Table 3.3 for the number of strides used in each of the individual SCI ensemble averages).

The entire filtering, processing and normalization procedure rendered a linear envelope EMG profile representing the dynamic muscle activity in time, expressed as a percentage of the gait cycle beginning with right heel contact (0%) and amplitude expressed as a percentage of the peak amplitude. This amplitude normalization procedure reportedly reduces inter-subject variability, yet removes any amplitude differences between subjects because each subject's EMG has the same "weight" (Winter and Yack, 1987). Therefore, no amplitude sensitive characteristics (e.g. average, peak activity, etc.) of the EMG signal can be compared between subjects.

Table 3.3: Number of strides used for the control subject grand ensemble average (GEA mean ± 1SD) and each of the spinal cord injured ensemble averages for all nine conditions tested. N/A, indicates that data is not available: SCI Subject 03 did not complete any of the 20% BWS conditions. † Indicates N=9; whereas N =10 for all other control conditions.

		20% BWS			40% BWS			60% BWS		
		0.5 km/hr	1.0 km/hr	1.5 km/hr	0.5 km/hr	1.0 km/hr	1.5 km/hr	0.5 km/hr	1.0 km/hr	1.5 km/hr
Control	GEA	28 ± 8	43 ± 7	53 ± 8	24 ± 7	48 ±8	55 ± 6	23 ± 9	42 ± 15	† 56 ± 13
SCI	01	14	29	34	17	30	36	19	34	39
	02	20	30	40	18	31	41	20	31	42
	03	N/A	N/A	N/A	24	43	51	24	44	50

The normalized intra-subject ensemble-averages were averaged across the control group (i.e. pooled) to render a grand ensemble average (GEA). The GEA was produced in order to qualitatively describe and compare the phasic muscular activity patterns of the control group to the individual ensemble averages generated from each of the SCI participants. Phasic EMG comparisons between the 18 control group GEA's (2 muscles X 9 conditions) and the individual ensemble averages for the SCI participant's were conducted using a Pearson cross correlation. This analysis renders a correlation coefficient (r) providing an index of similarity of the phasic activity between the two EMG profiles compared, with values closer to 1.0, being more similar in phase. This procedure is similar to that used by Aresenault, Winter and Marteniuk (1986b) and Yang

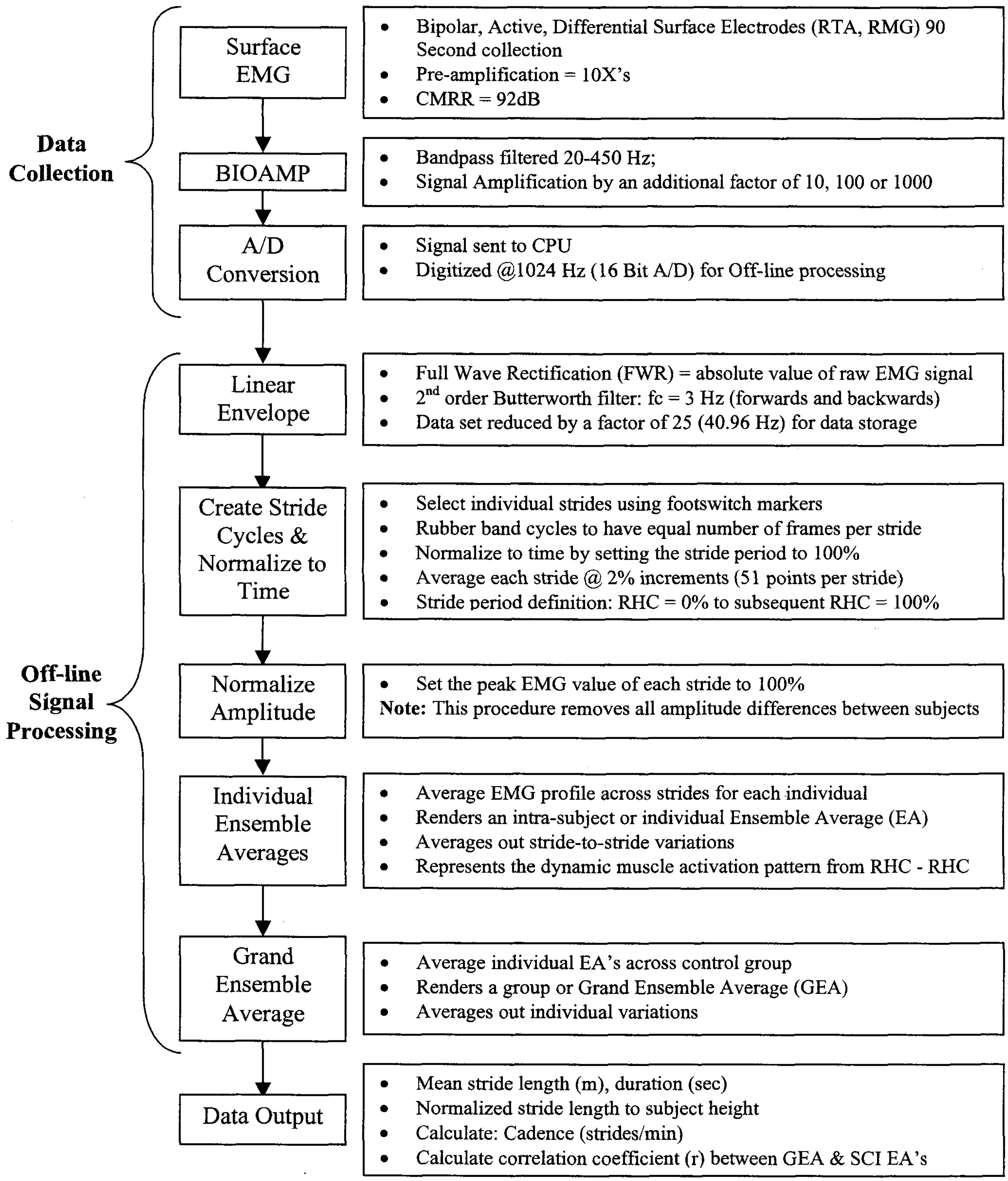

Figure 3.2: EMG data collection and Processing Flow Chart. See text for details

and Winter (1985). Correlation coefficient values greater than or equal to 0.80 were considered to indicate good phasic similarity. The 0.80 benchmark value was arbitrarily set for discussion purposes only.

3.3 Results

3.3.1 Pearson Correlation Coefficients

Table 3.4 displays the individual Pearson correlation coefficients (r) for each condition between the GEA and the ensemble-averages (EA) for the RMG and RTA for each of the SCI subjects. Only five conditions had coefficients greater than 0.80 (indicated by an asterisk "*") and all came from the same SCI participant (i.e. SCI 02), four of which were from the RMG and only one from the RTA.

All three of the speeds examined with 60% BWS for SCI participant 02 resulted in coefficients greater than 0.86 for the RMG whereas the only coefficient greater than 0.80 for the RTA was found at 60% BWS while walking at 1.5 km/hr (r = 0.85). The coefficients for the other two SCI participants (SCI 01 and 03), were all less than 0.80, with all but two of the coefficients for SCI 01 being greater than zero, and all but two of the coefficients for SCI 03 being negative. The two negative coefficients for SCI 01 occurred for both muscles under the same condition (i.e. 20% BWS at 0.5 km/hr), whereas the only two positive coefficients for SCI 03 occurred for both muscles while walking at 1.5 km/hr with 40% BWS.

In general, the coefficients for both muscles for SCI 01 and 02 increased as BWS increased. On the other hand, increases in speed were not observed to have the same trend on the coefficients.

3.3.2 Muscle Activity Patterns

Figures 3.3 – 3.5 show the muscle activity for both muscles across all speeds for the 20, 40 and 60% BWS conditions respectively. Each figure has six graphs lettered A – F, each showing the muscle activity over the period of one stride starting at RHC (i.e. 0%) and terminating at the subsequent RHC (i.e. 100%). The left column is the muscle activity from the RMG at the three speeds examined, starting at 0.5 km/hr (i.e. graph A) descending sequentially to graph C (i.e. RMG at 1.5 km/hr), whereas graphs D – F are in the right column and are the linear envelopes from the RTA, again starting at 0.5 km/hr

Table 3.4: Individual Pearson correlations (r) for each condition between the GEA and all three of the SCI participants' EA for the right medial gastrocnemius (RMG) and right tibialis anterior (RTA). The asterisk "*" indicates a Pearson coefficient greater than 0.80. SCI participant 03 did not complete any of the 20% BWS conditions and are indicated as N/A.

		SCI 01		SCI 02		SCI 03	
BWS (%)	Speed (km/hr)	RMG (r)	RTA (r)	RMG (r)	RTA (r)	RMG (r)	RTA (r)
	0.5	-0.15	-0.38	0.88 *	0.68	N/A	N/A
20	1.0	0.43	0.21	0.67	0.52	N/A	N/A
	1.5	0.31	0.23	0.50	0.59	N/A	N/A
	0.5	0.63	0.48	0.76	0.69	-0.68	-0.56
40	1.0	0.30	0.16	0.72	0.62	-0.87	-0.85
	1.5	0.57	0.08	0.59	0.70	0.17	0.35
	0.5	0.73	0.50	0.95 *	0.73	-0.68	-0.69
60	1.0	0.63	0.55	0.91 *	0.73	-0.90	-0.79
	1.5	0.77	0.09	0.87 *	0.85 *	-0.56	-0.56

(i.e. graph D) descending to graph F (i.e. RTA at 1.5 km/hr). Each graph within the figure shows the grand ensemble average plus-minus one standard deviation (GEA ± 1 SD = thick-black line ± thin-black lines) and each of the SCI participants' individual EA's across a stride. SCI participant 03 did not complete any of the 20% BWS conditions, whereas all three SCI participants completed all speeds while walking at 40 and 60% BWS.

3.3.2.1 Grand ensemble average

The GEA represents the average muscle activity across all ten control-subjects for each condition except for one condition (60% BWS at 1.5 km/hr) where data from only nine subjects data were available. The peak activity of the RMG-GEA for all conditions occurred between 30 and 50% of the gait cycle depending on the level of BWS and/or speed. In general, for all nine conditions, the activity pattern of the RMG was mono-phasic with the activity steadily increasing soon after heel-strike, rising to a peak presumably at or near heel-off, and then tapering off before swing. However, as speed increased the peak activity was observed to occur somewhat earlier in the gait cycle with the rise and fall of activity being more rapid resulting in a more well-defined pattern of activity during mid-stance compared to a wider, less distinct burst of activity at slower speeds. A similar trend was observed as BWS increased, with wider, less distinct burst patterns occurring at higher levels of BWS.

The activity of the RTA-GEA was observed to be typically bi-phasic, with the first peak of activity occurring at heel-strike (i.e. 0%) and the second peak occurring at approximately 60-70% of the gait cycle for all nine conditions. Wider areas of activity were observed at slower speeds and at higher levels of BWS. The RTA activity onset occurred earlier in the gait cycle at faster speeds. Furthermore, at the highest level of BWS at 1.0 and 1.5 km/hr, the RTA onset also appeared to be earlier in the stride cycle than at lower levels of BWS for the same speed.

3.3.2.2 Ensemble averages for SCI participant 01

In general, the muscle activity recorded from the RMG of SCI participant 01 occurred early, with the primary peak of activity occurring somewhere between 10-20% of the gait cycle and in some instances exhibited a bi-phasic pattern with a second peak occurring later on, just prior to heel-strike (i.e. ~ 100%). Only on two instances, while walking at 0.5 km/hr with 40 and 60% BWS respectively, did the primary burst from the RMG occur at approximately 40% of the stride.

The RTA pattern was mono-phasic for all nine conditions and was mostly quiescent from heel-strike (0%) up until approximately 60% of the gait cycle. This singular burst was generally very narrow, with a very rapid rise and fall in activity and lasted for only about 30% of the cycle.

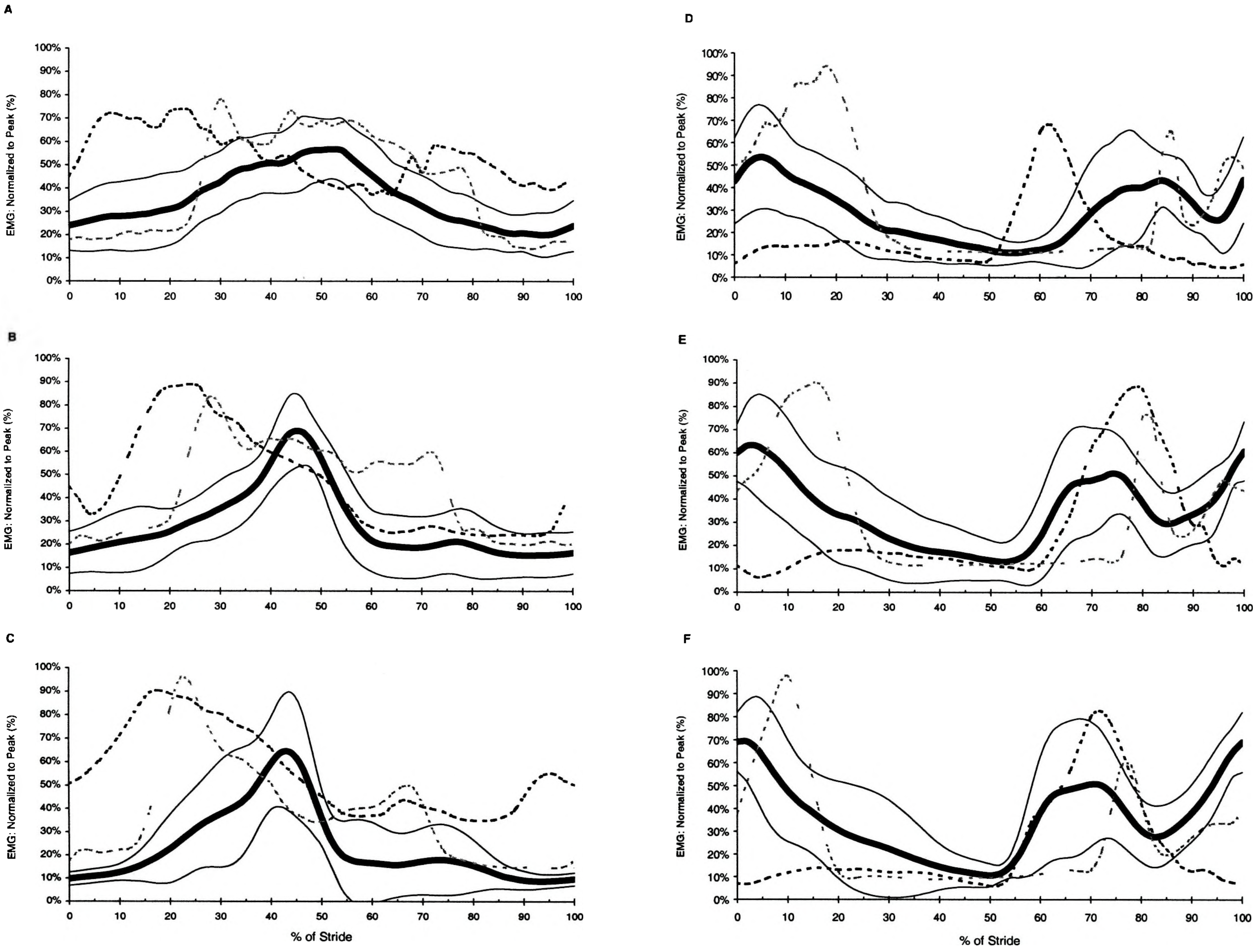

Figure 3.3: GEA ± 1 SD (▬▬ ± ——) and individual EA's for **SCI 01** (······) and **SCI 02** (·····) while walking with **20% BWS** for the medial gastrocnemius **A)** 0.5 **B)** 1.0 & **C)** 1.5 km/hr and tibialis anterior **D)** 0.5 **E)** 1.0 & **F)** 1.5 km/hr.

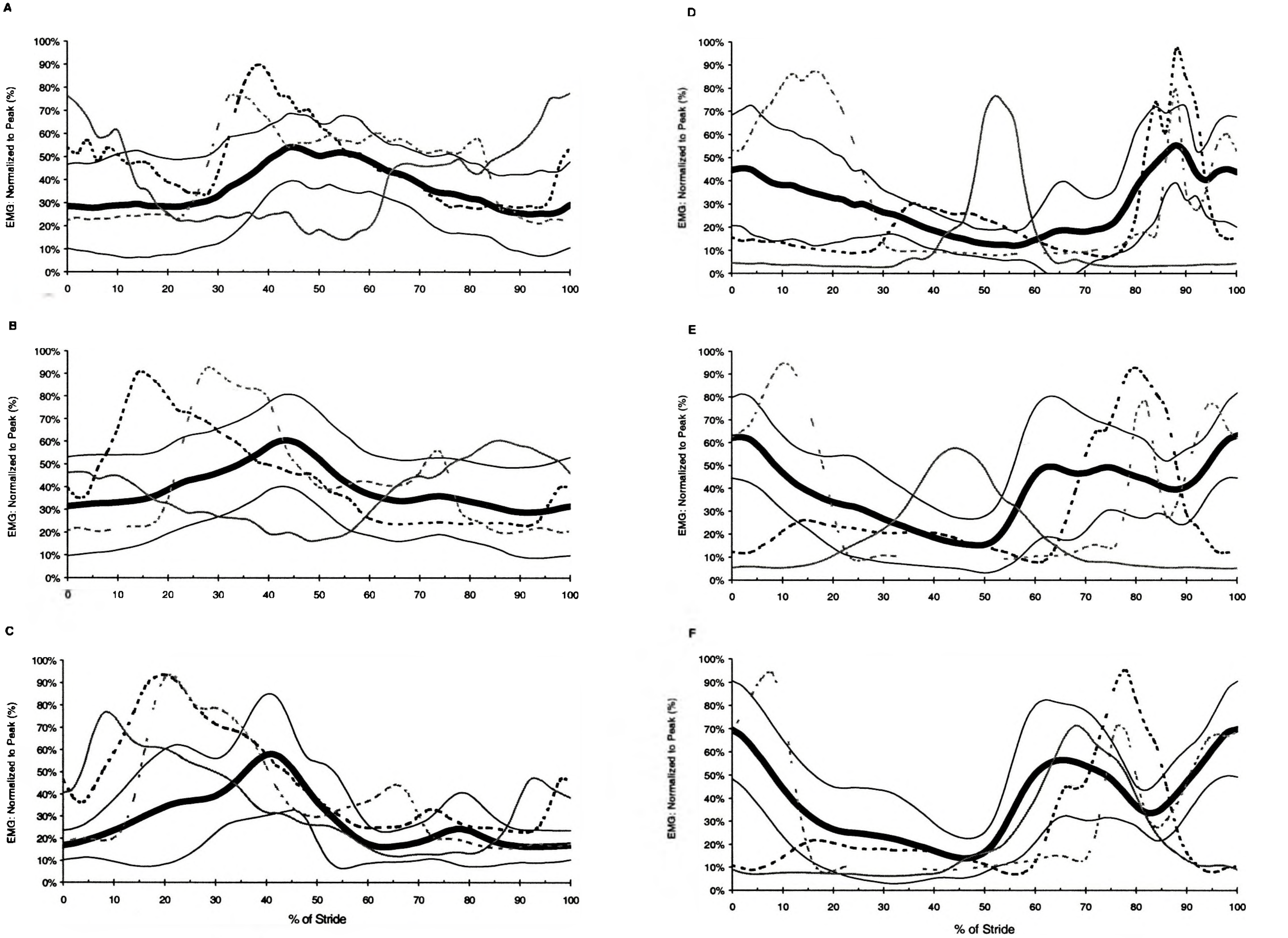

Figure 3.4: GEA ± 1 SD (▬ ± ——) and individual EA's for **SCI 01** (......), **SCI 02** (....) & **SCI 03** (——) while walking with **40% BWS** for the medial gastrocnemius **A)** 0.5 **B)** 1.0 & **C)** 1.5 km/hr and tibialis anterior **D)** 0.5 **E)** 1.0 & **F)** 1.5 km/hr

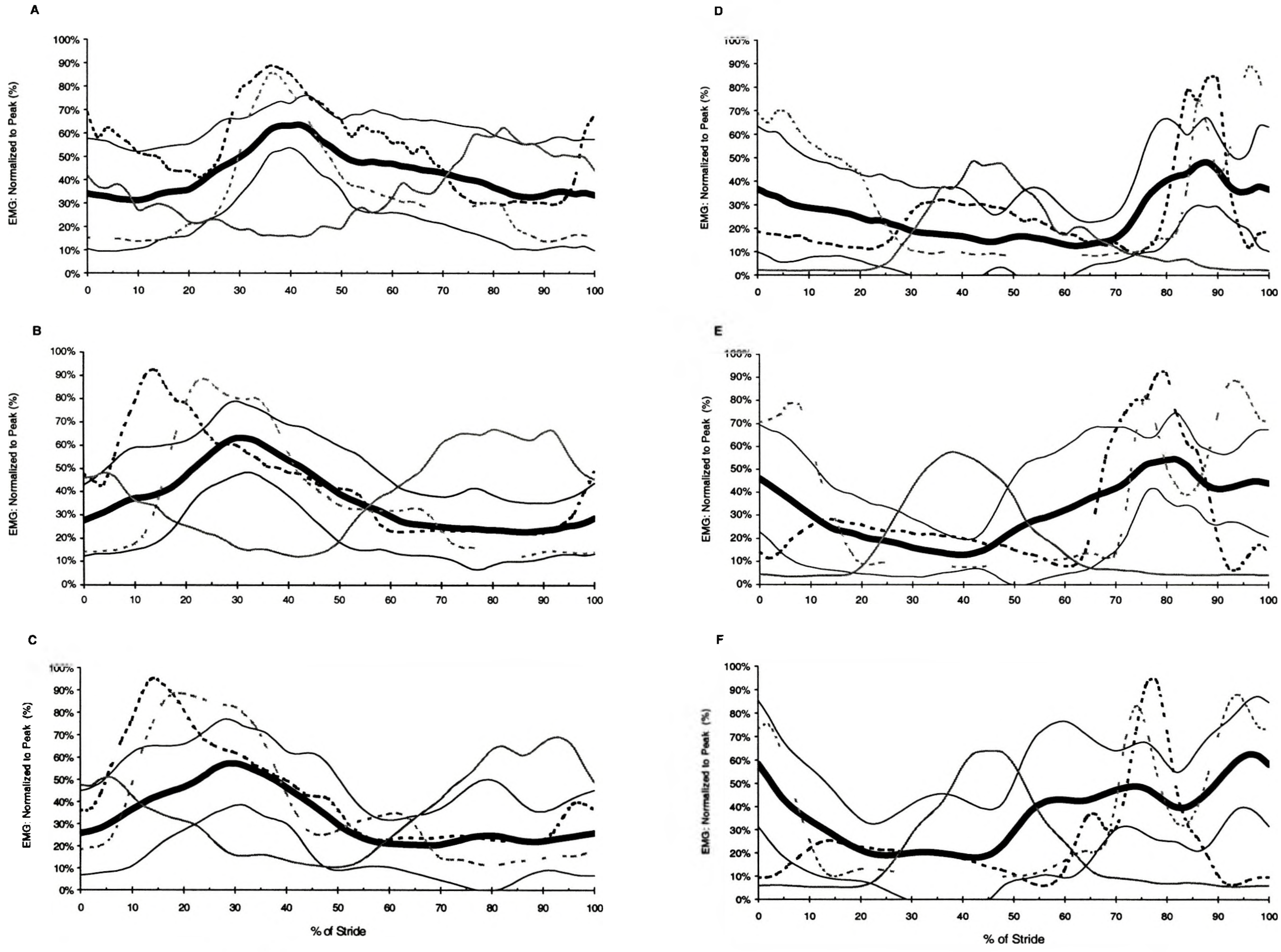

Figure 3.5: GEA ± 1 SD (▬ ± ——) and individual EA's for **SCI 01** (······), **SCI 02** (······) & **SCI 03** (——) while walking with **60% BWS** for the medial gastrocnemius **A)** 0.5 **B)** 1.0 & **C)** 1.5 km/hr and tibialis anterior **D)** 0.5 **E)** 1.0 & **F)** 1.5 km/hr

3.3.2.3 Ensemble averages for SCI participant 02

The RMG activity of SCI participant 02 can be characterized as biphasic for seven of the nine conditions. The muscular activity at heel-strike was minimal in all conditions, rising rapidly upwards to a peak at approximately 20-30% of the cycle, then decreasing slightly to a plateau and in some instances rising back up slightly into a second peak of activity, prior to going back down to a minimum at approximately 70-80% of the stride. As the speed increased, the activity was observed to start earlier yet end sooner. As BWS increased the RMG activity was also observed to start earlier but end sooner, thus maintaining the overall duration of the RMG throughout the gait cycle across all nine conditions to approximately 60%.

The RTA activity for this participant was tri-phasic for eight of the nine conditions with the first peak occurring anywhere between 0 and 20%, the second around 70-90% and the third at approximately 95% of the cycle. For the most part, the RTA activity was very consistent between conditions with speed shifting the first and second peaks to the left (i.e. earlier in the stride) with a very consistent period of low-level activity between the first and second bursts.

3.3.2.4 Ensemble averages for SCI participant 03

The third SCI participant's RMG activity exhibited numerous small fluctuations (high-frequency component) superimposed upon an overall low-frequency, bi-phasic level of activity across the stride as well as large variations from one condition to the next. In general, the activity was observed to begin anywhere between 45-85% of the stride, rising steadily towards a peak of activity between 70 and 100%, and gradually tapering off to its lowest level of activity anywhere from 20-60% after heel-strike. The durations of low-level activity across all nine conditions was less than 40% of the cycle, but typically lasted for about 20% of the stride.

The RTA activity displayed very distinct singular bursts of activity during mid cycle, starting anywhere from 20-40% and ending around 60-90%, with the peak of activity occurring around 37-67% of the stride for all nine conditions. Consistently smooth, low levels of activity were observed prior to and after the only burst of activity in this muscle.

3.3.3 Gait Variables

Table 3.5 summarizes all of the gait variables, mean stride length (m), mean stride length normalized to height, mean stride duration (sec), and mean stride rate (strides/min), for the control group and each of the three SCI participants. The control values are the group mean (N = 10), plus-minus one standard deviation for all conditions except one (1.5 km/hr with 60% BWS), where data from only nine controls was available. The individual values from the SCI participants are the mean value across strides only; SCI participant 02 did not complete any of the 20% BWS conditions and is indicated as not available (N/A).

In general, stride length and cadence tended to increase, whereas stride duration tended to decrease as the speed of ambulation increased for the control group and for each of the SCI participants. The stride lengths for SCI participants 01 and 02 were always longer than that of the mean stride length for the control group. On the other hand, the stride length for SCI participant 03 was less than that of the control group for three of the six conditions he participated in. The normalized stride length data for SCI participants 01 and 02 were also all less than that of the mean stride length for the controls with the exception of one condition for SCI 01 and SCI 02, in which their stride length was equal to and less than that of the mean stride length of the controls respectively. The normalized stride length data for SCI 03 tended to be less than that of the controls for five of his six conditions.

The stride durations and rates for SCI participants 01 and 02 were longer and smaller respectively than that for the control group for all nine conditions. SCI participant 03 had shorter stride durations for three conditions and a faster stride rate for one condition compared to the mean values for the controls. The stride rate for three other conditions was equivalent between SCI participant 03 and the controls.

In general, increases in BWS had no effect on stride length (normalized or absolute), or stride rate, yet a small increase in stride duration was observed for the control group. The effect of BWS on the gait parameters of the SCI participants was more variable. An increase in BWS tended to decrease the stride length for all three SCI participants but more dramatically for SCI 01. A similar decrease in normalized stride length was observed in SCI participants 01 and 02, whereas there was no change for SCI participant 03. As BWS was increased, stride duration decreased in all three SCI participants, whereas increases in BWS were observed to increase stride rate only for SCI participant 01.

Table 3.5: Gait variables: Stride length (m), Stride length normalized to height, stride duration (sec) and stride rate (strides/min). Control values are the group mean ± 1SD (i.e. collapsed across all 10 control subjects), whereas the SCI values are the mean only. SCI Subject 03 did not complete any of the 20% BWS conditions (indicated with N/A). † Indicates N=9; N = 10 for all other control conditions

			20% BWS			**40% BWS**			**60% BWS**		
			0.5 km/hr	**1.0 km/hr**	**1.5 km/hr**	**0.5 km/hr**	**1.0 km/hr**	**1.5 km/hr**	**0.5 km/hr**	**1.0 km/hr**	**1.5 km/hr**
Stride Length (m)	**Con**	**X**	0.43 ±0.11	0.57 ±0.1	0.71 ±0.09	0.53 ±0.15	0.53 ±0.11	0.67 ±0.08	0.55 ±0.13	0.63 ±0.19	† 0.69 ±0.16
	SCI	**01**	0.82	0.85	1.07	0.71	0.80	1.03	0.64	0.72	0.96
		02	0.59	0.82	0.92	0.63	0.77	0.90	0.60	0.78	0.86
		03	N/A	N/A	N/A	0.49	0.57	0.73	0.46	0.57	0.74
Stride Length (normalized to Height)	**Con**	**X**	0.26 ±0.07	0.34 ±0.07	0.42 ±0.05	0.32 ±0.09	0.31 ±0.06	0.40 ±0.05	0.33 ±0.08	0.38 ±0.11	0.37 ±0.15
	SCI	**01**	0.43	0.45	0.56	0.37	0.42	0.54	0.34	0.38	0.50
		02	0.32	0.44	0.50	0.34	0.42	0.48	0.32	0.42	0.46
		03	N/A	N/A	N/A	0.26	0.30	0.38	0.24	0.30	0.39
Stride Duration (sec)	**Con**	**X**	3.10 ±0.82	2.04± 0.37	1.68 ±0.21	3.77 ±1.06	1.88 ±0.39	1.58 ±0.19	3.92 ±0.92	2.25 ±0.68	1.64 ±0.39
	SCI	**01**	5.84	3.04	2.54	5.06	2.87	2.45	4.58	2.58	2.27
		02	4.21	2.91	2.20	4.51	2.76	2.14	4.29	2.80	2.05
		03	N/A	N/A	N/A	3.53	2.02	1.73	3.26	2.02	1.75
Stride Rate (strides/min)	**Con**	**X**	20 ±5	30 ±6	36 ±5	17 ±5	33 ±5	38 ±5	16 ±6	29 ±10	34 ±15
	SCI	**01**	10	20	23	12	21	24	13	23	26
		02	14	20	27	13	22	28	14	21	29
		03	N/A	N/A	N/A	17	29	34	18	29	34

3.4 Discussion

The present study was designed to perform direct comparisons between the phasic muscular activity of a group of SCI participants who had completed ten months of thrice weekly BWSTT to that of an able-bodied population walking under the identical conditions. Walking involves the cyclical activity of the lower limb musculature in order to produce rhythmical joint moments about the hip, knee and ankle. The moments of force about all joints must be appropriately sized and timed in order for the gait pattern to be fluid, coordinated and efficient. Thus, any pathology, or injury affecting the appropriate phasic and/or the magnitude of activation of the primary gait musculature will have a dramatic effect on the fluidity of ambulation.

Due to the complex nature of spinal cord injuries and the difficulties with EMG gait analysis in this special population, many studies have had to compare the muscular activity of controls to that of the SCI population while walking under different conditions such as speed, BWS and/or assistance from a therapist, assistive device or orthotics (Leroux, et al., 1999; Fung and Barbeau, 1989). However, both speed (Detrembleur et al., 1997; Murray et al., 1984; Shiavi et al., 1981; Shiavi et al., 1987 & Wooten et al., 1990) and BWS (Finch et al., 1990) have been shown to affect the phasic nature and amplitude of able-bodied EMG patterns while walking. Moreover, Harkema et al. (1997), also found that EMG amplitude was affected by BWS in individuals with incomplete and complete SCI. Consequently, some of the discrepancies previously reported in the literature between the two groups may be in part due to the dissimilarities between the walking conditions in which the measures were taken.

3.4.1 Grand Ensemble Average

Despite the manipulations in speed and BWS and the lack of additional footswitches and kinematic data to determine more precisely the moment of heel- and toe-off during the gait cycle, the GEA's observed for both the RTA and RMG, for all 9 conditions, appear to be consistent with their functional roles and, in general, exhibited classical bi-phasic and mono-phasic patterns respectively. In other words, the burst patterns from the RMG and RTA within the control group appeared to occur during the appropriate phase of the gait cycle and presumably suggest the occurrence of critical gait events such as heel- and toe-off. However, some irregularities from the patterns reported in the literature, as well as modulations to the timing and duration of muscle activity were observed. These irregularities and modulations were most likely caused by the artificial walking conditions employed in this study (i.e. extremely slow speeds with BWS).

More specifically, the RTA-GEA patterns in this study can be characterized as having the greatest amounts of activity during the first 20 and last 40-100% of stride cycle with a small period of low-level activity between the two, albeit not as quiescent as observed in previous reports (Detrembleur et al., 1997; Winter & Yack, 1987 & Yang & Winter, 1985). Yet the general EMG burst patterns are consistent with those reported in the literature (Arsenault et al., 1986a; Detrembleur et al., 1997; Inman, Ralston, & Todd 1981; Winter & Yack, 1987 & Yang & Winter, 1985). Both periods of RTA activity are presumably appropriate, with the first burst of activity being associated with the eccentric activity of the RTA while lowering the foot to the ground at heel-strike and the second causing dorsi-flexion of the ankle at toe-off to prevent toe-drag during swing. Also, the RMG-GEA profiles were consistent with those reported in the literature (Arsenault et al., 1986a; Detrembleur et al., 1997; Inman et al., 1981; Winter & Yack, 1987 & Yang & Winter, 1985), with maximum activity presumably occurring during mid- to late-stance (30-50% of the gait cycle). Functionally speaking, the initial portion the RMG activity would have allowed the leg and body to rotate over the ankle as the ankle was dorsiflexing, followed by a rapid increase in activity to a maximum, presumably when the heel was raised off the ground, just prior to toe-off.

Despite the overall patterns of the RMG and RTA being consistent with their previously reported functional roles and/or timing to the gait cycle, a more precise examination reveals that increases in both speed and BWS shifted the peaks of activity for both muscles to the left, with the effects of speed being more prominent at 20 than 60% BWS. Upon examination of the temporal gait variables reported in the literature, a greater proportion of the gait cycle is spent in the swing phase at faster speeds (Murray et al., 1984 & Nilsson, Thorstensson, Halbertsma, 1985) and with increasing BWS (Finch et al., 1991). Therefore, the leftward phase-shift observed in the GEA peaks of both muscles are likely due to a greater proportion of time spent in swing, and subsequently less time in stance, caused by the increases in both speed and BWS. Consequently, heel- and toe-off may have occurred earlier in the gait cycle under these conditions, which would be consistent with the results by Nilsson et al., (1985) and Detrembleur et al., (1997).

The total duration of muscle activity also appeared to be greater in the slower conditions and when more BWS was provided. This observation suggests that the muscles were chronically active with low levels of activity throughout the stride cycle with less distinct periods of rest under these conditions. The increased stride durations observed during slower speeds and higher levels of BWS in our study, both of which have been previously reported (Murray et al., 1984 & Nilsson et al., 1985 & Finch et al., 1991), may help explain this observation. Similarly, the shorter and longer durations of EMG activity observed for RMG and RTA activity respectively may be accounted for by the smaller proportion of time spent in stance and greater proportion of time spent in swing respectively at higher speeds and greater levels of BWS.

Further, the EMG profiles observed during the faster speeds and lower levels of BWS (e.g. 1.5 km/hr with 20% BWS) were well defined, narrower, and had more distinct periods of activity and inactivity for both muscles than conditions with more BWS at slower speeds (e.g. 0.5 km/hr with 60% BWS). The broader, less-distinct profiles observed while walking at slower speeds with more BWS may have been the result of smaller ground reaction forces exerted on the ground by the muscles to accelerate the body upwards and forwards with increased levels of BWS (Griffin, Tolani & Kram, 1999) and/or smaller moments of force required to accelerate and decelerate the limbs at slower speeds (Yang and Winter, 1985). Consequently, it is very reasonable to think that both speed and BWS have an effect on the mechanical demands of the task and will affect the forces required by the RMG and RTA while walking. Therefore, higher levels of BWS and slower speeds, may have required smaller forces applied over longer durations (i.e. stride duration increased) compared to higher forces applied over shorter durations when walking at faster speeds with less BWS. Therefore, the force-time relationship (i.e. impulse) under these conditions may resemble very closely the EMG curves recorded, with steeper slopes and sharper, better defined peaks occurring in the faster conditions with less BWS and broader, poorly defined curves with prolonged activity while walking slowly with more BWS.

Thus, the results from the present study support the notions previously reported that speed (Detrembleur et al., 1997; Murray et al., 1984; Shiavi et al., 1981; Shiavi et al., 1987 & Wooten et al., 1990) and BWS (Finch et al., 1990) have an influence on the timing and phasic characteristics of the EMG profiles of able-bodied individuals and are in contrast with those by Yang & Winter (1985), Kameyama et al., (1990) Milner et al., (1971) and Winter (1983), who reported no changes in the phasic EMG profiles with variations in speed.

Finally, it should be emphasized that the GEA's presented in figures 3.3 - 3.5 do not represent a universally true profile for the conditions in which they were recorded, but rather they represent the average profile within which the muscle is activated, because the individual variations have been averaged out (Arsenault, et al., 1986a). Furthermore, given the normalization procedure used in the present study, comparisons between EMG amplitudes between subjects cannot be made because each subject's EMG has the same weight (Winter & Yack, 1987). However, according to Delisa (1998), the timing of muscle activity is one of the most powerful characteristics obtained from dynamic EMG recordings and provides researchers and clinicians with a technique of determining whether or not a muscle is active or inactive and whether or not the activity is appropriately timed to critical gait events. Having said this, the GEA's recorded in this study indicate clearly that even under the artificial walking conditions employed, the RMG and RTA have similar functional roles as reported in the literature during full-weight bearing walking at comfortable speeds with peaks of activity generally occurring during the appropriate phase over the stride. However, increasing BWS and reducing speed does appear to induce modulations in the on-off timing, peaks, as well as the duration of activity.

Therefore, based upon our results, one must use the RTA and RMG GEA's compare the individual EA's recorded from the SCI participants while walking under the identical conditions. Otherwise differences in the phasic nature of the EMG profiles, GEA versus SCI cannot be attributed to SCI gait pathology because both speed and BWS were found to systematically modulate the phasic characteristics of the GEA in this study.

For example, previous comparisons between able-bodied individuals walking at their comfortable walking speed to that of SCI participants walking at much slower comfortable walking speeds (Dietz et al., 1994; Dietz et al., 1995; Dietz et al., 1999; Fung & Barbeau, 1989; Leroux et al.,1999 & Protas et al., 2001) may have led researchers to the false conclusion that the SCI participants had delayed and prolonged muscle activity relative to the controls. However, the results from the present study indicate that as the speed was reduced the peak activity for both the RTA and RMG occurred later in the gait cycle. Thus, the SCI participants may have been exhibiting normal EMG profiles yet were misdiagnosed because the peak activity occurs earlier at faster speeds (also observed in out study) and not necessarily because of any gait pathology in the SCI participants.

In conclusion, the results from the present study support the notion that direct comparisons between groups should only be performed when using identical walking conditions. Comparisons of this nature should allow for a more precise and objective analysis of the activity patterns recorded from the SCI participants than any other study conducted to date. Furthermore, dissimilarities between the GEA and the EMG profiles recorded from SCI participants in our study will be more easily attributed to their gait pathology because the effects of speed and BWS have been accounted for in the GEA.

3.4.2 Comparison of GEA to SCI Participants EA's

Despite the generic nature of the GEA, this profile, accompanied by the standard deviations (i.e. ± 1 SD) provides an index within which a muscle should expresses its peculiarities, thus allowing us to interpret and draw conclusions regarding the abnormalities in the SCI participants' gait patterns. However, the focus of the present discussion will be on the major discrepancies/similarities between the GEA and the individual EA's for each SCI participant

3.4.2.1 SCI 01 Vs GEA

The activity of the RMG and RTA were all poorly correlated to the respective GEA's for all nine conditions for this participant (i.e. none > 0.80), indicating that there were phasic dissimilarities between the EMG profiles of this subject and the GEA's. Upon examination of the EA's for the RMG of this participant while walking at 20% BWS at 0.5 km/hr, it is apparent that the muscular activity was very irregular, with a bi-phasic pattern, with a large proportion of activity occurring between 0-30%, then declining to a minimum by 65% of the cycle before rising back up again. This pattern of activity is almost the reciprocal profile of the GEA, hence the negative correlation. In fact, the RMG profiles for SCI 01 all exhibited a bi-phasic activity pattern across all nine conditions with the first peak of activity occurring within the first 20-40% and the second much later, around 90-100% of the cycle. The second burst of RMG activity, occurring later on during the stride is not easily explained. However, it is possible that inappropriate plantar flexor activity occurred during the swing phase just prior to heel-strike in order to assist with the stabilization of the ankle joint and provide support as the limb was loaded.

Although the first peak of RMG activity always occurred after heel-strike, which is presumably appropriate in order to stabilize the ankle during stance and provide the necessary effort to raise the heel prior to toe-off, it was observed always to occur before that of the GEA. In other words, the activity of the RMG for this participant was always premature. One possible explanation for this may be that this participant's gait pattern was markedly different than the controls'. More specifically, his strides were much longer than the controls, both absolute and normalized, likely due to his taller stature (190.5 Vs 167.8 cm for the controls), while his stride durations were also greater with much smaller cadence values. The dissimilarities in stride length, duration and cadence may help explain why the RMG activity occurred earlier in the stride period relative to the GEA. However, in the absence of more quantitative data (i.e. kinematics, and/or toe switch) it can only be speculated that less proportion of the stride time was spent in stance, similar to that observed in able-bodied individuals while walking at faster speeds, thus causing the first burst of activity from the RMG to occur earlier in the stride cycle than that observed in the GEA.

Interestingly, a similar leftward phase-shift in the first peak of RMG activity as speed increased was observed in this subject's EMG profiles as compared to the GEA.

The phase-shift observed in this participant can be presumably explained using the same logic applied to rationalize this pattern in the GEA. That is, as speed increased, less proportion of the gait cycle would have been spent in stance, thus, heel-off would have occurred earlier in the stride. Consequently, while walking at 1.5 km/hr with 60% BWS the primary peak of activity is very close to that observed in the GEA along with similar periods of inactivity, resulting in the highest correlation coefficient for this participant. Likewise, all the profiles recorded at 60% BWS had more closely timed peaks of activity to that of the GEA, paralleled by more appropriate periods of inactivity.

On the other hand the RTA activity for this participant can be characterized as mono-phasic as opposed to the typically observed bi-phasic pattern in the GEA. Generally, the singular burst of activity was very well defined, with a very steep slope as the activity rose and fell and was always preceded and followed by periods of relative quiescence. The absence of activity during heel-strike and phasic dissimilarities between the second burst of activity between the GEA and this participant can account for the low correlations for this muscle.

Other than the one condition (20% BWS at 0.5 km/hr), which had a negative correlation, the peak of RTA activity in each condition generally coincided well with the second burst of activity in the GEA. Consequently, it can be presumed that this participant did have appropriately timed dorsiflexion of the ankle in order to raise his toes off the treadmill to prevent toe-drag. However, it is also apparent that the RTA activity was not prolonged throughout swing as was that observed in the GEA, nor was it present at heel-strike to guide the foot to the floor to prevent foot-drop. Nonetheless, the RTA activity did appear to be more similar to that of the GEA with 40 and 60% BWS and at

0.5 and 1.0 km/hr, which is consistent with the higher correlation coefficients found for these conditions.

Lastly, despite the irregularities observed for both the RMG and RTA compared to the GEA, reciprocal activity between the RTA and RMG for this subject is readily observable in most conditions. This result suggests that some form of flexor-extensor modulation was occurring between these two muscles, also suggesting that relatively low levels of co-contraction were occurring which is indicative of a relatively efficient interchange between the two muscles.

3.4.2.2 SCI 02 Vs GEA

The correlation coefficients for this subject were the highest of all three SCI participants. Four high correlation coefficients (i.e. > 0.80) were observed for the RMG whereas only one high coefficient was observed for the RTA. Nonetheless, all of the correlations for the RMG and RTA were above 0.50 and 0.52 respectively for this subject. Similar to SCI participant 01, increases in BWS or decreases in speed were paralleled by increases in the coefficients for both muscles. Similarly, the correlation for the RTA was also observed to decrease when the speed was increased from 0.5 to 1.0 km/hr yet the correlation increased once the speed was increased to 1.5 from 1.0 km/hr. The high correlations for this participant indicate that the similarity in phasic activity between the GEA and his profiles are very good across the stride for all nine conditions. However, four of the five highest correlations occurred while walking with 60% BWS, suggesting that this condition allowed for the most "normal" phasic muscular activity patterns in this participant.

The activity of the RMG, in all instances, was observed to be appropriately quiet upon heel strike and generally rose rapidly to an initial peak in all conditions around 20-30% of stride cycle, which in all cases occurred distinctly before the peak activity of the GEA, except for one condition (60% BWS at 0.5 km/hr). After the peak activity, the profile either plateaued before decreasing or would decrease slightly before leveling off and at times would rise up again prior to going down to a minimum. This profile is unlike the steady decline in RMG activity exhibited in the GEA profiles. Thus, the activity of the RMG was unquestionably prolonged across the nine conditions and can be characterized as having an earlier onset with late suppression. Premature and prolonged activity of the RMG has been observed before in ambulatory SCI participants and was attributed to exaggerated responses to muscle stretch and impaired delivery of supraspinal signals descending to the spinal interneuronal network respectively (McKay et al., 1993). Prolonged activity during level walking was also observed by Leroux et al (1999) in their SCI participants while walking on level treadmills, yet was simply attributed to impaired motor recruitment. Another possible explanation for the observed RMG activity was that it may have been necessary in order to compensate for muscle weakness and prevent unwanted dorsiflexion during stance.

Generally, the RTA activity was not observed to begin until the RMG activity ended. Thus, very little co-activation between these two muscles was apparent in this subject and their activity could be described as reciprocal in nature. In fact, only a few of the conditions had any notable areas of co-activity in the two muscles during the stride period. Thus, this subject, like SCI 01, also exhibited reciprocal muscle activity patterns between the ankle antagonists.

The RTA activity exhibited by this subject was generally tri-phasic, unlike the typical double-burst profiles observed in the GEA. The peak of the first burst of activity was always observed to occur within the first 20% of stride yet was always delayed relative to the GEA. Nonetheless, the timing of this activity suggests that the RTA was acting appropriately during this phase of stride and presumably assisted with the lowering of the foot to the ground by controlling the amount of plantar flexion of the foot and preventing foot-drop. However, the activity of the RTA appeared to be delayed rather than anticipatory, as suggested by McKay et al (1993). After the first burst of activity, the RTA exhibited good suppression of activity during what is assumed to be the stance phase. After the period of low-level activity, the profile was observed to rise and fall very sharply at around 70-85% of the cycle, followed by a second sharp rise and fall shortly thereafter (shaped similarly to the letter “M”). The first portion of the “M” was likely timed appropriately in order to dorsiflex the ankle and provide toe-clearance during the early part of the swing phase, whereas the second burst may have been in preparation for the larger burst observed to occur after heel-strike. Either way, the last two, very abrupt and distinct bursts of the RTA, are not characteristic of this muscle and were not observed in the GEA suggesting some form of abnormal muscle recruitment.

The peaks of activity of both the RTA and RMG for this participant were observed to occur earlier in the stride as the speed and level of BWS was increased which was also exhibited in the GEA. As indicated for the control group, the proportion of time spent in swing phase increases as the rate of ambulation increases, thus possibly causing various critical gait events such as heel- and toe-off to occur earlier in the stride, which would be accompanied by a leftward phase-shift in the muscular activity. Therefore, similar modulations to the mechanical demands of the task would have likely occurred for the SCI participants and the earlier peaks of activity observed for this participant suggests that his motor patterns were adapted to suit the physical demands of the task. However, greater phasic dissimilarities occurred at higher speeds as indicated by the lower correlation coefficients.

3.4.2.3 SCI 03 Vs GEA

All of the correlations for SCI participant 03 were negative with the exception of two. Both positive coefficients were very low and occurred for both muscles under the same condition (i.e. 40% BWS at 1.5 km/hr). The negative correlations are due to the fact that both muscles activation patterns were almost 180 degrees out of phase with the GEA. In other words, the phasic characteristics of this participant's EMG profiles were unlike those produced by the control group. Nevertheless, the EMG profiles did exhibit a rhythmical and reciprocal pattern of activity in all six of the conditions he completed, despite being out of phase with the GEA. Thus, the poor correlations indicate that this participant's profiles are dissimilar to the expected patterns of muscle activation

The activity of the RMG was atypically bi-phasic with an unusual burst of activity occurring initially at heel-strike, which has been attributed to an exaggerated response to stretch (McKay et al., 1993). The activity at heel-strike was generally followed by a period of low-level activity with a second burst of activity starting as early as 45% of the cycle to as late as 85%. The secondary burst of activity gradually increased to a plateau and continued at this level before sloping downward, and in only one condition did the activity rise up again to a peak at 100% of the cycle (40% BWS at 0.5 km/hr). The abnormal profile of the RMG suggests that the ankle was either plantar-flexing (i.e. concentric) or dorsiflexing (i.e. eccentric) during either of these phases of the gait cycle, yet without kinematic data it cannot be determined conclusively whether or not this activity was concentric or eccentric.

Similarly, the activity of the RTA was atypically mono-phasic for all six conditions with the only burst of activity occurring mid-stride (i.e. 20-70%), which is inconsistent with the GEA's recorded in this study. This abnormal burst of activity from the RTA suggests that the ankle was being dorsi-flexed in order to raise the toes up to prevent them from dragging in preparation for the next stride. Additionally, with the absence of a second burst of activity it may be presumed that upon heel-strike that the foot was not being lowered downward to the belt, resulting in foot-drop.

Anecdotally, this participant was observed to have the greatest difficulty of the three SCI participants walking on the treadmill and was not able to complete any of the conditions at 20% BWS. The atypical activity observed in both the RMG and RTA as well as the poor correlation coefficients suggests that this participant did not use a "normal" gait pattern. Nonetheless, this subject did ambulate independently on the treadmill and the activity from both muscles was reciprocal and rhythmical in nature, suggesting that this individual's motor patterns were at least consistent stride-to-stride, and that the motor patterns had been adapted in order to produce gait, albeit abnormal.

3.5 Conclusion

The present study described and compared the EMG characteristics across the gait cycle in able-bodied individuals to that of three SCI participants who had completed ten months of BWSTT using the same nine conditions with varying combinations of BWS and speed. The activity profiles observed in the control group under slower speeds and greater levels of BWS were characterized by prolonged, low-levels of activity with less distinct peaks. Therefore, within the limitations of the present study, it can be concluded that both speed and BWS affect the timing and durations of the muscular activity of the control group's RMG and RTA activity while walking under these unusual conditions. From this observation, the results from this study support the notion that identical conditions should be employed when comparing the phasic EMG profiles between individuals with SCI to that of able-bodied controls in order to properly determine if the characteristics of the EMG profiles are abnormal. For instance, the patterns exhibited by the control group in this study displayed distinct phasic and timing variations in their RMG and RTA profiles from those reported in the literature. Consequently, the patterns produced by our control subjects could be considered abnormal relative to those found in the literature because the conditions employed in our study were vastly different. Furthermore, our GEA profiles also exhibited phasic and timing modulations from one condition to the next, providing additional evidence supporting the notion that speed and BWS affect the temporal characteristics of muscular activity during gait. Thus, our GEA profiles provide a better index to which the SCI participants EMG patterns can be contrasted.

It can also be readily concluded from our results that the phasic muscular activity patterns exhibited by our three SCI participants were plagued with numerous deviations from the patterns recorded from the control group across all nine conditions, including premature bursts, omissions, and prolonged low-levels of activity. However, for two subjects, the correlation coefficients increased with greater levels of BWS, suggesting that these conditions may provide a better opportunity for more "normal" gait patterns among certain SCI subjects. Nonetheless, additional studies measuring gait kinematics along with EMG recording would be necessary before drawing this conclusion.

All three subjects displayed rhythmical and reciprocal activity of the ankle musculature, with some signs of appropriate activity, although they were generally anomalous relative to the GEA. The variations in motor patterns suggest that compensatory movements must have been occurring in order for the participants to ambulate on the treadmill, and would have been detectable had kinematic data been collected. Additionally, a leftward phase-shift in the peak activity of two of the SCI participants occurred when speed and BWS were increased, a result which was similar to that observed in the GEA. These modulations suggest that the their motor outputs were adaptable and more importantly that they appeared to adapt appropriately as the demands of the task changed. This finding is in contrast to those reported by Leroux et al (1999), where their SCI participants' musculature showed very little adaptation to changes in walking grades.

In conclusion, certain characteristics of the GEA profiles are presumably associated with critical gait events (e.g. activity of the RTA and RMG during toe-off and heel-off respectively). Consequently, omissions, phase-shifts, etc. during various periods of the gait cycle observed in the EMG profiles of the SCI participants suggest that their gait and motor patterns are abnormal despite ten months of training on the treadmill. Nevertheless, despite obvious gait abnormalities in their gait patterns, it was apparent that all three of these individuals adapted their motor patterns in order to ambulate independently. The results from this study also suggest that these individuals have developed strategies to compensate for their various motor deficits, and that they are capable of modulating these patterns, allowing them to ambulate freely on the treadmill over various speeds and levels of BWS.

3.6 References

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4.0 SUMMARY OF FINDINGS

Other studies have compared the metabolic demands and EMG profiles of able-bodied individuals walking at their customary walking speeds to that of persons with incomplete SCI while ambulating with assistive devices, at slower speeds and/or with BWS. Our study attempted to isolate and account for the effects caused by slow speeds and BWS by making a direct and systematic comparison between these two populations under 9 different test conditions (3 levels of BWS). We hoped to end with a true representation of the metabolic demands and EMG profiles associated with treadmill walking for able-bodied individuals that could be used to contrast with the responses from individuals with incomplete SCI who had trained for ten months on the BWS treadmill.

The results from this study clearly illustrate that while walking on the treadmill metabolic rates for the SCI group were higher and less efficient. Furthermore, the EMG profiles from the SCI participants exhibited numerous phasic abnormalities relative to the control group for all nine conditions tested. The significant difference in oxygen consumption and EMG profiles between the two groups, across all three speeds and levels of BWS revealed that the SCI group was working much harder to ambulate and had adapted altered motor strategies.

The higher rates of oxygen consumption in the SCI group can be directly attributed to their obvious difficulties with production of a smooth and fluid gait pattern.

Presumably there were exaggerations in the motion of the body's center of mass in all directions, which interfered with the smooth and natural conversion of potential to kinetic energy and vice versa. The lack of conservation of energy observed in the metabolic data is further supported through examination of the EMG data. In general, the EMG data from the SCI group was plagued with irregular, prolonged, inappropriate and anomalous levels of muscular activity throughout the gait cycle relative to the control group. All of these abnormal muscular activities observed in the SCI participants suggest a lack of coordination, exaggerated movements and/or superfluous levels of muscular activity that would likely be contributing to the additional oxygen consumed by the SCI participants. In addition, given that the EMG collected in this study was from small, distal muscles, it may also be assumed that the larger, more proximal muscles would have exhibited pathological activity patterns as well, thus also contributing to the increased metabolic demands of SCI participants.

The highest rate of oxygen uptake for the SCI group occurred while walking at 1.5 km/hr with 20% BWS (12.75 ml/kg/min) and was calculated to be equivalent to that of an able-bodied person walking at 4.7 km/hr. This result alone allows us to conclude that the SCI participants are able to achieve an effective aerobic exercise stimulus while ambulating independently on the treadmill. This significant aerobic exercise stimulus carries with it meaningful health benefits, given previous reports that small increases in VO2peak above 20 ml/kg/min can substantially decrease the risk for cardiovascular disease. Presumably then, with the appropriate amount of training, a BWSTT program could allow for increases in the overall level of cardiovascular fitness (i.e. VO2peak measured during a graded exercise test) and subsequently reduce the risks of developing common ailments associated with the otherwise sedentary lifestyle of the SCI population.

Therefore, within the limitations of the present study, it can be concluded that both speed and BWS affect the temporal characteristics of ambulatory muscular activity and metabolic demands in both individuals with SCI and able-bodied persons. We found that BWSTT provides the opportunity for persons with SCI to experience upright, large muscle mass exercise, allowing for a greater exercise stimulus than that provided by other modalities. However, even with ten months of BWSTT altered motor strategies will continue to persist in individuals with incomplete SCI.

The results from this study suggest several things. First, comparisons between able-bodied individuals and those with SCI should be done, if possible, under similar conditions in order to control for the variance caused by dissimilar conditions. Second, the SCI population is much less efficient in ambulation than the able-bodied, a finding that is further supported by the numerous abnormalities observed in the EMG profiles recorded from the SCI participants. It is obvious that the neurological deficits suffered by these individuals as a result of their injury have resulted in impaired muscular recruitment strategies, which ultimately have significantly increased their metabolic demands of locomotion. Nonetheless, it is important to note that all of our SCI participants achieved the ability to ambulate independently on the treadmill after ten months of BWSTT, despite having different motor patterns from able-bodied persons. Furthermore, it is evident that BWSTT offers an effective cardiovascular workout to this population.

Appendix A: Modified Borg Scale

0	**NOTHING AT ALL**
0.5	**VERY, VERY, SLIGHT (just noticeable)**
1	**VERY SLIGHT**
2	**SLIGHT**
3	**MODERATE**
4	**SOMEWHAT SEVERE**
5	**SEVERE**
6	
7	**VERY SEVERE**
8	
9	**VERY, VERY SEVERE**
10	**MAXIMAL (almost maximal)**

Appendix B: Raw Data, ANOVA Tables & Post Hoc Analyses

Raw VO2 Data (ml/kg/min)

group	20X05	20X10	20X15	40X05	40X10	40X15	60X05	60X10	60X15
Control 01	6.23	6.74	6.80	5.41	6.53	6.69	4.77	5.19	5.49
Control 02	6.53	6.08	7.10	7.64	7.51	5.69	6.82	5.79	6.69
Control 03	6.38	7.12	8.55	5.92	5.89	6.86	5.30	5.87	5.95
Control 04	5.45	5.76	6.48	6.75	6.24	6.98	5.00	4.99	6.21
Control 05	5.47	6.21	6.50	5.51	5.83	5.64	5.24	5.63	5.74
Control 06	7.87	5.32	7.56	8.28	8.41	9.62	6.28	7.67	7.81
Control 07	4.79	5.21	6.35	5.40	7.44	6.98	4.96	5.94	6.51
Control 08	4.40	5.58	5.63	5.03	4.96	5.42	4.60	4.98	5.06
Control 09	4.60	5.89	5.37	4.15	4.76	5.52	3.92	4.16	4.62
Control 10	6.02	5.99	6.57	5.78	6.19	7.04	4.87	6.03	7.65
SCI 01	10.06	11.50	11.45	9.46	11.39	12.24	7.90	10.21	10.20
SCI 02	11.64	12.79	12.32	8.56	10.46	10.64	8.52	13.46	11.09
SCI 03	11.07	12.88	15.60	8.78	11.44	13.47	7.84	10.78	11.07
SCI 04	7.72	9.71	11.64	6.91	8.59	9.74	6.84	7.79	9.31

Anova Table for VO2 Data (ml/kg/min)

Summary of all Effects; design:
1-GROUP, 2-%BWS, 3-SPEED

	df Effect	MS Effect	df Error	MS Error	F	p-level
1 (Main Effect Group)	1	494.1	12.0	7.7	64.6	0.0
2 (Main effect % BWS)	2	12.8	24.0	1.0	12.5	0.0
3 (Main Effect Speed)	2	30.3	24.0	0.6	47.3	0.0
12 (Interaction Effect Group x BWS)	2	6.4	24.0	1.0	6.3	0.0
13 (Interaction Effect (Group x Speed)	2	10.2	24.0	0.6	15.9	0.0
23 (None)	4	0.5	48.0	0.3	1.4	0.2
123 (No 3 way interaction)	4	0.5	48.0	0.3	1.6	0.2

Tukey HSD test; VO2 Data, ml/kg/min
Probabilities for Post Hoc Tests
INTERACTION: *Group x % BWS*

			Control			SCI		
			20% BWS	**40% BWS**	**60% BWS**	**20% BWS**	**40% BWS**	**60% BWS**
			6.151588	6.336083	5.657406	11.53204	10.17934	9.584629
Control	**20% BWS**	6.151588	-------------					
Control	**40% BWS**	6.336083	0.98	-------------				
Control	**60% BWS**	5.657406	0.43	0.14	-------------			
SCI	**20% BWS**	11.53204	0.00	0.00	0.00	-------------		
SCI	**40% BWS**	10.17934	0.00	0.00	0.00	0.03	-------------	
SCI	**60% BWS**	9.584629	0.00	0.00	0.00	0.00	0.702	-------------

Tukey HSD test; VO2 Data, ml/kg/min
Probabilities for Post Hoc Tests
INTERACTION: Group x Speed

			Control			SCI		
			0.5 km/hr	**1.0 km/hr**	**1.5 km/hr**	**0.5 km/hr**	**1.0 km/hr**	**1.5 km/hr**
			5.65	6.00	6.50	8.77	10.92	11.60
Control	0.5 km/hr	5.65	-------------					
Control	1.0 km/hr	6.00	0.549058	-------------				
Control	1.5 km/hr	6.50	0.004426	0.177799	-------------			
SCI	0.5 km/hr	8.77	0.000138	0.000138	0.000138	-------------		
SCI	1.0 km/hr	10.92	0.000138	0.000138	0.000138	0.000145	-------------	
SCI	1.5 km/hr	11.60	0.000138	0.000138	0.000138	0.000138	0.321	-------------

Raw Metabolic Cost data (ml/kg/m)

group	20X05	20X10	20X15	40X05	40X10	40X15	60X05	60X10	60X15
Control 01	0.75	0.40	0.27	0.65	0.39	0.27	0.57	0.31	0.22
Control 02	0.78	0.36	0.28	0.92	0.45	0.23	0.82	0.35	0.27
Control 03	0.77	0.43	0.34	0.71	0.35	0.27	0.64	0.35	0.24
Control 04	0.65	0.35	0.26	0.81	0.37	0.28	0.60	0.30	0.25
Control 05	0.66	0.37	0.26	0.66	0.35	0.23	0.63	0.34	0.23
Control 06	0.94	0.32	0.30	0.99	0.50	0.38	0.75	0.46	0.31
Control 07	0.57	0.31	0.25	0.65	0.45	0.28	0.59	0.36	0.26
Control 08	0.53	0.33	0.23	0.60	0.30	0.22	0.55	0.30	0.20
Control 09	0.55	0.35	0.21	0.50	0.29	0.22	0.47	0.25	0.18
Control 10	0.72	0.36	0.26	0.69	0.37	0.28	0.58	0.36	0.31
SCI 01	1.21	0.69	0.46	1.13	0.68	0.51	0.95	0.61	0.41
SCI 02	1.40	0.77	0.49	1.03	0.63	0.43	1.02	0.81	0.44
SCI 03	1.33	0.77	0.62	1.05	0.69	0.54	0.94	0.65	0.44
SCI 04	0.93	0.58	0.47	0.83	0.52	0.39	0.82	0.47	0.37

Anova Table for Metabolic Cost Data (ml/kg/m)

Summary of all Effects; design: 1-GROUP, 2-%BWS, 3-SPEED	df Effect	MS Effect	df Error	MS Error	F	p-level
1 (Main Effect Group)	1	2.18685	12.00000	0.04255	51.39362	0.00001
2 (Main effect % BWS)	2	0.07447	24.00000	0.00442	16.86035	0.00003
3 (Main Effect Speed)	2	2.29840	24.00000	0.00818	280.8551	0.00000
12 (Interaction Effect Group x BWS)	2	0.03920	24.00000	0.00442	8.87448	0.00130
13 (Interaction Effect (Group x Speed)	2	0.06299	24.00000	0.00818	7.69730	0.00261
23 (Interaction Effect %BWS x Speed)	4	0.01515	48.00000	0.00185	8.18561	0.00004
123 (3 way Interaction Group x %BWS x Speed)	4	0.00875	48.00000	0.00185	4.73128	0.00267

Tukey HSD test; Metabolic Cost Data (ml/kg/m) Probabilities for Post Hoc Tests INTERACTION: Group x % BWS x Speed			Control								
			20% BWS			40% BWS			60% BWS		
			0.5 km/hr	1.0 km/hr	1.5 km/hr	0.5 km/hr	1.0 km/hr	1.5 km/hr	0.5 km/hr	1.0 km/hr	1.5 km/hr
			{1}	{2}	{3}	{4}	{5}	{6}	{7}	{8}	{9}
			0.69	0.36	0.27	0.72	0.38	0.27	0.62	0.34	0.25
control	20X05	0.69	--------------								
control	20X10	0.36	0.000	--------------							
control	20X15	0.27	0.000	0.002	--------------						
control	40X05	0.72	0.996	0.000	0.000	--------------					
control	40X10	0.38	0.000	0.999	0.000	0.000	--------------				
control	40X15	0.27	0.000	0.002	1.000	0.000	0.000	--------------			
control	60X05	0.62	0.042	0.000	0.000	0.001	0.000	0.000	--------------		
control	60X10	0.34	0.000	0.999	0.058	0.000	0.649	0.045	0.000	--------------	
control	60X15	0.25	0.000	0.000	1.000	0.000	0.000	1.000	0.000	0.003	--------------
SCI	20X05	1.21	0.000	0.000	0.000	0.000	0.000	0.000	0.000	0.000	0.000
SCI	20X10	0.70	1.000	0.000	0.000	1.000	0.000	0.000	0.147	0.000	0.000
SCI	20X15	0.51	0.000	0.000	0.000	0.000	0.001	0.000	0.008	0.000	0.000
SCI	40X05	1.01	0.000	0.000	0.000	0.000	0.000	0.000	0.000	0.000	0.000
SCI	40X10	0.63	0.508	0.000	0.000	0.070	0.000	0.000	1.000	0.000	0.000
SCI	40X15	0.47	0.000	0.012	0.000	0.000	0.135	0.000	0.000	0.001	0.000
SCI	60X05	0.93	0.000	0.000	0.000	0.000	0.000	0.000	0.000	0.000	0.000
SCI	60X10	0.63	0.658	0.000	0.000	0.118	0.000	0.000	1.000	0.000	0.000
SCI	60X15	0.42	0.000	0.708	0.000	0.000	0.996	0.000	0.000	0.192	0.000

Tukey HSD test; Metabolic Cost Data (ml/kg/m): Continued

		SCI								
		20% BWS			40% BWS			60% BWS		
		0.5 km/hr	1.0 km/hr	1.5 km/hr	0.5 km/hr	1.0 km/hr	1.5 km/hr	0.5 km/hr	1.0 km/hr	1.5 km/hr
		{10}	{11}	{12}	{13}	{14}	{15}	{16}	{17}	{18}
		1.21	0.70	0.51	1.01	0.63	0.47	0.93	0.63	0.42
control	20X05									
control	20X10									
control	20X15									
control	40X05									
control	40X10									
control	40X15									
control	60X05									
control	60X10									
control	60X15									
SCI	20X05	--------------								
SCI	20X10	0.000	--------------							
SCI	20X15	0.000	0.000	--------------						
SCI	40X05	0.000	0.000	0.000	--------------					
SCI	40X10	0.000	0.561	0.030	0.000	--------------				
SCI	40X15	0.000	0.000	0.990	0.000	0.000	--------------			
SCI	60X05	0.000	0.000	0.000	0.488	0.000	0.000	--------------		
SCI	60X10	0.000	0.686	0.018	0.000	1.000	0.000	0.000	--------------	
SCI	60X15	0.000	0.000	0.209	0.000	0.000	0.972	0.000	0.000	--------------

APPENDIX C: Sample Letter of Information and Consent

METABOLIC DEMANDS AND EMG PROFILES OF BWS TREADMILL WALKING IN PERSONS WITH SCI

The Body Weight Support Treadmill (BWST) is ideal for persons with poor lower body movement and impaired cardiovascular function because it has a specially designed harness, which partially supports a person while walking on a treadmill. This means that walking can be performed with minimal stress to the body. This technique provides an opportunity for us to safely evaluate your response to various intensities of exercise by varying the speed of the belt and level of body weight support.

The purpose of this study is to examine your physical responses to various intensities of walking on a body-weight supported treadmill (BWST). You will be asked to perform three 12-minute bouts of walking on the treadmill, each at a different level of body weight support and followed by five minutes of rest. The speed of the treadmill will be changed twice during each bout of exercise. During these exercise bouts photographs and/or video may be taken for the purposes of scientific presentations, scientific publications and/or instructional purposes. Below is a list of the procedures we will be using including the risks associated with each.

Metabolic Cart: The metabolic cart is used to analyse your expired air for several factors, for example, the amount of oxygen you are consuming. It involves breathing through a mouthpiece positioned in your mouth, a clip over your nose and a lightweight headpiece to secure everything in place. The mask is designed with valves that allow you to breathe room air while your expired air flows through tubing which gets analysed by a computer. There are no risks associated with this procedure.

Rating of Perceived Exertion (RPE) Scale: This scale is used to determine how hard you feel your legs are working and how hard you feel you are breathing. It is a 10-point scale on a piece of paper and when asked, you simply point to the number which best represents how you feel. There are no risks associated with this procedure.

Electromyography (EMG): Lower body muscular activity is measured through adhesive electrodes placed on your lower legs. In order to achieve a clear signal from the electrodes, the skin must be cleaned with rubbing alcohol and in some cases shaved with a new disposable razor. As a result of the skin preparation and electrode placement, you may experience some minor skin irritation, which should clear up within 2-3 days. This irritation is similar to that caused by a household band-aid.

*Initials:*_____

Heart Rate: A Polar™ Heart Rate Monitor will be used to monitor your heart rate, before, during, and after exercise. It involves positioning a plastic, reusable heart rate sensor on your chest, just below the breast line, the sensor is held in place using an adjustable elastic strap around your torso. The sensor is cleaned with alcohol after every use and transmits your hear rate signal to a wrist watch. There are no risks associated with this procedure.

All of the above procedures will be performed at designated intervals throughout the study.

Potential Risks and/or Discomforts: There are minor physical risks associated with participation in this study. These include muscle strains, chafing and/or bruising from the harness, skin irritation from the adhesive electrodes and possible increases or decreases in blood pressure for those with spinal cord injuries.

Additional foam padding will be positioned under the leg straps and areas where you identify discomfort to help reduce the amount of chafing and/or bruising caused by the harness. All electrode sites will be cleaned with rubbing alcohol and tissue paper to remove any excess adhesive. The student investigator has been trained to recognize the signs and symptoms of sudden increases and/or decreases in blood pressure and knows how to respond appropriately.

Potential Benefits: Participants will not benefit directly from this acute exercise session; however, this investigation is an essential step to determining the metabolic and muscular efficiency of spinal cord injured patients receiving extensive body weight support treadmill training as compared to able-bodied persons. The data collected from this study is also essential for prescribing exercise and monitoring the progress of current and future patient rehabilitation programs using body weight support treadmills.

The data derived from this study will also provide valuable insight into the main and interaction effects that speed of locomotion and percent of body weight support have on the physiological aspects of walking. This study will also provide normative data from which the scientific community may draw comparisons and conclusions when evaluating and prescribing exercise programs using body weight support treadmills.

Confidentiality: Any information obtained in connection with this study that can be identified with you will remain confidential and will be disclosed only with your permission. All data will be stored in a locked filing cabinet in the office of A. Hicks. Findings will be published in scientific journals and where necessary, subjects will be referred to by number only.

Participation and Withdrawal: You can choose whether to be in this study or not. If you volunteer to be in this study, you may withdrawal at any time without consequence of any kind. You may exercise the option of removing your data from the study. The investigator may withdraw you from this study if circumstances arise which warrant doing so.

***Initials:*______**

Rights of Research Participants: You may withdraw your consent at any time and discontinue participation without penalty. You are not waiving any legal claims, rights or remedies because of your participation in this research study. This study has been reviewed and received ethics clearance through the McMaster Research Ethics Board (MREB). If you have any questions regarding your rights as a research participant, contact:

MREB Secretariat
McMaster University
1280 Main St. W. CNH-111
Hamilton, ON, L8S 4L9

Tel: 905-525-9140 ext. 24765
E-mail: grntoff@mcmaster.ca
Fax: 905-540-8019

This work is being conducted/supervised by the following investigators. You are free to contact them with questions or concerns at any time during the study.

Dr. Audrey Hicks, Professor, Kinesiology. Ivor Wynne Centre A203. Tel. 525-9140 ext. 24643

Student Investigator: Nathaniel Dufresne, nldufres@hotmail.com, Tel. 905-524-3704

*Initials:*_____

METABOLIC DEMANDS AND EMG PROFILES OF BWS TREADMILL WALKING IN PERSONS WITH SCI

Consent of Participant

I have read, initialled and understood the information presented in the participant information letter and understand the procedures and risks involved in the study. I have received satisfactory answers to my questions related to this study. This project has been reviewed and received ethics clearance through the McMaster University Research Ethics Board (MREB). I understand that if I have any further questions or concerns resulting from my participation in this study, I may contact the Office of Research Ethics in person at CNH 111; by phone 905-525-9140 ext. 23142 or via e-mail: srebsec@mcmaster.ca. I am aware that I may withdraw from the study at any time. With full knowledge of all the foregoing, I agree, of my own free will, to participate as a subject in this study.

Print Name ____________________ Signature of Participant ____________________

Dated at McMaster, Ontario ____________________ Witnessed ____________________

I agree to allow video and/or photographs to be used in teaching or scientific presentations or publications of this work.

Print Name ____________________ Signature of Participant ____________________

Dated at McMaster, Ontario ____________________ Witnessed ____________________

***Initials:*_____**