METABOLIC DEMAND OF BODY WEIGHT SUPPORTED LOCOMOTION

COMPARISON OF OXYGEN DEMANDS AND MUSCLE ACTIVITY PATTERNS DURING DIFFERENT FORMS OF BODY WEIGHT SUPPORTED LOCOMOTION IN INDIVIDUALS WITH INCOMPLETE SCI

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ABSTRACT

Body weight supported devices available to enhance locomotor recovery following an incomplete spinal cord injury (SCI) include treadmills with (Lokomat^{TM}) and without (Manual Treadmill) robotic assistance, and the overground ZeroG^{TM} system. Cardiovascular and muscular demands of these devices were compared during steady-state locomotion at the *same* level of body weight support (BWS) in 7 individuals with incomplete SCI (42.6±4.29 years) and matched able-bodied controls (CON). Questionnaires evaluated consumer preference based on walking experience. Oxygen uptake (VO_2), heart rate (HR) and ratings of perceived exertion (RPE) were expressed as percentage of peak values obtained using arm ergometry. Additionally, VO_2 was expressed relative to resting metabolic equivalents (METS). Filtered electromyography (EMG) signals from tibialis anterior (TA), rectus femoris (RF), biceps femoris (BF) and medial gastrocnemius (MG) were normalized to ZeroG^{TM} stepping. Lokomat^{TM} sessions were the least demanding in terms of oxygen uptake compared to the Manual Treadmill and ZeroG^{TM}, and considered the least appropriate device for the SCI group's current level of function. For SCI, the a) ZeroG^{TM} required 3.0 METS, 54.7% of VO_2peak, 84.7% of peak HR, b) Manual Treadmill required 2.8 METS, 52.9% of VO_2peak, 80.8% of peak HR and c) Lokomat^{TM} required 1.7 METS, 30.1% of VO_2peak, 67.3% of peak HR. Central RPEs were 3.8, 3.7, 0.5 and peripheral RPEs were 5.1, 4.1, 0.7 for the ZeroG^{TM}, Manual Treadmill and Lokomat^{TM} respectively. For CON, walking required minimal effort (at most 31.5%

of VO_2 peak), with ZeroG^{TM} sessions requiring greater muscle activation. For SCI, muscle activation was higher in treadmill conditions compared to the ZeroG^{TM} due to increases in TA and BF activity. The Manual Treadmill and ZeroG^{TM} should be considered progressions following Lokomat^{TM} training where hip extension can be encouraged using the treadmill and additional components of gait (e.g. balance, torso stability) can be focused on using the ZeroG^{TM}.

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“Your talent is God’s gift to you.

What you do with it is your gift back to God.”

- Leo Buscaglia -

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LISTS OF ABBREVIATIONS

ADL	Activities of Daily Living
ANOVA	Analysis of Variance
AIS	American Spinal Injury Association Impairment Scale
BF	Biceps Femoris
BMI	Body Mass Index
BPM	Beats Per Minute
BWS	Body Weight Support
CON	Able-Bodied Controls
EMG	Electromyography
HR	Heart Rate
MET	Metabolic Equivalent
MG	Medial Gastrocnemius
RF	Rectus Femoris
RMS	Root Mean Square
ROM	Range of Motion
RPE	Rating of Perceived Exertion
RPEC	Rating of Perceived Exertion (Central)
RPEP	Rating of Perceived Exertion (Peripheral)
RPM	Revolutions Per Minute
SCI	Spinal Cord Injury
TA	Tibialis Anterior
VO_2	Metabolic Rate of Oxygen Consumption

CHAPTER 1: REVIEW OF THE LITERATURE

1.1 SPINAL CORD INJURY

1.1.1 Description and Epidemiology

The spinal cord consists of nerve bundles that connect the brain with the rest of the body. ***Afferent signals*** (e.g. sensory) send information *towards* the central nervous system that can be processed by the brain (except for reflexes), which can then send responses (***efferent signals***) to muscles and/or glands. This intricate information superhighway is protected by the bony vertebral column. Any damage experienced by the spinal cord, however, can interrupt these signaling pathways resulting in alterations to normal function (motor, sensory and/or autonomic).

As of June 2013, the incidence and prevalence of spinal cord injury (SCI) in Canada was 4,300 and 86,000 respectively (Rick Hansen Institute, 2013). ***Traumatic SCIs*** are caused by accidents, most commonly resulting from falls (43.2%) or motor vehicles (42.8%) (Spinal Cord Injury Ontario, 2013). In Ontario, males represent 68.4% of tramautic SCI, with the highest incidence of injury between the ages of 20-29 and over 70 years old. ***Non-traumatic SCI*** can result from illness or congenital causes (e.g. tumours, spinal disk degeneration) and usually occur later in life (> 60 years old) (Rick Hansen Institute, 2013). Despite healthcare advancements there is still no cure for someone who has suffered an injury and life expectancy is still less than the general population. Canadians who experience a traumatic SCI are expected to live 15-30 years less compared to the general population. A recent study conducted by Middleton and colleagues (2012) in Australia estimated individuals who are injured between the ages of 25 to 65 years might attain 64-97% of the general population's life expectancy. Years of survival post SCI are negatively related to the extent of neurological impairment. Injury severity also has an impact on total lifetime costs for an individual, which can range between $1.5-$3.0 million for traumatic SCI (Rick Hansen Institute, 2013). Cardiovascular disease is currently the leading cause of morbidity and mortality, with earlier onset and increased prevalence experienced in individuals with SCI compared to the general population (Hicks and Martin-Ginis, 2008).

1.1.2 Classification of SCI

Medical imaging techniques (e.g. magnetic resonance imaging) can help determine the exact location and extent of damage following an injury. Classification of the injury can be made based on the spinal level involved (tetraplegia vs. paraplegia) and an assessment of neurological function (American Spinal Injury Association Impairment Scale). Generally, the higher the injury, the more body parts that are affected and the lower the individual's physical capacity compared to lower injury levels or non-injured controls.

Tetraplegia vs. Paraplegia

Individuals with ***tetraplegia*** have sustained an injury in the cervical spine affecting the arms, trunk, legs and pelvic area. Additional complications include difficulty regulating heart rate (HR), blood pressure and body temperature (e.g. sweating). An individual with ***paraplegia*** has an injury in the thoracic and/or lumbar region affecting pelvic organs (bowel and bladder), legs, and possible trunk impairments. While tetraplegia and paraplegia occur approximately equally, paraplegia more commonly results in a ***complete*** injury than tetraplegia (Phillips et al, 1998).

Complete vs. Incomplete Injuries and the American Spinal Injury Association Impairment (AIS) Scale

Neurological function (motor and sensory) is characterized using the ***American Spinal Injury Association Impairment Scale (AIS)*** (Appendix A) (Rick Hansen Institute, 2013). When using this assessment tool the level of an injury is determined as the most caudal segment with *normal* sensory and motor function. Sensory function (122 point score) is measured on a 3-point rating ability to feel a pin prick/light touch along the dermatomes (skin region supplied primarily by a single spinal nerve) from C2-S5. Motor function (100 point score) is measured on a 6-point rating of movement ability for 10 pairs of key muscles. AIS A describes a ***complete*** SCI, where individuals have total ***paralysis*** or loss of sensation and mobility in the sacral region. If an individual retains any function in S4-S5, which innervates the anus, the injury is classified as ***incomplete*** (AIS B-D subtypes). Individuals classified as AIS B retain only sensory function, AIS C are able to actively contract less than half of the key muscles against gravity, with AIS D able to move more than half of these muscle groups through full range of motion against resistance. The final category, AIS E, refers to “normal”, where an individual has suffered an SCI but has not experienced functional loss in sensation or movement. For both complete and incomplete SCI, the likelihood of ***spontaneous recovery*** decreases six months after the injury (Giangregorio et al, 2005).

1.1.3 Physiological Changes After SCI

Locomotor Capacity

In the acute phase of an injury, retention of sensory and motor function influences the extent of locomotor recovery. For example, the ability to detect pinprick sensations within the first three days and/or quadriceps femoris muscle strength above grade three by three months following an injury is an indicator of ambulatory potential (Behrman et al, 2006). Locomotor activity is evident following both complete and incomplete SCI, which has provided evidence for the existence of primitive motor patterns generated at the level of the spinal cord, which will be addressed in subsequent sections. Even though individuals with complete SCI experience total loss of supraspinal input, they are still capable of generating locomotor activity due to the adaptation of spinal neuron networks following the injury. Unfortunately there is no evidence that this leads to improved overground locomotor ability which suggests some supraspinal input may be necessary for independent ambulation. Evidence suggests if a minimum of 10-15% of descending spinal tracts remain intact following an injury some locomotor recovery is possible, irrespective of noticeable changes to AIS lower extremity motor scores (Dietz, 2008). Functional magnetic resonance images of brain activation suggest spared connectivity with the ***cerebellum*** is a requirement for overground locomotion, which is only possible for those with incomplete SCI (Winchester et al, 2005). Connection between the body and the brain, particularly the cerebellum, appears to determine whether improvements will translate into independent overground locomotion. Research suggests this functional achievement is limited to those with incomplete SCI, primarily AIS C or D (Fenuta and Hicks, 2011).

Speed of locomotion is often used as an overall measure of locomotor ability. Following an SCI, locomotion requires greater concentration and is usually only achieved at slow speeds. A study involving community ambulators with SCI found mean comfortable locomotion speed of 0.96 m/s, which is slower compared to speeds the majority of the general population prefer (1.0-1.67 m/s) (Waters et al, 1994). Additionally, individuals with SCI have difficulty adapting to speeds faster than their natural cadence. This is due to an inability to increase stride frequency, limited ability to increase stride length, (Pépin et al, 2003) and the requirement of longer double-support time in order to complete gait cycles (van Hedel et al, 2005).

Weakness and/or spasticity can result in impairments throughout the different phases of gait. These commonly include persistent plantarflexion and hip/knee flexion during swing phase, improper foot placement at heel strike and inadequate hip extension during stance phase (Ditunno and Scivoletto, 2009). ***Discoordination*** of the lower limbs often will result in difficulty weight bearing and contributes to locomotor inefficiency. When maneuvering over obstacles or ambulating uphill, individuals with SCI have been known to use different strategies to complete the same task compared to non-injured adults (e.g. reliance on upper extremity, hip hiking or circumduction) (Leroux et al, 1999). The slow speed and high rate of energy expenditure typically associated with ambulation in individuals with SCI is the reason many of these individuals are usually prescribed a wheelchair for their primary means of mobility.

Muscular Function, Activity and Strength

Function

The extreme catabolic state experienced by paralyzed tissue during the acute phase after an injury makes this population more susceptible to secondary health complications. In the first year following traumatic SCI as much as 68% of muscle atrophy occurs, especially in slow muscle fibres, resulting in muscle weakness (Castro et al, 1999). Muscle fibre type distribution shifts from type I and IIa fibres to an increased proportion of type IIx fibres resulting in decreased endurance. Additional difficulties voluntarily activating skeletal muscle below the lesion level contribute to decreases in oxidative metabolism and glucose intolerance. Collectively, these changes make this population more susceptible to developing type 2 diabetes (Hicks and Martin-Ginis, 2008).

Strength

Lower extremity muscle strength is considered to be a primary determinant of locomotor ability for those with SCI, strongly related to ambulation speed and the rate of oxygen consumption (Waters and Mulroy, 1999). Individuals with SCI considered to be community ambulators demonstrate pelvic control with better hip flexion and knee extensor strength compared to those limited in their ability to independently ambulate in the community (Hussey and Stauffer, 1973).

Individuals with incomplete tetraplegia are less capable of utilizing upper extremity assistive devices for ambulation due to upper extremity paralysis making preservation of lower extremity musculature even more important. Wirth and colleagues (2008) have suggested *dynamic* strength loss is the major motor impairment affecting the lower limbs in incomplete SCI. Chronically paralyzed soleus muscles, for example, achieve 20-30% of initial peak torque values after a bout of repetitive activation (Dudley-Javoroski and Shields, 2008). Thus, while the amount of force one can generate with the lower extremity is important, the rate at which this force can be produced (***power***) also must be considered.

Baseline volitional muscle strength has been shown to be predictive for identifying ***responders*** to body weight supported treadmill training, essentially referring to those who eventually gain functional improvements in overground locomotion. A study by Yang and colleagues (2011) found greater preservation of muscle strength after injury in knee extensors, knee flexors, ankle plantar flexors and hip abductors was indicative of a greater potential to respond to locomotor retraining. These same researchers concluded lower extremity manual muscle testing scores for the quadriceps and hamstrings were the strongest predictors of responsiveness. Furthermore, Gorassini and colleagues (2009) found responders to body weight supported treadmill training had on average twice the volitional muscle strength as measured by manual muscle testing scores compared to non-responders.

Activity

Spasticity (e.g. increased muscle tone, exaggerated spinal reflexes) is present in both complete and incomplete injuries (Dietz, 2008). ***Electromyography (EMG)*** signal amplitudes during gait are significantly lower in injured compared to non-injured individuals. Inappropriate tibialis anterior (TA) and gastrocnemius activity during stance phase is a common gait impairment (Dietz et al, 1997). Through gait retraining, decreases in TA and increases in gastrocnemius activation occur during stance phase resulting in the achievement of more similar muscle activity patterns to non-injured adults.

Similar to baseline volitional muscle strength, baseline muscle activity has been shown to be predictive for identifying responders to body weight supported treadmill training. Research suggests that responders show higher leg EMG during gait pre-training compared to non-responders, particularly higher amplitudes of quadriceps and hamstring muscle groups (Yang et al, 2011). These were the same muscle groups whose manual muscle testing scores Yang and colleagues (2011) determined were predictive of responsiveness to gait retraining as previously mentioned. Individuals with complete injuries exhibit less activity of the lower extremity compared to incomplete injuries, and at levels insufficient to translate to improved overground locomotor ability.

Energy Expenditure and Oxygen Cost of Activities (METS)

Energy expenditure for a given *relative* workload is generally lower for SCI compared to non-injured adults (Price, 2010). One ***metabolic equivalent (MET)*** is the ratio of energy expended during a physical activity compared to resting energy expenditure. The commonly accepted value in healthy adults for 1 MET is 3.5 mL/kg/min (Byrne et al, 2005). A more precise value (2.7 mL/kg/min) has been proposed for those with SCI to account for the physiological changes that occur secondary to paralysis (e.g. loss of muscle mass, decreased sympathetic nervous system availability), which reduces resting metabolic rate (Collins et al, 2010).

During sustained submaximal exercise it takes a few minutes for the rate of oxygen consumption to meet the energy demands of the working tissues. Cardiac output, HR, and respiratory rate then plateau and an individual achieves ***"steady state"***. Steady state rates of oxygen consumption relate to the level of physical effort and energy expended to perform a given task (Waters and Mulroy, 1999). It is a measure of the demand placed on the cardiovascular system to deliver oxygen to working muscles during aerobic tasks (e.g. locomotion), and provides a global estimate of muscular activity (Hornby et al, 2012).

Normal self-selected speed of locomotion has been shown to approximate the most economical ambulation speed for an individual, enabling individuals to keep oxygen consumption below the ***anaerobic threshold*** (RER = 0.90; 50% VO_2 max) (Waters and Mulroy, 1999; Cerretelli et al, 1975). As previously mentioned, a study involving community ambulators with SCI found comfortable walking speed averaged 0.96 m/s, which is slower than speeds for non-injured adults who have slow, normal and fast speeds ranging between 0.62-1.65 m/s. Following an SCI, even at comfortable speeds, locomotion becomes less efficient, with studies indicating higher-energy expenditures in this population compared to normal locomotion. For example, community ambulators with SCI had slightly higher mean rates of oxygen uptake and HR at comfortable walking speeds compared to non-injured adults [14.6 mL/kg/min and 108 beats per minute (BPM) respectively]. This is in agreement with earlier work, which found rate of oxygen uptake during locomotion to be 38% greater in SCI compared to non-injured adults (16.4 mL/kg/min vs. 11.9 mL/kg/min) (Waters and Lunsford, 1985) and more recent work which found 24% higher HR in this population during locomotion (Teixeira da Cunha-Filho et al, 2003). Locomotion for people without mobility impairments is not generally taxing and can be sustained for a prolonged period of time. This is because locomotion remains below the anaerobic threshold, requiring 32-48% of VO_2 max for non-injured individuals between 20-75 years of age (Waters and Mulroy, 1999). Additionally, there is no evidence to suggest significant differences in energy expenditure between treadmill and overground locomotion at controlled velocities.

During locomotion in individuals without mobility impairments the rate of oxygen consumption rises approximately linearly with increases in comfortable ambulation speeds (Waters and Mulroy, 1999). In SCI, the relationships between oxygen rate, oxygen cost and speed are dependent on lower extremity strength, since the greater the loss of lower limb support the greater the reliance on the upper-extremity for weight bearing, which is more energetically demanding. Studies in clinical populations have found that the rate of rise in energy expenditure usually does not exceed levels seen in non-clinical populations during locomotion unless strenuous exertion of the upper extremities is required (Waters et al, 1978). Specific to those with SCI, energy expenditure during locomotion has been shown to be similar to able-bodied individuals only in those who have intact lumbar and sacral function and a sufficient range of hip abduction and extension to independently maintain erect posture.

Other factors that have been shown to influence oxygen consumption during locomotion include spasticity and load. Spasticity and peripheral loading of the body (e.g. foot versus trunk) increases oxygen consumption during locomotion (Teixeria da Cunha-Filho et al, 2003; Waters and Mulroy, 1999). Body weight supported gait retraining interventions (e.g. treadmill training) manipulates the amount of load to replace uncontrolled spinal reflexes and compensate for motor losses which occur following an injury allowing individuals with paralysis to practice locomotion earlier in their rehabilitation.

Cardiovascular Function and Peak Aerobic Capacity

Function

Cardiovascular adaptations occur within the first two months following an injury including decreased lumen diameter and arterial blood flow, and increased vascular resistance of the femoral artery (Hicks and Martin-Ginis, 2008). There is evidence to suggest SCI have higher total triglyceride and cholesterol (e.g. low-density lipoprotein cholesterol) levels, increased insulin resistance and arterial stiffness and difficulty regulating blood glucose levels (Hicks and Martin-Ginis, 2008). These are all considered risk factors for cardiovascular disease and type 2 diabetes.

Generally, individuals with tetraplegia have lower than normal blood pressure, while individuals with lesions below T6 tend to have more normative levels (Phillips et al, 1998). Additionally, individuals with tetraplegia are more prone to cardiac complications including autonomic (hyperreflexia) dysreflexia and orthostatic (postural) hypotension. ***Autonomic (hyperreflexia) dysreflexia*** is a transient hypertensive state caused by non-specific noxious stimuli (e.g. distended bowel or bladder) below the lesion level, which triggers a sympathetic reflex *increasing* systolic blood pressure (e.g. 20-40 mm Hg above baseline) and decreasing HR (Phillips et al, 1998). The resulting hypertension can potentially be life threatening and persists until stimulus removal. On the other hand,

orthostatic (postural) hypotension, which is also more common in cervical injuries, consists of an acute or progressive *decrease* in mean blood pressure (> 10-15 mm Hg) in the erect position (e.g. from sitting to standing).

Peak Aerobic Capacity

Maximal aerobic capacity (VO_2 max) is the highest oxygen uptake an individual can achieve during physical work at sea level and has been used as an indicator of physical fitness (Waters and Mulroy, 1999). The achievement of VO_2 max is based on the primary criteria of a leveling off of oxygen uptake (e.g. < 2.1 mL/kg/min) with an increase in work rate. Secondary objective criteria include high levels of lactic acid in the blood post exercise (e.g. 7.9-8.4 mM), elevated respiratory exchange ratio (e.g. ≥ 1.15) and the ability to achieve a certain percentage of an age-adjusted estimate of maximal HR (e.g. ≥ 90%) (Howley et al, 1995). Body size and composition affect the amount of oxygen consumed therefore it is often expressed relative to body weight to allow for comparison between individuals (mL/kg/min) (Waters and Mulroy, 1999).

Oxygen demand is directly related to the muscle mass involved, therefore the VO_2 max during upper limb exercise is less than with the lower limbs, with only modest predictability of peak oxygen uptake between modalities. VO_2 max during arm ergometry in men generally varies between 64-80% of values attained during leg ergometry (Phillips, 2008). HR max values also are lower during arm compared to leg ergometry with upper body testing resulting in 88-98% of lower body values (Phillips, 2008). For these reasons arm exercise is considered to be less efficient and less effective than lower body exercise in developing and maintaining cardiovascular fitness. Interestingly, while maximum physiologic responses are generally lower during arm exercise compared to leg, for a given submaximal workload, oxygen uptake, HR and perceived exertion are greater when completed with the upper extremity.

Despite the limitations of arm ergometry testing this modality provides useful information for individuals whose occupation and recreational activities primarily involve the upper extremity or who have disabilities affecting the lower limbs (e.g. SCI). In general, the SCI lesion level will determine the amount of active muscle mass and sympathetic tone (e.g. vasoregulation) available, which will influence venous “muscle pump” in the legs and the potential to increase oxygen uptake (Phillips, 2008). Typically, the higher the lesion level, the lower the physical capacity to increase oxygen uptake with some exceptions. A study conducted by van Loan and colleagues (1987) compared responses to maximum arm ergometry between individuals with SCI and non-injured controls. VO_2 max values were highest for non-injured adults, followed by individuals with paraplegia and tetraplegia (28.2 vs. 25.3 vs. 12.0 mL/kg/min respectively). The same trends were present when considering max HR values, with the highest values attained in non-injured participants followed by those with paraplegia and tetraplegia (168 vs. 160 vs. 109 BPM respectively) (van Loan et al, 1987). Interestingly, elite athletes with an SCI have been shown to achieve VO_2 peak values ranging from 32 mL/kg/min to 44.9 mL/kg/min (Gass and Camp, 1979).

Peak HR response obtained from an arm ergometry test can be used to prescribe exercise intensity. This is because there is a linear relationship between the rate of oxygen uptake and HR in non-injured individuals (Franklin, 1985) and those with paraplegia (Hooker et al, 1993). For individuals with complete tetraplegia however, there is dissociation in this relationship as they rely predominantly on vagal withdrawal to augment HR during arm ergometry (Hooker et al, 1993). For this subgroup it would be more appropriate to use a percentage of peak arm ergometer test results or ratings of perceived exertion (RPE) to prescribe and monitor aerobic exercise intensity.

1.2 BODY WEIGHT SUPPORTED LOCOMOTOR TRAINING

Many individuals with chronic incomplete SCI express the importance of recovering independent ambulation as one of their rehabilitation goals and the majority of these individuals do regain some locomotor function especially when provided with appropriate training interventions. Body weight supported locomotor training is a modern approach to SCI rehabilitation, which provides an *interactive* training environment that is believed to promote “rewiring” of synaptic connections (***neuroplasticity***) (Fenuta and Hicks, 2011). Additionally, increases to corticospinal drive to the lower limb muscles, have been associated with improved locomotor ability following body weight supported treadmill training interventions (Thomas and Gorassini, 2005).

1.2.1 History of Training Method: Animal Models

Animal models (cats, rats, monkeys) have provided the basis for understanding locomotor capacity following SCI and the principles behind locomotor retraining as a form of rehabilitation. Studies suggest that the spinal cord is plastic, especially in younger animals (Barbeau and Rossignol, 1987), and can be trained when provided with a ***task-specific*** stimulus. For example, spinally transected animals trained to stand (e.g. using a harness) are unable to step, with the opposite condition also true; cross training is however possible with time (Field-Fote, 2000). This emphasizes the fact that if the goal is to learn to walk, one must practice walking. The important role of the ***cerebellum*** in coordinating bilateral leg activation during gait has also been proposed in cats (Dietz, 2004).

Similarities exist between stepping behaviour in spinalized cats (quadrupedal locomotion) and human infants (bipedal locomotion). Both are able to adjust stepping rate in response to changes in treadmill speed and rely on hip position and load for coordination of lower limbs (e.g. initiation of swing dependent on hip position) (Fenuta and Hicks, 2011). Spinalized cats are able to adapt to increasing treadmill speeds, but are limited to maximum speeds between 0.8-1.0 m/s (Pépin et al, 2003). Infant stepping is regulated by spinal neurons which generate the basic motor pattern (***central pattern generators***) and with maturity the central nervous system descending pathways (e.g. brainstem, thalamus) take over control, modifying gait parameters in response to environmental cues (e.g. peripheral cutaneous and proprioceptive input) (Fenuta and Hicks, 2011). Studies of individuals with neurologically complete SCI offer evidence for the existence of central pattern generators in adult humans (Dietz, 2008). Despite the loss of supraspinal influences post injury spinal neurons in these individuals can spontaneously initiate movement and/or can be triggered to do so through task-specific interventions.

Body weight supported treadmill training has shown to improve the rate of recovery, decrease the amount of muscle atrophy experienced post injury and even increase force-generating capacity beyond what would be expected by spontaneous recovery alone (Hicks and Martin-Ginis, 2008). EMG activities in step-trained cats and humans begin to normalize at comparable speeds following treadmill-training interventions. This improvement in motor response cannot be explained by muscle stretch reflexes alone (Dietz, 2008), but are also attributed to the increased chances of activating muscles in a temporally appropriate manner (“at the right place, at the right time”) with training (Fenuta and Hicks, 2011). Unfortunately these benefits do not translate to improved overground locomotor ability for those with complete SCI due to the loss of neural connections, coordination and balance. Additionally, these benefits are more difficult to maintain in this subgroup without continued provision of a task specific stimulus who demonstrate a “use it or lose it” phenomenon.

1.2.2 Available Rehabilitation Devices (Treadmill vs. Overground)

Currently, there are a variety of training options available to individuals with SCI when considering a body weight supported gait rehabilitation program (e.g. land vs. water based, combined with functional electrical stimulation), each with its own respective advantages and disadvantages. A harness is often used to provide ***body weight support (BWS)*** and assist balance and has been shown to significantly reduce the amount of knee flexion during gait (Pépin et al, 2003). The devices used in the presented thesis are described below and depicted in Appendix B.

Treadmill-Based: Lokomat^{TM} vs. Manual

The Andago GmbH treadmill (Loko, Germany) can be used with or without the robotic driven gait orthosis (LokomatPro, Hocoma, Switzerland). The unweighting system is dynamic and motor-driven, allowing for constant BWS (Winchester and

Querry, 2006). Manual treadmill training sessions are often completed with therapists sitting beside each leg on either side of the treadmill and manually moving the legs through the gait cycle (Appendix B). Training with the robotic orthosis (e.g. Lokomat^{TM}) requires use of the same unweighting system. The orthosis is carefully lined up to the hip and knee joints, and then secured to the individual with chest and pelvic straps, as well as cuffs that go around the thigh, below the knee and above the ankle. Ankles are positioned in neutral dorsiflexion using spring-assisted elastic straps to assist in limb clearance during the swing phase of gait. The robotic device produces normal physiologic gait patterns and is able to control joint kinematics synchronized to gait speed (patient coefficient) using computer-controlled motors at the hips and knees. The training system is designed to limit lateral or anterior/posterior motion, while allowing normal vertical movement of an individual during locomotion. The system software offers assessment tools that can measure movement characteristics using force transducers connected to the joint motors. The L-Force module, for example, measures the isometric force exerted by an individual for hip and knee flexion and extension while secured to the robotic limbs (Winchester and Querry, 2006).

Reducing the amount of guidance force is important for relearning as it allows individuals to be more actively involved in a session as they attempt to independently achieve limb movement instead of being moved by the device. This is consistent with the ***guidance hypothesis,*** which suggests continuous passive assistance during practice negatively affects later performance and retention compared with unconstrained practice (Israel et al, 2006). Studies in individuals with SCI have demonstrated the importance of being actively engaged in Lokomat™ sessions, as those who simply allow the robot to passively move their legs through the range of motion (ROM) experience decreased EMG activity and metabolic expenditure (Kressler et al, 2013).

Similar gait kinematics can be achieved during Lokomat™ and Manual Treadmill sessions with both able to maintain a variable stepping pattern, which research supports as an effective gait training strategy (Fenuta and Hicks, 2011). However, differences do exist with respect to speed, hip and ankle extension, ROM and stepping consistency, which can better be achieved with robotic assistance. Depending on the ability of the individual, manual treadmill sessions can be conducted with or without assistance from therapists at one or both legs. Therapists are typically used for individuals who cannot independently complete the gait cycle with trainer skill shown to correlate with achieving better knee extension during stance and improved toe clearance (Galvez et al, 2011). Better knee extension in this study correlated with larger knee horizontal assistance force, while toe clearance correlated with better knee and hip extension. Inconsistency in the stepping pattern with trainer fatigue or patient spasticity, however, has been identified as a potential concern.

Overground Based: ZeroG™

Overground training provides individuals with greater degrees of freedom and is the most task-specific thus facilitating stepping transfer to functional locomotion. The newly developed ZeroG™ system (ZeroG™; Aretech, LLC, Ashburn, VA) by Hidler and colleagues (2011) has an unloading system capable of providing 300 lb of static and 150 lb of dynamic (constant force) BWS using a custom-series elastic actuator. The unloading system is mounted to an active trolley, which automatically follows an individual (0-4 mph) on the overhead rail as they practice gait and balance activities. This unique set up minimizes the horizontal forces felt by the individual using the system.

Dynamic BWS has been shown to result in more natural ground-reaction forces and gait characteristics compared to static BWS, allowing individuals to move through a greater ROM when practicing activities of daily living (ADLs). A recent study by Fenuta and Hicks (2013) collected normative EMG data from able-bodied participants during locomotion using the ZeroG™ at varying degrees of BWS (0/20/40/60/80% BWS). It was concluded that the dynamic BWS provided by the ZeroG™ decreases muscular demands of the lower limbs without significantly altering patterns of muscle activation throughout gait.

1.2.3 Providing an Interactive Training Environment to Promote Neuroplasticity and Enhance Locomotor Function

Prior to designing any training program it is important to know if individuals have any contraindications to participating in rehabilitation sessions using these devices; a few examples are presented in Appendix B.

Similar to cat models, there are two afferent (sensory) inputs important to facilitate appropriate leg muscle activation for stepping post SCI: a) *hip extension*, which stretches the hip flexors to trigger swing phase and b) *load*, which affects the amount of muscular demand and forces experienced by the lower extremity (Fenuta and Hicks, 2011). It is important that despite offsetting a portion of body weight a *challenging* stimulus is still provided to intact neuromuscular structures. The goal is to eventually require less than or equal to 30% of body weight offset as these levels have been shown to most closely resemble independent overground locomotion, and better allows individuals to maintain upright posture (Fenuta and Hicks, 2011).

Body weight supported treadmill training is more effective than conventional physiotherapy for improving gait parameters, with trained individuals achieving increased velocity, distance, step length, swing phase, hip extension and plantarflexion during stance compared to non-trained individuals (Fenuta and

Hicks, 2011). A recent study found responders to treadmill training, who eventually recover functional overground locomotion, acquire a) greater tibialis anterior (TA) and hamstring muscle activity, b) decreases in duration and cocontraction of quadriceps and hamstrings, and demonstrate c) 3-4 x greater quadriceps and hamstring activity compared to non-injured controls during locomotion at comparable speeds and level of BWS (Gorassini et al, 2009). Regardless of the specific training modality (treadmill vs. overground) all have provided evidence of improved overground speed of locomotion post intervention. However, one study found slower walkers and faster walkers at baseline saw the greatest improvements from treadmill and overground training respectively (Protas et al, 2001). This emphasizes the importance of evaluating an individual pre-training in order to provide a rehabilitation program that is most appropriate to challenge current abilities in order to maximize recovery of locomotor function.

Normalization of the gait pattern is possible for complete and incomplete SCI regardless of responsiveness to body weight supported treadmill training (Dietz, 2008; Gorassini et al 2009). With training, inappropriate tibialis anterior activity decreases, while appropriate gastrocnemius activity increases during stance phase. Increases in the magnitude of EMG patterns are above and beyond what would be expected by enhanced stretch reflexes alone (Fouad and Pearson, 2004). This altered activation pattern results in greater weight bearing capacity of the extensors with training. Leg extensor EMG activity has been shown to remain elevated even three years after training in those with incomplete SCI who regularly maintain locomotor activity (Dietz, 2004). Long-term participation in these training programs will help to ensure continued benefits are experienced by individuals post injury, which is particularly important for those with complete injuries, who as previously mentioned, demonstrate a “use it or lose it” phenomenon.

To date only one study has investigated chronic training effects of different modalities. Alexeeva and colleagues (2011) compared overground and treadmill locomotion to standard physical therapy in participants with chronic, incomplete SCI (n=35). After 13 weeks of training 1 hour per day, 3 times per week, aerobic capacity increased but this change was not significant for any group. Significant increases in strength and walking speed/distance did occur, however, in all groups, with balance increasing significantly only for the overground training group. Based on their findings, these researchers suggest the “optimal” approach for improving locomotor ability in this population is overground locomotor training with 30% BWS that takes place 1 hour per day, 3 times per week, for 10 weeks in duration.

1.2.4 Evidence of Benefits Additional to Improved Locomotor Function

It is important to note that the benefits of body weight supported locomotor interventions extend beyond strictly improvements in locomotor function, which is of particular importance for those with complete SCI (who may not experience changes in locomotor ability).

Cardiovascular Function

Locomotion targets the lower limbs and has an associated metabolic cost. Individuals become more efficient at completing the task of walking following a treadmill training intervention, with metabolic costs decreasing by as much as 68% (Protas et al, 2001). Training with electrical stimulation either on a treadmill or overground has been shown to be even more effective in improving walking VO_2, velocity and economy (Kressler et al, 2013).

Supported treadmill training has shown beneficial effects on peripheral circulation (e.g. femoral artery compliance) (Ditor et al, 2005). There is evidence that in both injury subtypes (complete and incomplete) interventions can have a positive impact on improving the blood lipid profile, glucose tolerance, insulin sensitivity and autonomic regulation (e.g. HR, blood pressure) (Hicks and Martin-Ginis, 2008).

Muscular Function

Body weight supported treadmill training provides a hypertrophic stimulus and therefore can be used to prevent and treat the muscle atrophy experienced post SCI. Interventions implemented in the acute phase of injury, as little as two times per week, is sufficient to prevent the expected muscle atrophy and fiber type redistribution, which has beneficial effects on increasing glycemic regulation (Adams et al, 2004; Giangregorio et al, 2005).

1.3 RATIONALE OF MASTER'S THESIS

Optimal training strategies to enhance locomotor function have yet to be established for individuals with incomplete SCI. To this author's knowledge, only one study, by Israel and colleagues (2006), has *simultaneously* collected metabolic and EMG measurements during robotic and therapist assisted treadmill locomotion in individuals with incomplete SCI. Simultaneous measurements allow for examination of potential relationships between muscle activity patterns and whole-body metabolic costs during locomotion. These researchers determined that the metabolic cost of robotic-assisted locomotion was significantly *lower* compared to therapist assisted locomotion when subjects were asked to *match* the kinematics of the robot ("walk with the robot") (9 mL/kg/min vs. 14 mL/kg/min, 14 mL/kg/min vs. 15 mL/kg/min respectively). This difference was partially attributed to reduced metabolic activity observed during quiet standing in the robotic assisted condition and decreased hip flexor EMG activity during locomotion. Interestingly, when asked to *maximize* efforts the above-mentioned differences were minimized. Based on their results, it was concluded that *voluntary* effort is required during robotic-assisted locomotion to achieve metabolic costs and hip flexor EMG activity similar to those associated with therapist-assisted locomotion. It was suggested that therapist-assisted treadmill training programs should be used as progressions from robotic-assisted sessions in order to enhance the benefit of locomotor rehabilitative strategies. Additional emphasis was placed on the need for randomized controlled trials to compare the relative effectiveness of training with robotic and therapist-assisted treadmill locomotion in individuals with incomplete SCI.

1.3.1 Purpose and Objectives

The presented thesis hopes to contribute to the literature by comparing oxygen demand and muscle activity during locomotion across three body weight supported devices (Lokomat^{TM}, manual treadmill *without* therapist assistance, ZeroG^{TM}) at a *constant* BWS. This will be the first study of its kind to include an evaluation of the newly developed overground supported training system known as the ZeroG^{TM}.

To date, no study has systematically investigated which assisted locomotion devices consumers prefer to use during their rehabilitation sessions and more importantly *why* individuals would choose one modality versus another. The available training devices can be expensive and labour intensive on the part of the therapist and thus the presented comparative study will help provide information to determine which option(s) will provide the most benefit to the patient physiologically and psychologically, while also considering effective use and allocation of resources.

In summary, the objectives of the presented thesis include the following:

- To investigate oxygen demand and muscle activation patterns at the same level of body weight support (BWS) while participants [incomplete spinal cord injury (SCI) gender- and age-matched with able-bodied adults] complete a locomotor training session using a) the treadmill with robotic assistance (Lokomat^{TM}), b) the treadmill without therapist assistance (Manual Treadmill), and c) the overground training system (ZeroG^{TM}) (Appendix B).
- To acquire information regarding palatability of different body weight supported training devices from the consumers who will be using them in a rehabilitation setting.

1.3.2 Hypotheses

Based on the primary objectives of the presented thesis the following hypotheses were proposed:

- The highest levels of oxygen uptake and muscle activation will occur during locomotion using the ZeroG^{TM} and this will be higher in persons with incomplete SCI versus gender and age matched able-bodied adults.
- Individuals will prefer the ZeroG^{TM} device because it most resembles and is thus the most transferrable to unsupported overground locomotion.

CHAPTER 2: METHODS

2.1. PARTICIPANTS

2.1.1 Recruitment

Participants were recruited using posters in the MacWheelers SCI Rehabilitation Program at McMaster University (Hamilton, ON) and Regional Rehabilitation Centre at Hamilton General Hospital (Hamilton, ON) (Appendix C). A version was also published in the 2013 spring issue of SCI Ontario's magazine (Outspoken) and was posted on their on-line website (E-Spoken) (Appendix C). Interested participants were asked to contact the student investigator and were screened to confirm their eligibility to participate in the study and set-up testing sessions.

2.1.2 Eligibility Criteria

Eligible participants consisted of individuals with incomplete SCI, and healthy able-bodied adult controls between the ages of 18-65 years of age; the normative population was gender-and age-matched (± 5 years) to the SCI participants. Individuals with SCI were able to participate if their injury was a) chronic (1+ year post), b) incomplete (AIS C-D, sensory and/or motor function at S4/5) and c) if they could comfortably complete walking sessions using the Andago GmbH treadmill (Loko, Germany) with manual and robotic (LokomatPro, Hocoma, Switzerland) assistance, as well as overground using the ZeroG^{TM} (ZeroG^{TM};

Aretech, LLC, Ashburn, VA). For the ZeroG^{TM} session, participants were required to be able to walk with less than or equal to 68 kg (150 lbs) offset, as this was the maximum amount of dynamic BWS (constant rope tension) the machine could provide. Able-bodied controls (CON) were defined as individuals who a) did not have a musculoskeletal disease affecting the lower extremity (e.g multiple sclerosis, cerebral palsy) b) had not experienced an injury to the lower extremity (e.g. ligament sprain, muscle strain) within the past year and c) were able to comfortably walk 30 m without the use of assistive devices. Device limitations that needed to be considered for both SCI and CON included the following: all of the devices used had only been tested for individuals with a maximum body weight of 135 kg (298 lbs) and the Lokomat^{TM} was only designed to accommodate adults with femur lengths of approximately 35-47 cm and pelvic widths of 29-51 cm. Participants needed to comfortably fit into each device's safety harness and the 3 Lokomat^{TM} cuffs (around thigh, below knee and above ankle) that attached to the robotic limbs.

2.2 BASELINE TESTING AND DEVICE FAMILIARIZATION

All testing and data collection was completed at the Robert Fitzhenry Specialized Rehabilitation and Exercise Lab in the Ivor Wynne Centre (McMaster University, Hamilton, ON). Participants did not consume any food (except water) in the four hours preceding testing and did not consume caffeine or alcohol or participate in strenuous exercise in the 24 hours prior. Participants were informed prior to testing to wear comfortable clothing including a t-shirt, shorts, socks and running shoes. During the pre-screening process, the participants were familiarized with the laboratory environment, equipment and test procedures. Written informed consent was obtained from all participants with a protocol approved by the McMaster Research Ethics Board (Appendix C).

2.2.1 Anthropometric Measurements

At the first session, demographic and anthropometric measures of height (cm) and weight (kg) were recorded (Appendix D). Participants with SCI who could not stand upright for the height measurement using a stadiometer or weight measurement using a standard electronic scale, had the height assessment performed using a measuring tape while lying supine (top of cranium to bottom of heel), and weight taken using an electronic scale that was wheelchair compatible (BRW 1000 Wheelchair Scale, Detecto, Webb City, MO). If this scale was used participant weight was determined as follows: Weight of Person (kg) = (Weight of Person + Wheelchair) – (Weight of Wheelchair). The CON group had their height measured using a stadiometer and weight measured using a standard electronic scale. Upper thigh and lower leg lengths were also recorded (using a measuring tape) for set-up of the Lokomat^{TM}, which was used for isometric strength testing and a body weight supported walking session. Upper thigh length (cm) was measured from the greater trochanter to lateral knee joint space and lower leg length (cm) was measured from the lateral knee joint space to the ground. Finally, participants were sized for the 3 Lokomat^{TM} cuffs (above thigh, below knee and above ankle) that attached to the robotic limbs.

2.2.2 Body Weight Support (BWS) Determination Using the ZeroG^{TM}

Participants were fitted to the overground training system to determine the appropriate size harness and minimal amount of BWS required to complete a walking trial. The amount of support required was determined to be appropriate if the participant was able to take steps on the ZeroG^{TM} without buckling at the knees. Participants were encouraged to complete a couple of lengths of the track to become familiar with the device and finalized support settings.

2.2.3 Peak Aerobic (VO_2 Peak) Test on an Arm Ergometer

Participants performed a VO_2 peak test to volitional fatigue on an electronically braked arm ergometer (Angio V2; Lode BV, Groningen, The Netherlands) (Appendix D). The MOXUS Modular Metabolic System (AEI Technologies Inc., Bastrop, TX) was calibrated prior to each test using a syringe, known gas concentration and MOXUS software system. The arm ergometer was firmly secured to a wall using a height-adjustable bracket. The height of the arm ergometer was adjusted so that from a seated position the centre of the participant's shoulder joint was in line with the centre of the arm ergometer crank axis. Participants sat upright with feet flat on the ground and back firmly against the chair at a distance from the arm ergometer such that the elbow was almost fully extended at the furthest point in the range of motion during cranking. Participants with insufficient handgrip had their hands secured to the arm handles with tensor bandages.

A mask attached to the metabolic cart and nose clip was worn in order to continuously measure expired gas and ventilatory parameters, which were sampled at 30-second epochs throughout the test. Resting data was collected for 5 minutes; oxygen uptake [AEI Metabolic System (Moxus) software, Pittsburgh, PA], heart rate (Polar T31 heart rate monitor; Polar Electro Inc, Woodbury, NY, USA) and brachial blood pressure (DINAMAP Pro 100, GE Healthcare, Finland) were recorded at 1, 3 and 5 minutes (Appendix D). Throughout the progressive (ramp) exercise testing protocol each participant was encouraged to give a maximal effort and maintain cadence between 60 to 80 revolutions per minute (RPM); a speedometer provided visual feedback. Subjects were verbally encouraged throughout the duration of each test, but were not informed of the time that had elapsed or of the workload they had achieved.

The test began with a 2-minute warm-up without resistance (0 W). The resistance increased thereafter every minute by ~5 W for participants with tetraplegia, and

~10 W for participants with paraplegia (Hol et al, 2007) and able-bodied adults (Verellen et al, 2011). The exact progression was individualized in order to ensure test duration was between 8-12 minutes in length as recommended by the American College of Sports Medicine (ACSM, 2006). Expired gas and ventilatory parameters were measured continuously throughout the protocol and averaged every 30 seconds. Heart rate (HR) was also sampled every 30 seconds, and ratings of perceived exertion (RPE) using Borg's 0-10 scale (Appendix D) was assessed every minute to gauge subjective perception of physical effort both centrally (heart and breathing) and peripherally (arms). These measures of effort were intended to also serve as an anchor for participants (how to rate progressively harder exercise on a scale of 0 to 10) in order to assess their physical effort during subsequent body weight supported locomotion sessions.

Participants continued arm cycling until a) volitional fatigue, b) they were unable to maintain imposed cadence for 15 seconds or c) there was an obvious plateau of oxygen uptake from one minute to the next. Following cessation of the test, participants were instructed to rest quietly for 5 minutes before beginning a cool-down at 30 W at a self-selected crank rate. Brachial blood pressure, HR [beats per minute (BPM)] and oxygen consumption measurements were obtained 1, 3 and 5 minutes following cessation of the test in order to ensure appropriate recovery.

2.2.4 Isometric Strength Testing Using the Lokomat™

Participants were fitted with the appropriate size harness for the body weight supported training device and strapped into the robotic limbs to provide lower limb isometric strength production measurements (Nm). Using the L-FORCE v2.0 tool in the Lokomat™ software program (Lokomat™ System V5.0), participants were fully suspended from the device's unweighting system with lower limb joints locked in position (30° hip and 45° knee flexion) to measure isometric strength of hip flexion and extension, as well as knee flexion and extension on the right and left sides respectively (Appendix D). Participants were asked to exert a maximal voluntary isometric contraction against the robotic limbs so that force transducers within the device could measure the exerted force from each of the aforementioned movements. The L-FORCE v2.0 has been technically and clinically validated (Bolliger et al, 2006).

Following these strength assessments, participants were provided the opportunity to walk on the Andago GmbH treadmill (with and without the Lokomat™) at the assigned BWS from the ZeroG™ assessment for 5-10 minutes to become familiar with the device and to establish comfortable walking speed on the treadmill. The harness size and device (Lokomat™ and Andago GmbH treadmill) settings were recorded for future assessments (Appendix D).

2.3 RANDOMIZED BODY WEIGHT SUPPORTED WALKING SESSIONS

The subsequent 3 body weight supported walking sessions all used the same amount of support that had been established at the first session using the ZeroG™. Sessions were conducted on separate days if fatigue was determined to be a limiting factor. Participants completed walking trials in a randomized order using the Andago GmbH treadmill system (with or without the Lokomat™) or overground ZeroG™. Electromyography (EMG) electrodes (Delsys Trigno Wireless EMG Surface Electrodes, Delsys, Boston, MA) were placed on tibialis anterior (TA), rectus femoris (RF), biceps femoris (BF) and medial gastrocnemius (MG) muscles of both legs for a total of 8 EMG channels. Footswitches (Delsys Wireless Force Sensitive Resistors, Delsys, Boston, MA) were placed on the bottom of both feet (sensor on base of first toe and calcaneus respectively) to synchronize the muscle activation signals with heel strike as the subject walked. All sensors were calibrated prior to each session according to Delsys system configurations. The subjects put on a Polar HR Monitor and then were fitted to the appropriate harness and device settings based on the pre-screening evaluation and randomization order. Finally, the participant was attached to the portable metabolic cart (MedGraphics V02000, Medical Graphics Corp., St. Paul, MN) using the patented preVent mask™, which was calibrated using the system's auto-calibration feature prior to each session. Resting measures (5 minutes) were taken before and after each walking session; oxygen uptake (sampled in 30-

second epochs), HR and brachial blood pressure were recorded at 1, 3 and 5 minutes (Appendix D). Participants then completed a minimum of 2 minutes of steady state walking [(± 5 ml/kg/min and ± 5 BPM)] at a comfortable self-selected speed at the same level of BWS regardless of the device. Oxygen uptake and HR were recorded at 30-second epochs and RPE were monitored every minute. RPE was monitored to gauge perceived central (heart and lungs) and peripheral (limbs) effort; Borg's 0-10 scale was anchored to the VO_2 peak test (Appendix D). Muscle activity was continuously monitored throughout the duration of the steady-state walking session and collected using EMGworks Acquisition software program (Delsys EMGworks 4.0, Delsys, Boston, MA). For the Lokomat™ walking trial, all participants were asked to contribute as much to the "walking motion" as possible; the Guidance Force Control system was set to 100% for all sessions indicating full robotic assistance. When the robotic orthosis was not used, participants stepped on the treadmill without assistance and no corrections were made by assisting investigators to normalize stepping patterns. Participants were allowed to use the bilateral handrails on the treadmill to maintain postural stability, although they were asked to minimize upper-extremity weight bearing. For all walking trials, able-bodied participants completed 2-minute steady state sessions a) at baseline (0 BWS) and b) at a percentage of BWS determined by their matched SCI participant (% BWS). Following all sessions, regardless of randomization order, participants were required to take a couple of steps using the ZeroG™ in order to have muscle activity data to normalize and control for potential day-to-day variation in EMG signal conduction and/or electrode placement; the ZeroG^{TM} was used for this as it was assumed to provide maximum activation of muscle groups for both SCI and CON.

2.3.1 Consumer Preference

Immediately following each walking session the participants were asked to fill out a satisfaction questionnaire of the respective device used (Appendix D). Upon completion of all testing sessions participants were asked to complete one final questionnaire directly comparing their subjective experience using the different body weight supported devices (Appendix D). This information was valuable to acquire in order to gain an idea of consumer preference regarding the different training modalities.

2.4 DATA ANALYSIS

MOXUS [AEI Metabolic System (Moxus) software, Pittsburgh, PA] and VO2000 (Medical Graphics Corp., St. Paul, MN) software programs were used for analysis of peak and walking VO_2 metabolic measures respectively. EMGworks Analysis (Delsys EMGworks 4.0, Delsys, Boston, MA) was used for analysis and filtering of muscle activity signals. The rectified EMG records were low-pass filtered using a second order Butterworth filter with 3 dB passband ripple, 40 dB attenuation, and corner frequencies of 100 Hz and 400 Hz. SPSS software program was used for statistical analysis of acquired data. The Shapiro-Wilk test was used to test for normality of the data due to the small sample size; non-parametric statistics were used where appropriate. Statistical significance was set at $p < 0.05$ for all analyses. Bonferroni's correction was used for post-hoc testing when necessary. All reported values are expressed as mean ± standard error.

Main Objective Analysis

Oxygen consumption was expressed relative to resting metabolic equivalents (METs), which are considered 2.7 mL/kg/min and 3.5 mL/kg/min for SCI and CON respectively (Collins et al, 2010). Repeated measures ANOVAs were conducted on the steady state a) cardiovascular data [normalized to 1) resting (METS) and 2) VO_2 peak] and b) filtered root mean square (RMS) muscle activity signals over 3 consecutive gait cycles (normalized to ZeroG^{TM} filtered activity signals at the same level of BWS for each given day) to investigate differences in participant measures (SCI and CON respectively) while completing walking sessions using the Lokomat^{TM}, Manual Treadmill and ZeroG^{TM} devices. For CON this analysis was completed for both body weight supported conditions [e.g. 0 BWS vs. % BWS (matched)].

Additional Analyses

Bivariate correlations (Pearson) were conducted for participants with SCI to investigate potential relationships between the amount of required BWS and a) flexion:extension isometric strength at the hip and b) flexion:extension isometric strength at the knee. One-way ANOVAs were used to look at differences between SCI and CON.

CHAPTER 3: RESULTS

3.1 PARTICIPANTS

Demographic data of the 7 males with incomplete SCI (42.6 ± 4.29 years) and 7 gender and age matched able-bodied controls (42.7 ± 4.29 years) who participated in the study can be found in Table 1. There were no statistically significant differences between groups for all demographic measures including age, height, weight, and body mass index (BMI). The majority of incomplete SCI were traumatic in nature (e.g. motor vehicle accidents, falls), with only two participants having non-traumatic injuries (e.g. spinal stroke). Average duration of injury was 4.0 ± 0.62 years, with injury levels ranging from C3-L5 (2 tetraplegia, 5 paraplegia), and AIS C-D. Of the 7 participants with SCI, 3 had previous experience using the Lokomat™, 3 had previous experience using the Manual Treadmill and 3 had previous experience using the ZeroG™. No adverse events were experienced during study completion.

Table 1: Demographic variables for SCI and CON.

	SCI	CON	Between Groups (P-Value)
Age (years)	42.6 ± 4.29	42.7 ± 5.40	0.98
Height (cm)	179.1 ± 1.56	177.5 ± 2.91	0.84
Weight (kg)	89.6 ± 6.39	84.8 ± 5.88	0.52
BMI (kg/m^2)	27.9 ± 2.03	26.1 ± 1.46	0.48
BWS (kg)	37.8 ± 9.62	32.8 ± 7.40	0.67
BWS (%)	41.3 ± 10.16	41.3 ± 10.16	1.00
Time Since Injury (years)	4.0 ± 0.62	N/A	N/A

Mean ± standard error.

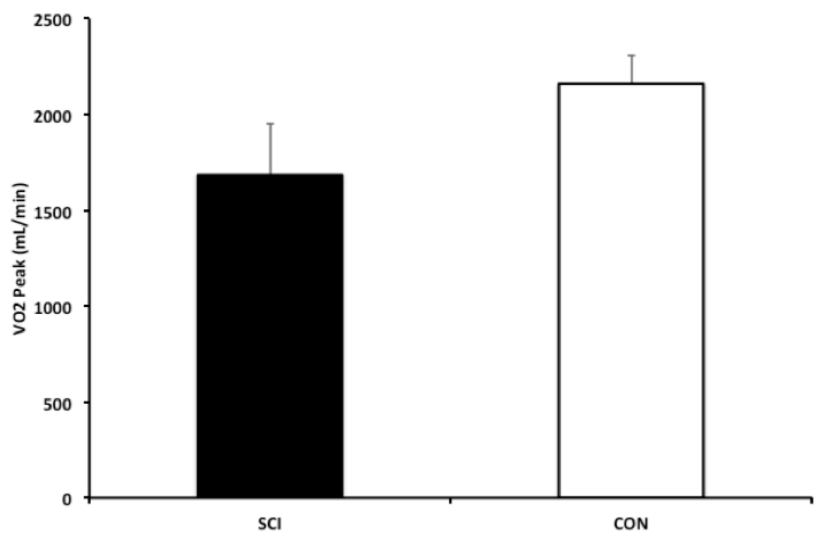
3.2 BASELINE TESTING AND DEVICE FAMILIARIZATION

3.2.1 Peak Aerobic (VO_2 Peak) Test on an Arm Ergometer

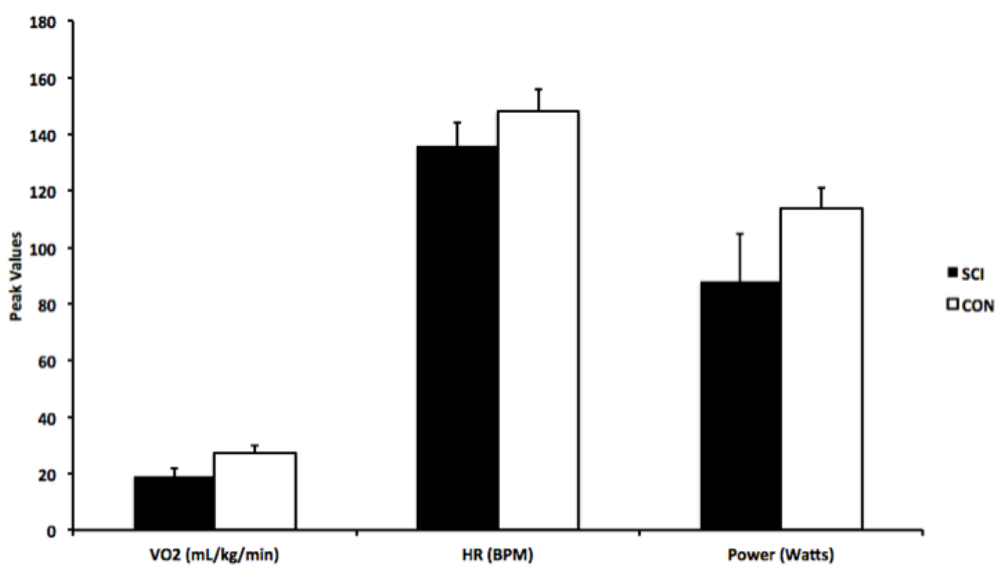
There were no statistically significant differences between groups (SCI vs. CON) for all measures obtained during the VO_2 peak test (Figure 1 a-c).

Figure 1: VO_2 peak test variables for SCI and CON.

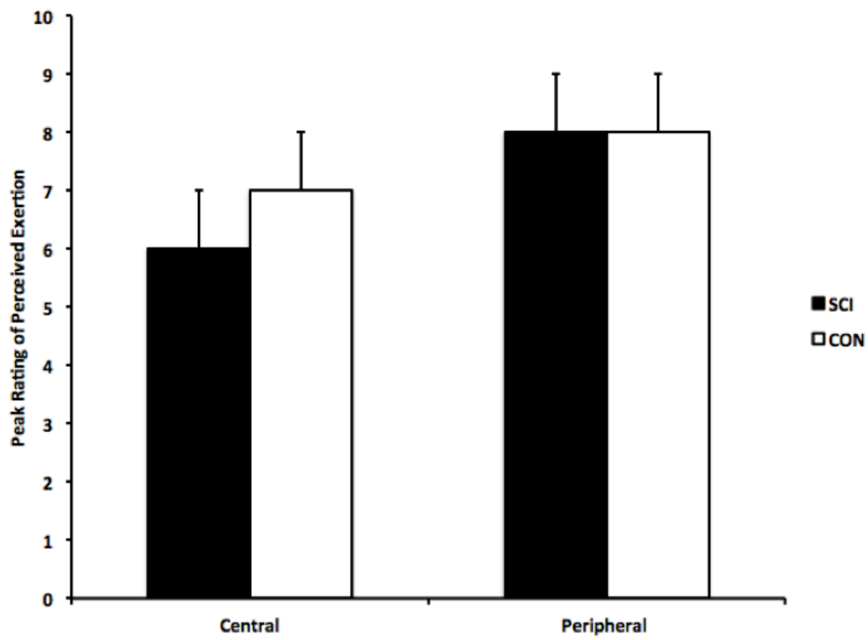
a) Absolute VO_2 peak (mL/min).



b) Relative VO_2 peak (mL/kg/min), peak heart rate (HR) expressed in beats per minute (BPM) and maximum power achieved (Watts).



c) Rating of perceived exertion (central – heart/lungs and peripheral - arms); 10 = maximal effort.

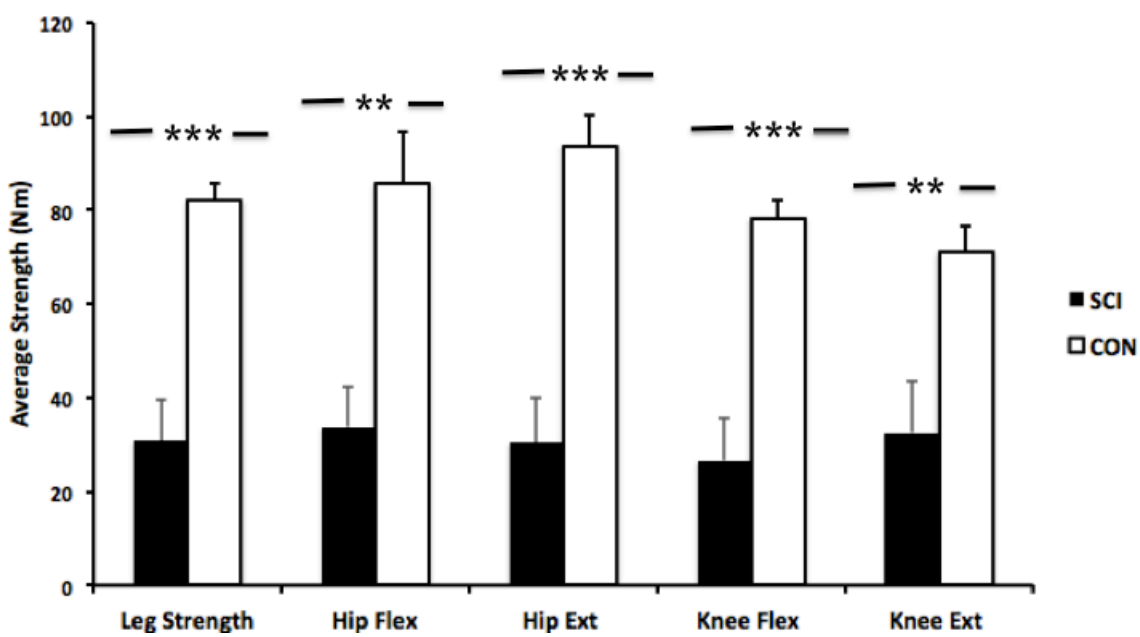


3.2.2 Isometric Strength Testing Using the Lokomat^{TM}

Statistically significant differences in lower leg strength between SCI and CON existed at all four maximum isometric efforts (Figure 2). The greatest difference was evident for hip extension and knee flexion in participants with SCI who produced only 31.2% and 29.4% respectively of the force seen in the CON group. While there were no statistically significant differences in the ratio between flexion:extension strength at the hip or knee between SCI and CON (Figure 3), there was a trend for a higher flexion:extension ratio at the hip in the SCI group.

Further, an interesting relationship was noted between the strength ratio at the hip joint and the amount of BWS required to complete a walking session using the ZeroG™ (e.g. without excessive knee flexion). A strong positive correlation (R^2=0.72) was found between flexion:extension ratio at the hip in participants with SCI and the amount of BWS required to complete the overground walking session; the higher the flexion:extension ratio, the more support that was required (Figure 4). The greater contributor to this relationship at the hip joint was the decrease in isometric hip extension strength compared to flexion strength (R^2=-0.69 vs. R^2=-0.36).

Figure 2: Isometric strength measures for SCI compared to CON.



P < 0.01 = **, P < 0.001 = ***.

Figure 3: Flexion:extension ratios at the hip and knee for SCI and CON.

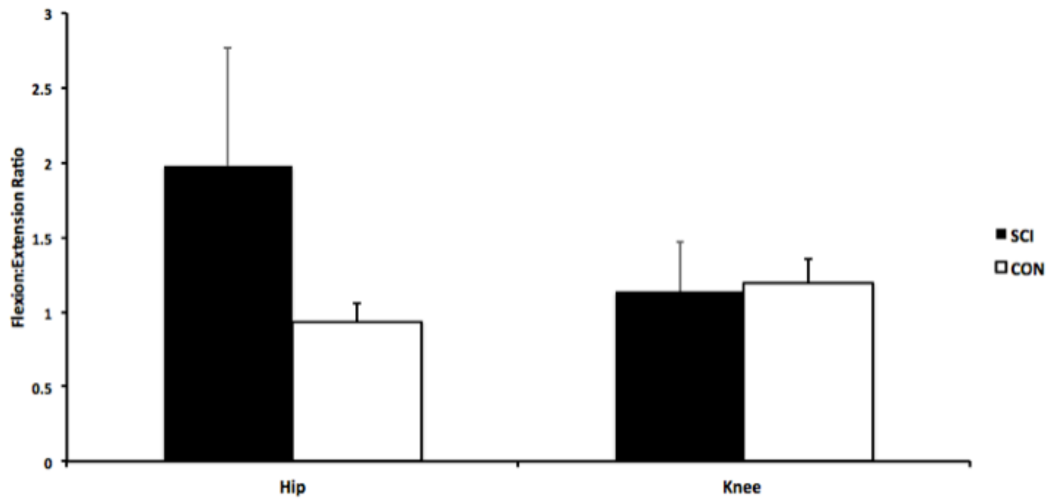
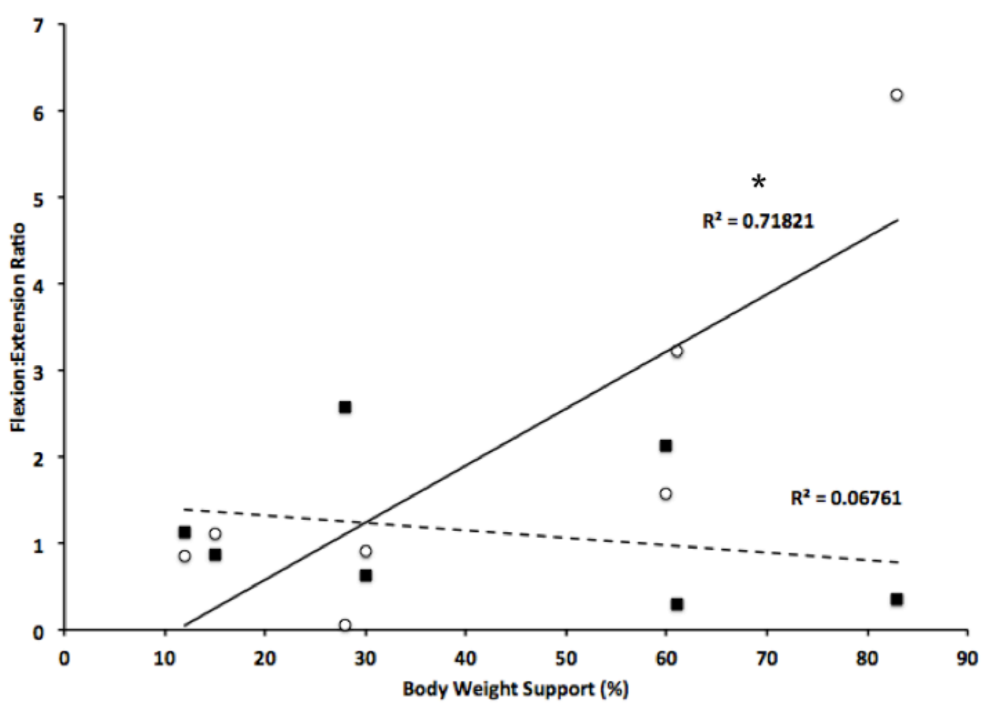


Figure 4: SCI flexion:extension ratio correlations with the amount of BWS required for locomotion.



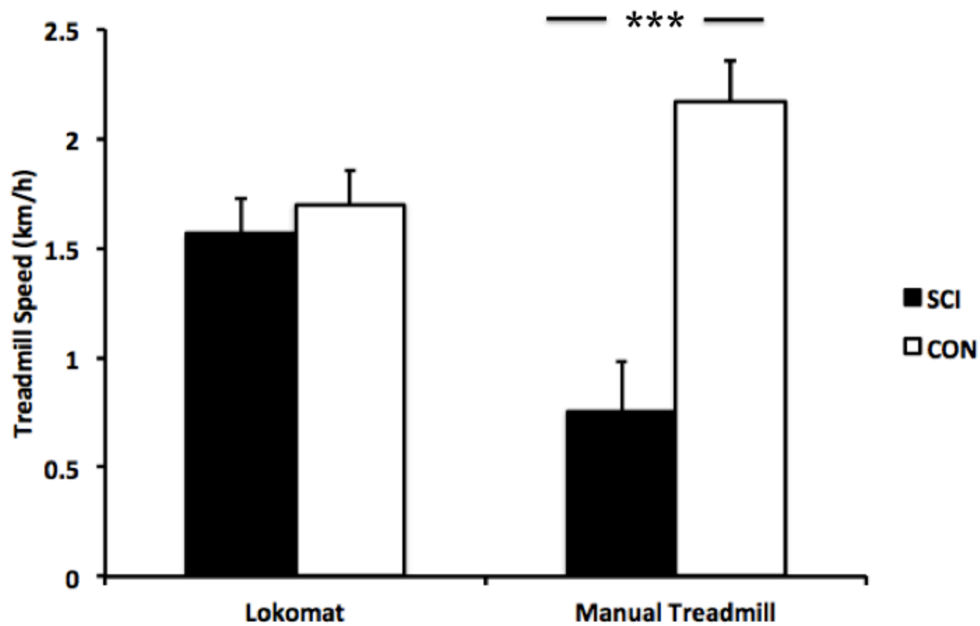
Open circles and solid line indicate relationship at the *hip*. Closed squares and dashed line indicate relationship at the *knee*.
$P < 0.05$ = *.

3.3. RANDOMIZED BODY WEIGHT SUPPORTED WALKING SESSIONS

All 7 matched pairs completed the Manual Treadmill and ZeroG™ walking sessions, but one of the pairs did not complete the Lokomat™ walking session due to spasticity (exaggerated stretch reflexes). The average amount of BWS that was used for the walking sessions was 41.3% ± 10.16% (Table 1). All participants reached steady state after 2 minutes of walking.

For the treadmill sessions (Lokomat™, Manual Treadmill) there were significant differences between participants with SCI and their matched controls in terms of comfortable walking speed while walking on the treadmill without the robotic limbs, with CON having a faster natural speed compared to participants with SCI (2.2 ± 0.18 vs. 0.8 ± 0.23 km/h) (Figure 5). While participants with SCI preferred to walk at much slower speeds while walking on the Manual Treadmill compared to the Lokomat™ (0.8 ± 0.23 vs. 1.6 ± 0.19 km/h), no statistically significant differences were found for walking speeds in their matched controls between the two treadmill devices (1.7 ± 0.18 vs. 2.2 ± 0.18 km/h).

Figure 5: Preferred speed for SCI compared to CON during treadmill locomotion.



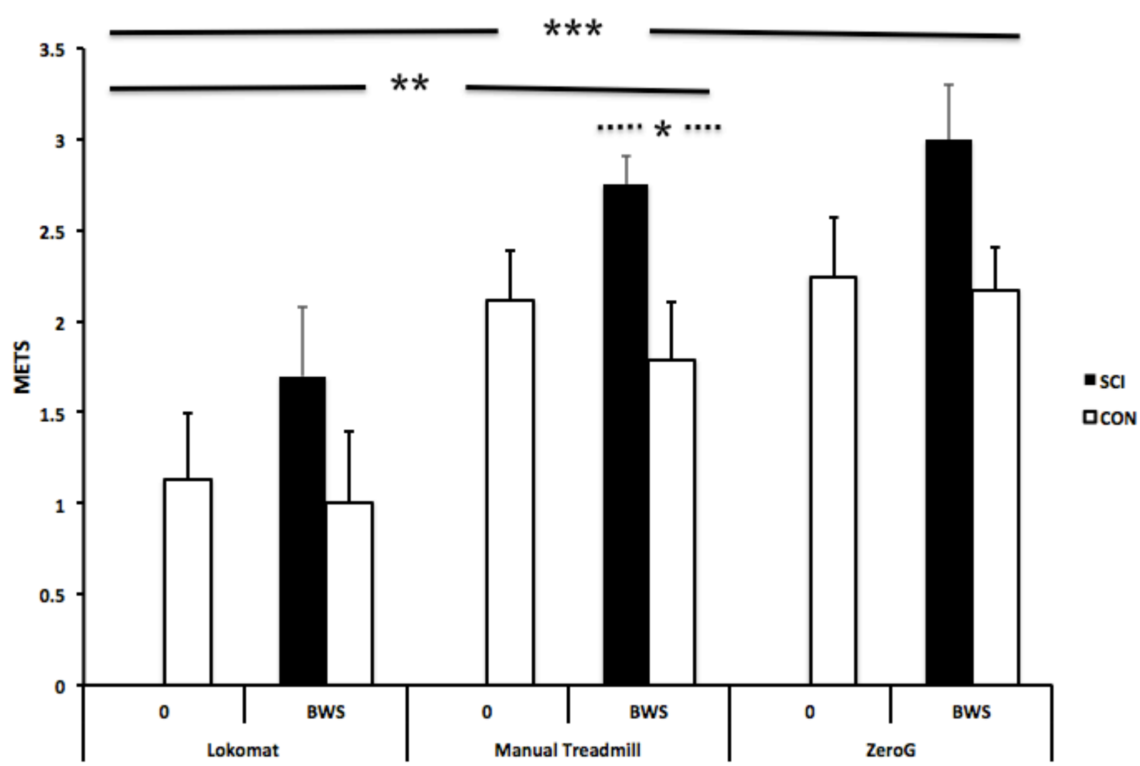
P < 0.001 = ***.

3.3.1 Aerobic Demand Expressed Relative to Resting Values

Lokomat^{TM} sessions resulted in significantly lower MET values when compared to the Manual Treadmill or ZeroG^{TM} sessions (Figure 6). The highest MET values were attained during ZeroG^{TM} walking, although these were not statistically different compared to the Manual Treadmill, for both SCI and controls (3.0 ± 0.30 vs. 2.8 ± 0.16, 2.2 ± 0.24 vs. 1.8 ± 0.32 respectively). Individuals with SCI achieved a significantly higher MET value compared to CON during the Manual

Treadmill session (2.8 ± 0.16 vs. 1.8 ± 0.32). There was no statistically significant effect of BWS observed for CON during any of the walking sessions.

Figure 6: METS during locomotion for SCI and CON with (BWS) and without (0) body weight offset.



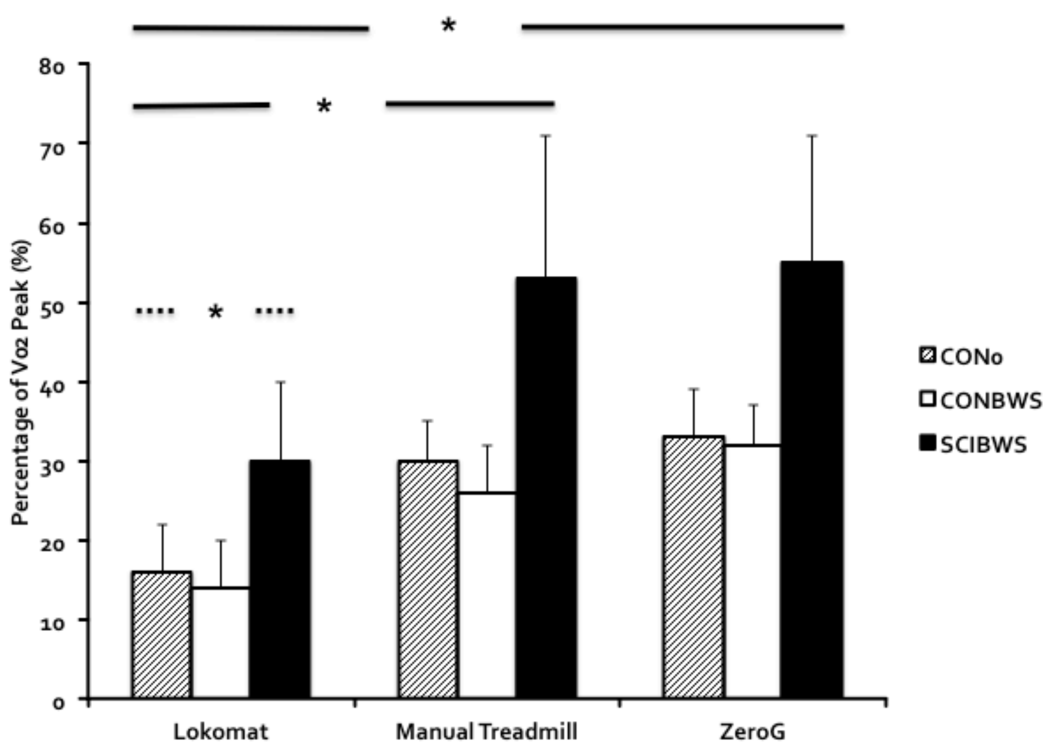
Solid line = significant between *device* comparisons, and dashed line = significant between *group* comparisons. P < 0.05 = *, P < 0.01 = **, P < 0.001 = ***.

3.3.2 Aerobic Demand Expressed Relative to Peak Values

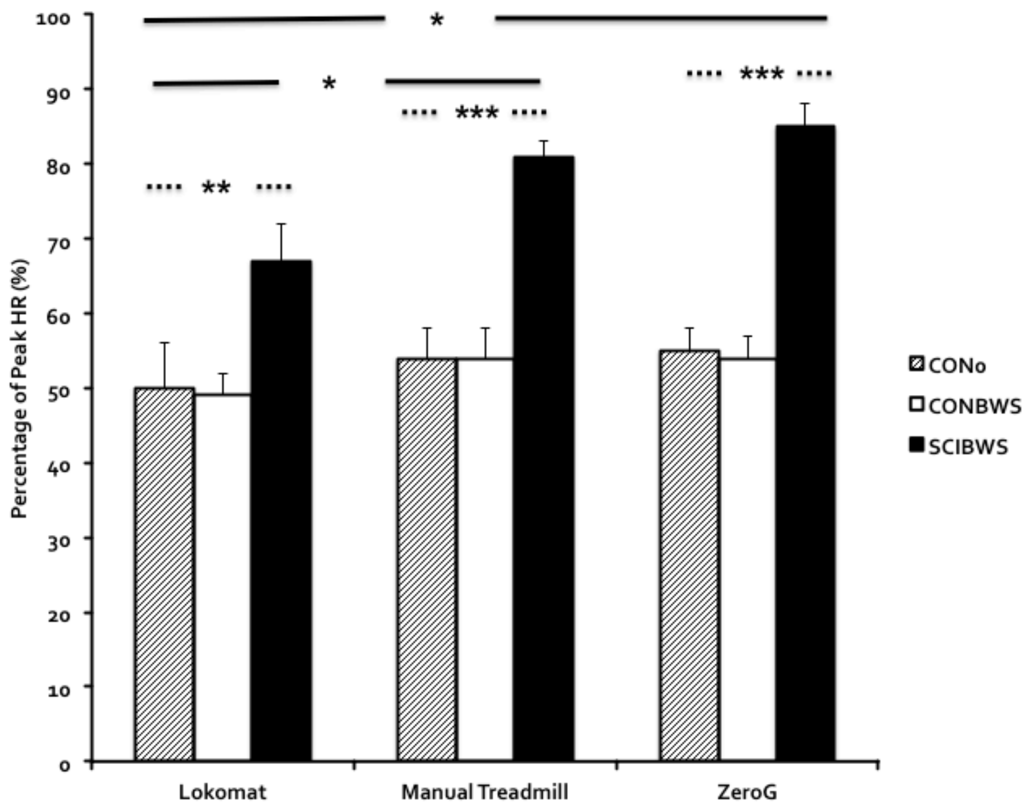
Cardiovascular measures were expressed relative to percentage of peak values obtained during the arm ergometer test (Figure 7a-b, Table 2).

Figure 7: Cardiovascular measures during locomotion expressed as percentage of peak values obtained from the arm ergometer test.

a) VO_2



b) HR



CON0 = control unsupported walking; *CONBWS* = control walking at matched BWS; *SCIBWS* = SCI supported walking. Solid line = significant between *device* comparisons, and dashed line = significant between *group* comparisons.

P < 0.05 = *, P < 0.01 = ** and P < 0.001 = ***.

Table 2: Absolute central (RPEC) and peripheral (RPEP) RPE ratings during locomotion.

	Lokomat™		Manual Treadmill		ZeroG™	
	RPEC	**RPEP**	**RPEC**	**RPEP**	**RPEC**	**RPEP**
CON0	1.0 ± 0.20	0.9 ± 0.14	0.9 ± 0.14	0.9 ± 0.14	1.0 ± 0.00	1.0 ± 0.00
CONBWS	0.9 ± 0.14	0.9 ± 0.14	0.9 ± 0.14	0.9 ± 0.86[d]	1.0 ± 0.00	1.0 ± 0.00[d]
SCIBWS	0.5 ± 0.19[ab]	0.7 ± 0.18[ab]	3.7 ± 1.3[a]	4.1 ± 0.64[acd]	3.8 ± 1.14[b]	5.1 ± 1.18[bcd]

CON0 = control unsupported walking; *CONBWS* = control walking at matched BWS; *SCIBWS* = SCI supported walking. Mean ± standard error. Significant ($p < 0.05$) differences relative to peak values between *devices*; Lokomat™ vs. Manual Treadmill = [a], Lokomat™ vs. ZeroG™ = [b], Manual Treadmill vs. ZeroG™ = [c]. Significant ($p < 0.05$) between *groups* differences = [d].

VO_2

The Lokomat™ session was significantly less demanding (expressed as a percentage of peak) compared to the Manual Treadmill and ZeroG™ sessions (e.g. 30.1% ± 10.06% vs. 52.9% ± 17.60% vs. 54.7% ± 16.42% for participants with SCI). Differences between groups existed during the Lokomat™ session only, with participants with SCI requiring a significantly greater percentage of peak VO_2 compared to CON (30.1% ± 10.06% vs. 14.4% ± 6.32%). For CON, the Lokomat^{TM} session was also found to be the least demanding session compared to the Manual Treadmill and ZeroG^{TM} with no statistically significant effect between BWS conditions (e.g. 16.0% ± 6.09% vs. 14.4% ± 6.32% for Lokomat^{TM} session).

HR

Lokomat^{TM} sessions resulted in significantly lower HRs (expressed as a percentage of peak) in comparison to the Manual Treadmill or ZeroG^{TM} sessions (e.g. 67.3% ± 4.77% vs. 80.8% ± 1.95% vs. 84.7% ± 3.03% for participants with SCI). Individuals with SCI achieved a significantly higher percentage of peak HR values compared to CON during all 3 walking sessions (e.g. 84.7% ± 3.03% vs. 54.2% ± 3.20% during the ZeroG^{TM} walking session). For CON, no statistically significant differences existed between devices or BWS conditions (e.g. 49.8% ± 3.57% vs. 48.8% ± 3.25% for Lokomat^{TM} session) for HR.

Central RPE

Lokomat^{TM} sessions were perceived by participants to be significantly less demanding on the heart and lungs when compared to the Manual Treadmill or ZeroG^{TM} sessions (e.g. 0.5 ± 0.19 vs. 3.7 ± 1.28 vs. 3.8 ± 1.14 for participants with SCI). Individuals with SCI perceived the Manual Treadmill and ZeroG^{TM} sessions to be significantly more demanding compared to CON (3.7 ± 1.28 vs. 0.9 ± 0.14; 3.76 ± 1.14 vs. 1.0 ± 0.00 respectively). For CON, non-statistically significant differences were found between devices and BWS conditions (e.g. 1.0 ± 0.20 vs. 0.9 ± 0.14 for Lokomat^{TM} session) with respect to perceived central effort.

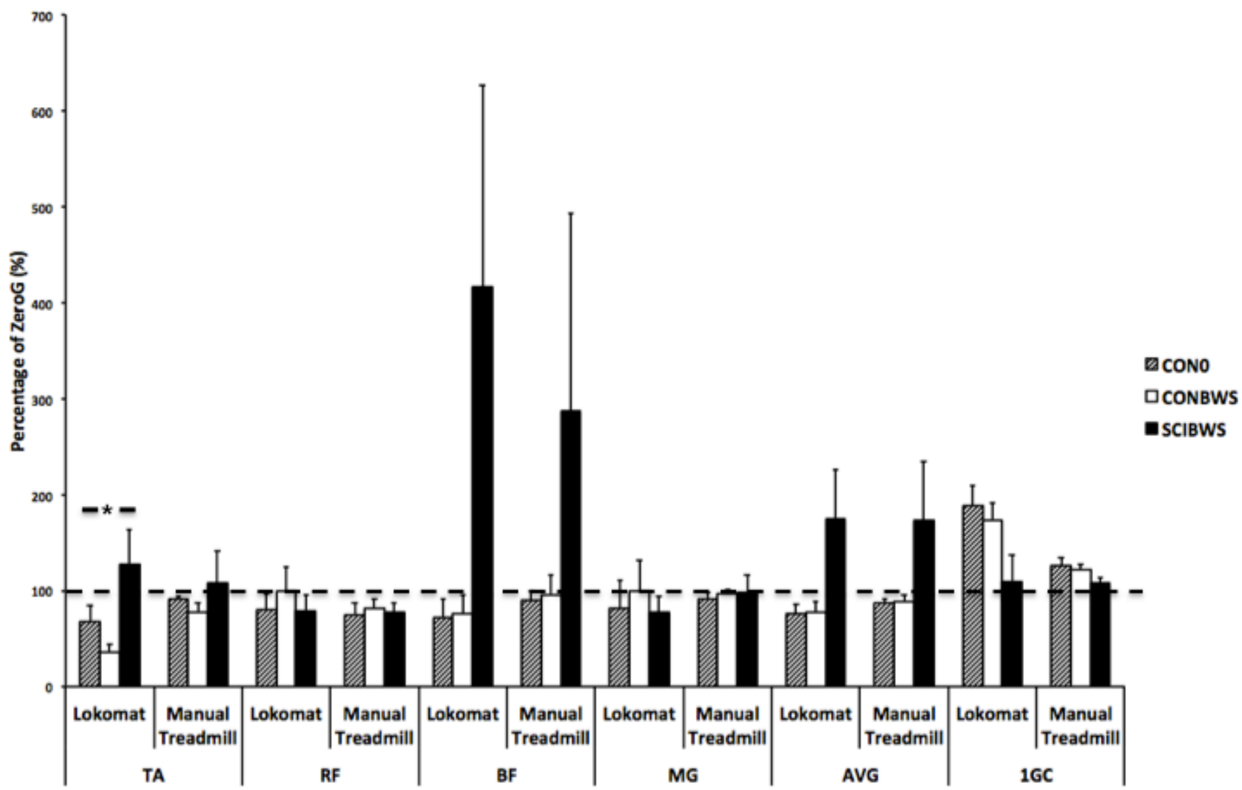
Peripheral RPE

Lokomat^{TM} sessions were perceived by participants to be significantly less demanding on the lower limbs when compared to the Manual Treadmill or ZeroG^{TM} sessions (e.g. 0.7 ± 0.18 vs. 4.1 ± 0.64 vs. 5.1 ± 1.18) for participants with SCI). Individuals with SCI perceived the Manual Treadmill and ZeroG^{TM} sessions to be significantly more demanding compared to CON (4.1 ± 0.64 vs. 0.9 ± 0.86, 5.1 ± 1.18 vs. 1.0 ± 0.00 respectively). For CON, non-statistically significant differences were seen between devices or BWS conditions (e.g. 0.9 ± 0.14 vs. 0.9 ± 0.14 for Lokomat^{TM} session) with respect to perceived peripheral effort.

3.3.3 Muscle Activity Expressed Relative to Walking While Using the ZeroG^{TM}

Non-statistically significant differences in muscle activity gait parameters were found between the three devices (Figure 8). For individuals with SCI, average muscle activation tended to be higher for both treadmill conditions compared to the ZeroG^{TM} session, which could be attributed to increases in TA and BF activity. Conversely, the ZeroG^{TM} session tended to require greater muscle activation compared to the treadmill sessions for CON. The only statistically significant difference between groups occurred during the Lokomat^{TM} session, which elicited greater relative TA activation for participants with SCI compared to CON (128.3% ± 35.07% vs. 36.0% ± 8.22%). For CON, significant differences between devices existed with respect to TA activation, which was higher during supported walking using the Manual Treadmill compared to Lokomat^{TM} (77.7% ± 9.03% vs. 36.0% ± 8.22%). No statistical differences were evident, however, between BWS conditions during any of the walking sessions in CON.

Figure 8: Muscle activity of independent muscle groups during treadmill locomotion expressed relative to overground walking using the ZeroG^{TM}.



100 = value obtained while walking on the *ZeroG^{TM}*. TA = tibialis anterior, RF = rectus femoris, BF = biceps femoris, MG = medial gastrocnemius, AVG = average muscle activity over a gait cycle and 1GC = gait cycle completion time. *CON0* = control unsupported walking; *CONBWS* = control walking at matched BWS; *SCIBWS* = SCI supported walking. Solid line = significant between *group* comparisons. P < 0.05 = *.

3.3.4 Muscle Activity Versus Oxygen Demand Across Devices

Non-significant statistical differences existed in the ratio of root mean square (RMS) activity (μV) of individual muscle groups to aerobic demand (%VO_2 peak) during supported walking for individuals with SCI (Table 3). The only statistical difference between groups occurred during the Lokomat™ session, which had a higher ratio for MG in CON compared to SCI (24.3 ± 20.97 vs. 1.4 ± 0.29). In the CON group, higher ratios existed for RF and MG during supported walking on the Lokomat™ compared to the Manual Treadmill and ZeroG™ (e.g. MG ratio: 24.3 ± 20.97, 1.6 ± 0.36, 1.2 ± 0.23).

Table 3: Ratio of root mean square (RMS) muscle activity (μV) to oxygen uptake (%VO_2 peak) across independent muscle groups for SCI and CON.

	Lokomat™		Manual Treadmill		ZeroG™	
	SCI	CON	SCI	CON	SCI	CON
TA	2.2 ± 0.71	3.1 ± 1.75	1.5 ± 0.49	1.4 ± 0.38	2.5 ± 1.63	1.4 ± 0.39
RF	0.7 ± 0.14	2.0 ± 0.69[abd]	0.7 ± 0.28	0.6 ± 0.14[a]	0.9 ± 0.42	0.6 ± 0.19[bd]
BF	15.9 ± 9.41	1.7 ± 0.52	3.3 ± 2.84	1.1 ± 0.24	1.2 ± 0.53	1.2 ± 0.48
MG	1.4 ± 0.29[c]	24.3 ± 20.97[abcd]	2.9 ± 1.55	1.6 ± 0.36[a]	2.9 ± 2.06	1.2 ± 0.23[bd]

TA = tibialis anterior, RF = rectus femoris, BF = biceps femoris, MG = medial gastrocnemius. Mean ± standard error. Significant ($p < 0.05$) between *device* differences; Lokomat^{TM} vs. Manual Treadmill = [a] and Lokomat^{TM} vs. ZeroG^{TM} = [b]. Significant ($p < 0.05$) between *group* differences = [c]. Significant ($p < 0.05$) differences between *BWS* conditions = [d].

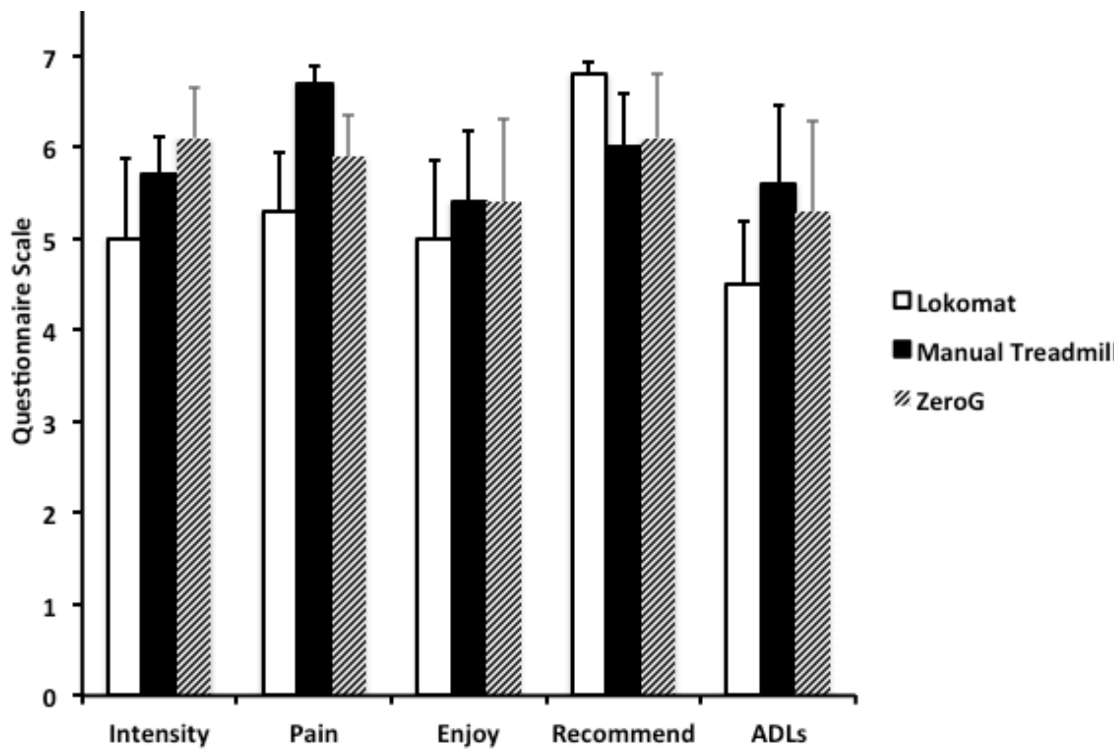
3.3.5 Consumer Preference

Information regarding the subjective experience of participants with SCI after the different walking sessions were collected from all 7 participants for the Manual Treadmill and ZeroG^{TM} sessions and 6 participants for the Lokomat^{TM} session. Only the data from the 6 participants with SCI who experienced *all* three devices were included in the final analysis.

Overall, participants seemed to equally enjoy completing all 3 walking sessions (Figure 9). Following individual session completion, participants were unanimous in strongly agreeing that exercise facilities should purchase the Lokomat^{TM}. Participants found the Manual Treadmill to be the least painful (e.g. most comfortable) session to complete and believed this would be most beneficial in developing the fitness and skills needed to complete activities of daily living (ADLs). Finally with respect to appropriate level of intensity of the session (e.g. BWS, treadmill speed), these individuals found the Manual Treadmill and

ZeroG^{TM} devices to be more appropriate for their ability compared to the Lokomat^{TM}.

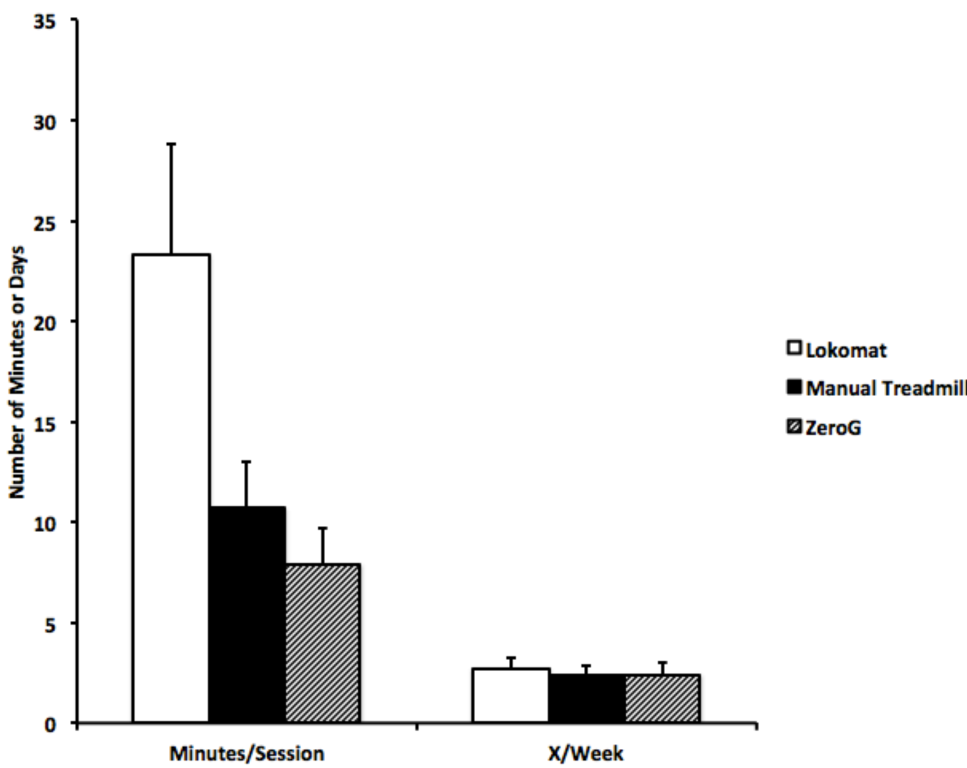
Figure 9: Subjective evaluation of walking experience.



The following aspects were evaluated on a scale from 1 to 7 (strongly disagree to strongly agree): a) appropriateness of session *intensity* (e.g. support, speed), b) session completion without *pain*, c) *enjoyment*, d) *recommendation* for exercise facility investment in equipment and d) appropriateness of session to improve ability to complete *activities of daily living* (ADLs).

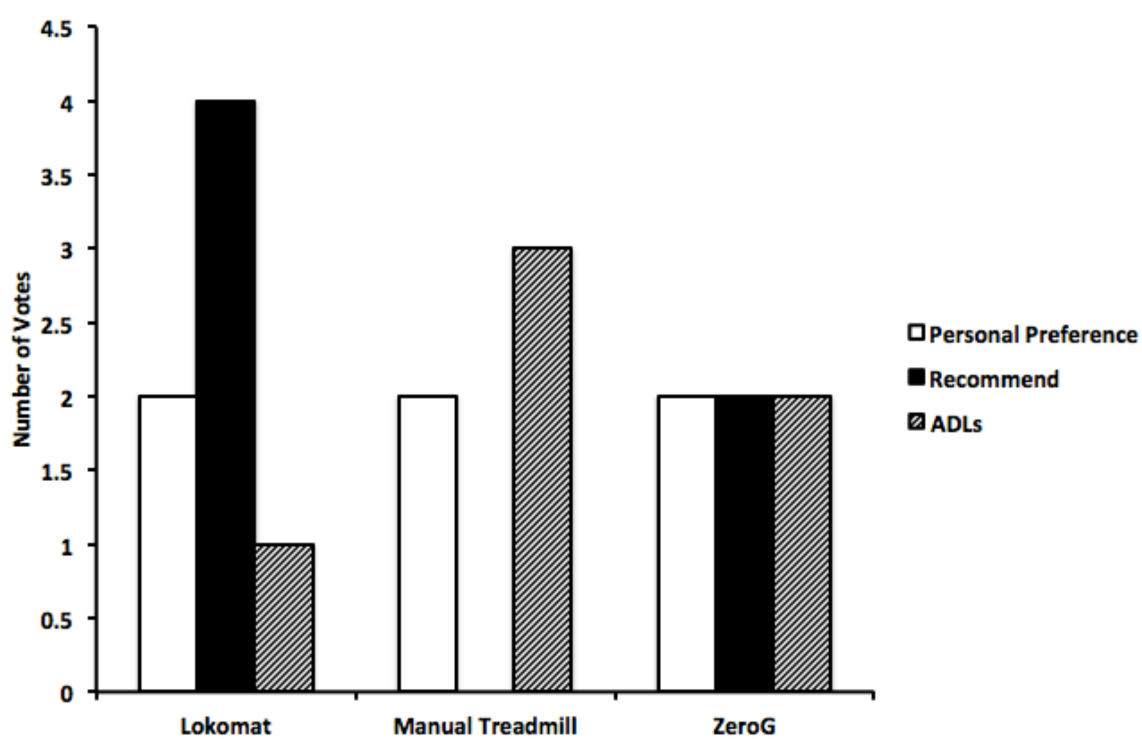
In general, participants indicated they would be interested in using these body weight supported devices 2-3 times per week (Figure 10). Participants believed they could complete longer sessions using the Lokomat^{TM} compared to the Manual Treadmill or ZeroG^{TM} devices (23.3 ± 6.54 vs. 10.7 ± 2.30 vs. 7.9 ± 1.84 min respectively).

Figure 10: SCI determined exercise parameters (length, times per week) based on subjective evaluation of intensity of walking experience.



Consistent with the individual session questionnaires, overall, participants equally enjoyed the 3 different walking sessions (Figure 11). Based on their experience, participants would recommend exercise facilities invest in the Lokomat^{TM} over the other two devices because of its dual-purpose ability, as it can be used with and without the robotic limbs. Additional benefits mentioned included the following: the robotic treadmill allows for feedback using the computer system, it allows individuals to go through a greater range of motion (ROM) during gait and provides greater freedom for therapists allowing for more individualized and focused sessions. Finally, in terms of a perceived applicability to the performance of ADLs, participants showed slight preference towards the Manual Treadmill and ZeroG^{TM} compared to the Lokomat^{TM}.

Figure 11: SCI final device selection.



Number of votes (maximum = 6) indicating 1) most *preferred* device, 2) which device they would *recommend* exercise facilities to invest in, and 3) which device session would best assist individuals complete *activities of daily living* (ADLs).

CHAPTER 4: DISCUSSION

Based on the primary objectives of the presented thesis it was hypothesized that the highest levels of oxygen uptake and muscle activation would occur during locomotion using the ZeroG^{TM} and this would be higher in persons with incomplete SCI versus gender and age matched able-bodied adults. Additionally it was anticipated that individuals would prefer the ZeroG^{TM} device because sessions using this device most closely resemble unsupported overground walking. As expected walking sessions were physiologically more demanding for individuals with SCI compared to CON. Contrary to what was hypothesized, both the Manual Treadmill and ZeroG^{TM} sessions were considered more demanding and were the preferred training modalities compared to the Lokomat^{TM}.

4.1 DEMOGRAPHICS AND BASELINE TESTING

Demographic variables of the SCI group were comparable to the CON group. VO_2 peak values obtained from the men with incomplete SCI (1.7 ± 0.27 L/min) were compared to normative physical capacity values; the 5 participants with paraplegia (1.7 ± 0.38 L/min) and 2 participants with tetraplegia (1.7 ± 0.18 L/min) performed at levels that were considered fair (1.34-1.72 L/min) and excellent (≥1.19 L/min) respectively (Janssen et al, 2002). Peak VO_2 values from able-bodied participants (2.2 ± 0.15 L/min) were similar to those reported by van Loan and colleagues (1987) (2.1 L/min) and more recent unpublished data from our lab (2.4 L/min). Interestingly, the peak aerobic data in this study was similar between the two groups, which are contrary to other studies indicating those with SCI usually obtain lower values (Cerny, 1980) due to a decreased amount of active muscle mass and sympathetic tone limiting venous "muscle pumping" action and the ability to increase oxygen uptake (Phillips, 2008). Zwiren and Bar-Or (1975), however, found no significant differences in VO_2 max in matched wheelchair-active and normal active subjects, suggesting conditioning levels of the participants in the current study may have been more similar than typically expected (e.g. two participants with SCI were competitive athletes). In addition, the majority of participants with SCI had injuries in the mid-thoracic and lumbar regions, indicating a greater amount of active muscle mass would have been available for them to utilize to achieve higher peak values compared to a sample consisting of primarily higher lesion levels.

Wirth and colleagues (2008) have emphasized the need for a dynamic assessment tool to detect and follow motor deficits following an incomplete SCI. This was in response to evidence that electrically stimulated muscle contractions are similar between SCI and CON, however rate of torque development is drastically reduced in SCI. To this author's knowledge this is the first study to use the Lokomat™'s L-Force module to assess lower extremity isometric strength of the hip and knee in individuals with incomplete SCI. The reduced hip extension and knee flexion strength measures obtained from participants with SCI compared to CON is in agreement with previous studies which suggest this population has difficulty voluntarily activating muscle below the lesion level making weight bearing and toe-clearance during gait difficult (Ditunno and Scivoletto, 2009). Further investigation into the flexion:extension ratio at both the hip and knee determined that a positive correlation existed between flexion:extension at the hip (primarily due to a reduction in hip extensor isometric strength) and the amount of BWS required to complete an overground walking session using the ZeroG™. The importance of the hip extensors is in agreement with a study by Yang and colleagues (2011), who found that manual muscle testing scores for the hamstrings in addition to the quadriceps were the strongest predictors of ***responsiveness*** to body weight supported gait training. In fact, responders on average had twice the volitional muscle strength as that of nonresponders. Whether the isometric strength measures obtained from this study are able to predict responsiveness to body weight supported gait training, however, requires further investigation.

4.2 RANDOMIZED BODY WEIGHT SUPPORTED WALKING SESSIONS

4.2.1 Body Weight Support

The average amount of BWS used during the walking sessions was 41.3%, with only three of the seven participants with SCI able to complete walking sessions with the recommended less than 30% BWS. These lower levels of support have been shown to better resemble independent overground walking patterns while allowing individuals to better maintain upright posture. For the CON group that tested in unsupported and matched conditions it was expected that there would be a reduction in oxygen cost during supported compared to unsupported walking sessions which could have been attributed to a) energy expenditure being related to body weight and b) reduced mechanical work required to vertically lift the body during gait. Although there was a trend for a decrease in all physiological measures with BWS this did not reach statistical significance for any walking session in the CON group. This is in contrast to studies in clinical populations (e.g. stroke, paraplegia) which found reduced oxygen cost and heart rate during body weight supported walking sessions when as little as 30% BWS was provided (Danielsson et al, 2000). With respect to leg muscle activation, differences in RF and MG ratios of muscle activity to aerobic demand were evident between BWS conditions when using the Lokomat^{TM} and ZeroG^{TM}. Differences between unsupported and matched conditions were primarily due to alterations in muscle activity for similar levels of %VO_2; this will be further addressed in subsequent sections.

4.2.2 Comfortable Walking Speed

Average speed for subjects with incomplete SCI in this study was less than half that for non-injured subjects when the robotic orthosis was not used (e.g. 0.61 vs.

2.1 km/h during Manual Treadmill sessions). Another study by Waters and Lunsford (1985) compared walking speeds of individuals with paraplegia to non-injured adults and found 67% slower speeds in the injured group, which is similar to the 65.6% difference found in the present study. Collectively, these findings are in agreement with Hidler and colleagues (2011) who suggest gait speeds typical of individuals who use the ZeroG^{TM} are well below that of community ambulators, which is approximately 0.7 m/s. Training at faster speeds has been shown to increase afferent and efferent activity during walking and has been shown to be more effective at increasing overground walking speed in stroke patients (Domingo et al, 2007). It has been suggested that, compared to able-bodies individuals, those with SCI have limited abilities to adapt faster natural cadence due to a limited ability to increase stride length and an inability to increase stride frequency (Pépin et al, 2003). Domingo and colleagues (2007) suggest that by providing manual assistance during treadmill walking (which was not done in the present study) may help individuals safely train at faster speeds while keeping muscle activation patterns similar to CON. From a therapeutic perspective these are all aspects of the training program, which must be considered.

4.2.3 Metabolic Demand

Previous research has indicated the robotic orthosis results in lower metabolic costs (approximately 20%) compared to the manual treadmill (Hornby et al, 2012), with the potential to minimize these differences if the participant is encouraged to exert maximal effort (Israel et al, 2006). In this study, despite encouraging participants to maximally contribute to the walking motion in all walking conditions differences in metabolic costs were evident, with the Lokomat^{TM} resulting in the lowest metabolic demand (approximately 23.8% of VO_2 peak for the SCI group) compared to the Manual Treadmill and ZeroG^{TM} sessions, which had similar oxygen costs. While this may suggest that perhaps participants were not providing maximal efforts during these sessions, it is also important to note that in order to standardize walking conditions between participants 100% guidance force was set on the robotic orthosis which may have limited the ability for a maximal effort of participants. Additionally, the study by Israel and colleagues (2006) provided therapist assistance during treadmill sessions, an option that was not provided for participants in this study. Other features of the Lokomat^{TM} that could have contributed to the differences in metabolic costs include posterior support assisting with forward propulsion, and stability at the pelvis and trunk. The evidence from this study suggests the use of the robotic orthosis is not entirely passive, with increases in oxygen uptake evident during training sessions, which according to Krewer and colleagues

(2007) can be attributed to loading during stance phase resulting in associated muscle activation.

In this study walking sessions completed with the ZeroG^{TM} and Manual Treadmill sessions required greater than 50% of VO_2 peak values for the SCI group (e.g. 55% and 53% of VO_2 peak respectively). This would suggest that for these individuals walking using the aforementioned devices is above the ***anaerobic threshold*** resulting in reduced endurance and earlier onset of fatigue. In contrast, the ZeroG^{TM} and Manual Treadmill sessions required only 32% and 26% of peak oxygen uptake for the CON group. This is consistent with evidence obtained by Waters and colleagues (1983) who found individuals without mobility impairments require minimal effort during walking with rates of oxygen consumption of 30% of maximum aerobic capacity. The similarity in oxygen costs between the treadmill session without the robotic orthoses and the overground walking session is in agreement with evidence in able-bodied populations which suggest no significant differences exist in energy expenditure between treadmill and overground walking at controlled velocities (Waters and Mulroy, 1999). It is important to note that walking velocity in this study was not controlled, although all participants received the same instruction to walk at a comfortable speed. Rate of oxygen uptake while walking overground in this study was 41.8% higher in SCI compared to non-injured adults which is similar to earlier work which found the rate of oxygen uptake while walking to be 38% higher in SCI compared to

CON (Waters and Lunsford, 1985). In the present study individuals with incomplete SCI had 36.5% higher HR compared to CON during overground walking sessions which is slightly greater than the 24% greater increase found during walking by Teixeira da Cunha-Filho and colleagues (2003).

In order to complete a walking session using the overground device, at the same level of BWS, the participants with SCI required three times resting metabolic rate while their matched controls only required double. For all subjects this is equivalent to most customary activities of daily living including domestic work (e.g. bed making) (Jette et al, 1990; Collins et al, 2010). With respect to physical training, as Jette and colleagues (1990) suggest, this MET value may provide a sufficient training stimulus to develop cardiorespiratory fitness for individuals whose functional capacity is less than 6 METS, however the average in this study for the SCI group based on arm ergometry was 7 METS.

4.2.4 Muscle Activity

Muscle activity for SCI tended to be greater with treadmill than the ZeroG™ due to increases in TA and BF activity. For the SCI group, the “foot lifters” used with the Lokomat™ may have provided afferent feedback to the lower limb during gait encouraging dorsiflexion. Evidence from this study supports this as Lokomat™ sessions resulted in greater TA activation in SCI compared to CON. While the “foot lifters” used with the robotic orthosis may have provided beneficial feedback for the SCI group, reduced EMG activity of TA during gait in the CON group may have occurred using this same device as the foot lifters may have inhibited normal activation of this muscle group. It is interesting that, Domingo and colleagues (2007), showed higher EMG activities in TA and thigh muscles generally are higher in SCI compared to CON during treadmill walking, with activity increasing with increases in speed whereas for CON they remain relatively constant. Increased activity of these muscles groups could potentially be attributed to attempting to compensate for a lack of lower limb stability. Additionally, individuals with SCI tend to have more flexed knees during stance, requiring greater extensor activation in order to prevent collapse of the lower limb during gait.

It has been suggested that muscles with greater cortical projections, such as TA and more proximal muscles such as the hamstrings (Brower and Ashby, 1990) are the most affected in individuals with SCI following body weight supported training (Lucareli, 2001). Improved hip extension with training is common with treadmill training as the belt encourages hip extension forcing individuals to “pull up” (e.g. increasing knee flexion) during swing. The large variability in BF activation in this study may be an indicator of potential responders to body weight supported training. Gorassini and colleagues (2009) showed only responders increased TA and hamstring muscle activation during treadmill walking. While responders increased amplitude of hamstring activity, burst duration decreased resulting in less ***co-contraction*** with quadriceps activation. Thus, the ability to modify muscle activation patterns post SCI may predict responsiveness to training.

No differences in muscle activation existed between the two treadmill modalities, which is in agreement with previous work by Foreman and colleagues (2005) who found no difference in muscle activity patterns between robotic and therapist assisted walking in SCI. It is important to note that while manual assistance was not provided to individuals with SCI during treadmill sessions in the present study there is no evidence to suggest significant differences in muscle amplitude or activation patterns exist when conducting treadmill training with or without manual assistance (Domingo et al, 2007).

Unlike SCI, muscle activity for CON tended to be greater with overground sessions rather than treadmill sessions, as well as having higher TA activation during supported Manual Treadmill versus Lokomat^{TM} sessions. Muscular work associated with forward propulsion is thought to be a primary determinant of the metabolic costs of walking in subjects without neurological injury (Israel et al, 2006) which would be greater when using the ZeroG^{TM} compared to the treadmill sessions, especially when the robotic orthosis is used. As previously mentioned, “foot lifters” used with the robotic orthosis may have inhibited normal activation of

TA for the CON group, resulting in the obtained differences between the two treadmill modalities.

While both groups were able to successfully complete all walking sessions, greater muscle activity was evident in SCI during treadmill sessions compared to during ZeroG^{TM} sessions for the CON group. This may provide evidence in support of the idea of ***motor equivalence*** (Grasso, 2004). Essentially this principle suggests that a given motor task goal (e.g. walking) can be achieved using different muscle synergies. This is advantageous for individuals with lesions to the spinal cord as they can take advantage of the redundancies of the neuromuscular system to accomplish motor tasks. For example, individuals with SCI often will use their arms and/or axial muscles to assist with swing phase during gait. Coupling of movement provides sensory feedback to help generate temporally appropriate muscle activity patterns. From a therapeutic perspective body weight supported training interventions help individuals to learn to produce new motor strategies in a controlled setting with the hopes of transferring this rehearsed pattern to unsupported overground walking.

4.2.5 Muscle Activity Versus Oxygen Demand Across Devices

There were no differences between devices in the ratios of muscle activity to aerobic demand in the SCI group. On the other hand, for the CON group, the ratios that were most affected across devices were RF and MG. The differences were generally accounted for by changes in muscle activity for the same level of %VO_2. For RF, activity decreased with BWS when using the Lokomat^{TM} and ZeroG^{TM}. For MG, activity increased with BWS when using the Lokomat^{TM} and decreased when using the ZeroG^{TM}. Thus, for the same level of BWS, greater activity of MG could be obtained for a given %VO_2 in the CON group when using the Lokomat^{TM} compared to the ZeroG^{TM}. The susceptibility of reduced RF and MG activity to BWS in individuals without mobility impairments when using the ZeroG^{TM} is in agreement with a study by Fenuta and Hicks (2013), which found offloading body weight by as much as 80% BWS significantly decreased heel strike activity of RF and MG by 62.8% and 35.5% respectively.

4.2.6 Consumer Preference

Upon subjective evaluation of the walking experiences in this study, participants with SCI appeared to equally enjoy all three devices, would be interested in using the devices 2-3 times per week and showed slight preference for use of sessions without the robotic orthosis in order to better transfer learned skills to completion of activities of daily living [e.g. ZeroG^{TM} sessions required 3 METS which is the equivalent requirement for bed-making (Collins et al, 2010)]. The fact that participants equally enjoyed all three devices provides therapists with greater freedom in terms of device selection depending on the purpose of the session. In general, subjective experience of participants matched the physiological response of the walking sessions. The ZeroG^{TM} and Manual Treadmill sessions appeared to exceed the anaerobic threshold which would explain why these sessions in particular required more effort physiologically and in terms of perceived exertion compared to the Lokomat^{TM}, which required only 30% of peak oxygen uptake. The length of time an individual can perform prolonged exercise at work rates above the anaerobic threshold decreases inversely with the rate of energy expenditure due to the restricted availability of anaerobic energy supply and accumulation of lactate in the muscle. The current evidence supports these patterns as individuals with SCI upon subjective evaluation of the walking experiences anticipated they would be able to complete longer sessions using the Lokomat^{TM}, compared to the Manual Treadmill and finally the ZeroG^{TM}. These are all aspects of program design that need to be considered in order to provide an intervention that is beneficial to and can be tolerated by the individual.

4.3 STRENGTHS AND LIMITATIONS

The aim of this study was to compare the metabolic and muscle activation responses to currently available locomotor training devices. A strength of the study was its repeated measures design, which allowed all participants to act as their own controls. Participants with incomplete SCI were eligible to participate in the study only if they were at least 1-year post injury (average 4 years) to control for ***spontaneous recovery*** potentially influencing the obtained results. In addition, the incorporation of a gender- and age-matched control group further strengthens this study, providing comparative information between injured and non-injured individuals on the responses to bodyweight-supported ambulation. The *simultaneous* use of the portable metabolic cart and wireless EMG system allowed for an evaluation of the relationship between muscle activation and oxygen costs of walking sessions, particularly with the ZeroG^{TM} which is a new overground training device used for gait rehabilitation. Normalizing to peak aerobic measures and muscle activity during ZeroG^{TM} sessions at every testing session allowed for comparison between individuals as well as controlling for potential day-to-day variation in EMG signals. Manual treadmill sessions were conducted without therapist assistance to see how individuals would naturally respond to body weight support *without* external assistance, which is a novel approach. This is also the first study of its kind to report isometric strength data using the L-Force software module in the Lokomat^{TM} system to evaluate its potential usefulness as a tool for therapists to use to track progress and/or determine readiness for gait training. To this author's knowledge, this is also the first study to acquire information from consumers regarding preference of devices used during gait rehabilitation, which was conducted to hopefully provide insight into which interventions provide ***efficacious*** and ***effective*** treatments to this population.

A few limitations of this study must be addressed. The small sample size, consisting of entirely males with incomplete SCI may limit the generalizability to the entire SCI population. In terms of methodology, it is important to consider that the Lokomat^{TM} was programmed to provide maximal assistance to the legs during all phases of gait in order to standardize the guidance force, therefore these results are only relevant to Lokomat^{TM} sessions completed under these conditions. Additionally, the participants were allowed to use the treadmill handrails or the arms of the therapist during the overground session for balance but were discouraged to use them for weightbearing. While attempts were made to ensure consistent upper-extremity use across stepping conditions no objective measures of force (e.g. force plates on handrails) were made to ensure of this. Additionally, without the use of arm swing individuals were unable to take advantage of ***neuronal coupling***, which may have influenced lower extremity muscle activity as suggested by Ferris and colleagues (2006). While the portable metabolic system used in this study allowed for an evaluation of the overground training system, there is limited information regarding its accuracy and reliability. Some studies have indicated systematic biases in VO_2 values of portable systems compared with standard metabolic carts. Crouter and colleagues (2006) compared the VO2000, TrueOne2400 portable metabolic systems with the Douglas bag method during cycling stages of increasing intensity in healthy males. Their results found the VO2000 portable metabolic system to be less reliable for measuring VO_2 and VCO_2 compared to the other two methods. It tended to underestimate VO_2 at rest (53%) and overestimate VO_2 during cycling (8%) at most cycling work rates. While additional research is needed to confirm these results particularly during walking, these small systematic biases should not have affected the validity of the values obtained in this study (since the device was used in each modality).

The variability of muscle activity during the different walking sessions, particularly for the participants with SCI, may have contributed to why no significant differences were found between devices. The inability to control for walking speed, particularly with respect to the overground walking session may have influenced the results obtained, as reduced speed has been associated with increased signal variability and decreased muscle activation. According to Winter (1983), changes in walking speed affect the acceleration of lower limbs during gait, primarily activity at the hip and knee versus the ankle. Additionally this study looked at average muscle activity across the *entire* gait cycle and did not break down the activity into phases (e.g. swing vs. stance phase), which may potentially be where differences between devices exist.

4.4 FUTURE CONSIDERATIONS

Recently there has been an interest in using functional electrical stimulation in combination with these body weight supported devices in those who can tolerate the stimulus to improve temporal activation of paralyzed muscle during gait training. Repeating this study with a larger sample size using these devices in combination with functional electrical stimulation will provide additional information regarding the acute benefits of the simultaneous use of these modalities and if greater differences will become apparent (particularly between the Manual Treadmill and ZeroG^{TM}) with this external stimulus assisting with the gait pattern. Follow up training studies with these devices used with or without functional electrical stimulation also need to be conducted. Although controversial, there is some experimental evidence to suggest a gender bias in favour of females in terms of functional recovery following an SCI (Chan et al, 2013) and thus inclusion of females is important in order to generalize the findings of the present study. Inclusion of additional muscle groups (e.g. postural muscles) and controlled velocities will help to provide a more complete understanding of the unique features of therapeutic sessions using the three different devices. All of these factors are important to consider in determining which modalities best result in improvements in endurance, muscle strength, functional capacity and/or in-home/community ambulation.

Whether the isometric strength measures obtained from this study are able to predict responsiveness to body weight supported gait training requires further investigation. Use of this measurement in studies at baseline and post-training with each of these devices will help to determine whether changes in isometric muscle strength as measured objectively by the L-Force module can be used as a predictive measure of improvement following gait training. If so, this may provide a valuable tool for therapists to use to determine readiness for body weight supported gait training and monitor progression. It would be worthy to note that although strengthening of muscles may contribute to better walking, it cannot replace the need to relearn muscle activation patterns in a task specific manner.

4.5 CONCLUSIONS

The main objective of this study was to compare oxygen demand, muscle activation and consumer preference between 3 body weight supported devices (Lokomat™, Manual Treadmill and ZeroG™). Consistent with our hypothesis walking sessions were more demanding for participants with SCI compared to CON. Contrary to our hypothesis the ZeroG™ was not considered the most physiologically demanding session, with both the ZeroG™ and Manual Treadmill sessions eliciting significantly greater VO_2 values compared to the Lokomat™. When expressed relative to resting values, overground sessions required 3.0 METS for participants with SCI, compared to 2.8 and 1.7 METS for the Manual Treadmill and Lokomat™ sessions respectively. In addition, contrary to what was expected, the ZeroG™ did not elicit significantly greater lower limb muscle activity in the four muscle groups included in this investigation and was not the selectively preferred modality by consumers. In general, consumers believed both the ZeroG™ and Manual Treadmill to be more appropriate for their ability and most beneficial in transferring the skills obtained towards the achievement of unsupported overground walking.

The observed positive relationship between the ratio of hip flexion:extension isometric strength and the amount of BWS required to complete a walking session using the ZeroG™ was interesting. Further investigation is required to see if this variable can be used as a predictor of individual support required for body weight supported walking in individuals with incomplete SCI. This may provide a useful tool to therapists in terms of exercise prescription, determining readiness for locomotor training, as well as monitoring rehabilitation progression. The evidence from this study would suggest using the Lokomat™ to work on isolated hip extension strength. Therapists can take advantage of the feedback system of the device, the greater BF activation of this treadmill-based exercise, as well as having the ability to conduct longer sessions due to the decreased cardiovascular and muscular demands imposed on the patient using the robotic device. The Manual Treadmill and ZeroG™ should be used as more intense progressions where hip extension can continue to be encouraged while using the treadmill and additional components of gait (e.g. balance and torso stability) can be focused on while using the overground device.

CHAPTER 5: MUSCLE ACTIVATION DURING BODY WEIGHT SUPPORTED LOCOMOTION WHILE USING THE ZEROG^{TM}

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Muscle Activation During Body Weight Supported Locomotion While Using the ZeroG^{TM}

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ABSTRACT

The ZeroG™ provides dynamic body weight support (BWS) using a harness while individuals with mobility impairments (e.g. spinal cord injury) ambulate overground. Muscle activity during locomotion using this device was studied in thirteen able-bodied adults (23.8 ± 2.71 yrs). Electromyography (EMG) recordings were collected from tibialis anterior (TA), medial gastrocnemius (MG), rectus femoris (RF) and biceps femoris (BF) muscles during randomized walking trials at preferred speeds under 5 levels of BWS (0,20,40,60,80%). Filtered EMG signals from each trial were normalized to 0% BWS and correlated to gait phases. Muscle activity, averaged across muscles, decreased significantly at heel strike by 33.4% with increasing BWS. Offloading significantly decreased heel strike activity of RF (62.8%), MG (35.5%) and TA (25.9%) respectively. Gait cycle completion time increased with BWS primarily due to increased swing phase time. These results summarizing the effect of BWS on muscle activation during ambulation can now be compared with clinical populations using the ZeroG™.

KEY WORDS:

Dynamic body weight support (BWS), electromyography (EMG), gait, mobility impairments, muscle activity, rehabilitation, ZeroG^{TM}

ABBREVIATIONS:

Biceps femoris (BF), body weight support (BWS), electromyography (EMG), medial gastrocnemius (MG), rectus femoris (RF), root mean square (RMS), tibialis anterior (TA).

INTRODUCTION

Body weight supported locomotor training has created enthusiasm in clinical populations and service providers as a rehabilitation tool to improve overground walking ability and other health outcomes [1]. Varying the amount of body weight supported by such devices alters the intensity of training sessions; unloading the individual decreases the muscular demands of the lower limbs [2]. For individuals with neurological impairments (e.g. spinal cord injury) there is evidence to suggest that by offsetting body weight, appropriate gait pattern expression may be facilitated and even normalized. Adult spinal cat models provided initial indications of the possibility of locomotor recovery following interactive treadmill gait training programs, which provided graded weight support [3]. Persons with spinal cord injury (complete and incomplete) have demonstrated the ability to increase weight bearing capacity of lower limb extensors through body weight supported treadmill training, in addition to increasing appropriate gastrocnemius and decreasing inappropriate tibialis anterior activation [4]. An overground body weight supported walking study (0/30/100% BWS at natural cadence) in individuals with chronic stroke found offloading resulted in reduced hip range of motion, with individuals responding to the support by decreasing their walking speed and taking shorter stride lengths [5].

As the above information suggests, there are a variety of therapeutic training approaches that offset body weight as a patient progresses through their rehabilitation program. These options include manual or robotic assisted treadmill training, as well as treadmill or overground training with or without electrical stimulation of the peroneal nerve [6]. Each training approach has its respective advantages and disadvantages; for example, robotic assisted training provides stepping consistency, but does not allow the movement variability associated with overground walking [7]. In addition this therapeutic option is treadmill based and as a result, is not as task-specific or transferable to daily living as overground training programs [5].

Currently there is a need for a body weight supported overground walking device equipped with a motor which moves along an overhead track independent of the user, in order to eliminate the possibility that a drag force created by the system itself is affecting gait performance [5]. The newly developed ZeroG^{TM} system (Aretech, LLC, Ashburn, VA) created by Hidler and colleagues, attempts to provide this dynamic body weight support (BWS) through the use of a custom-series elastic actuator during overground walking and balance training (Fig. 1) [8]. Through the use of dynamic BWS, it has been suggested that more natural ground-reaction forces and gait characteristics can be achieved [8]. To date, no study has investigated the effects of BWS on muscle activation patterns during walking using the ZeroG^{TM}. It would be beneficial to know "normal" responses to partial weight bearing using this device before applying its use in a rehabilitation setting. Therefore, the purpose of this study was to collect normative data from walking trials in able-bodied participants using the ZeroG^{TM} at varying degrees of BWS (0/20/40/60/80% BWS). It was hypothesized that decreased levels of activation in primarily the ankle plantarflexors (e.g. calves) would be experienced with increased BWS. The results from this study will eventually be used as a model for comparison with individuals that have mobility impairments who use this rehabilitation tool for gait therapy.

Figure 1: ZeroG^{TM} overground BWS system. A custom series elastic actuator travels along an overhead trolley providing static and dynamic BWS allowing individuals to safely practice activities of daily living.

METHODS

Subjects

Thirteen “able-bodied” individuals (6 males, 7 females) participated in the study with an average age of 23.8 ± 2.71 years and weight of 68.3 ± 7.98 kilograms.

“Able-bodied” was defined as an individual without musculoskeletal disease (e.g. multiple sclerosis, cerebral palsy) or injury within the past year (e.g. sprain, strain) of the lower leg and who was able to comfortably walk 30 m unassisted.

Protocol

Subjects were informed prior to testing to wear comfortable clothing including a t-shirt, shorts, socks and running shoes. All testing took place at the Robert Fitzhenry Specialized Rehabilitation and Exercise Lab in the Ivor Wynne Centre at McMaster University (Hamilton, ON) in accordance to the approved protocol by the McMaster Research Ethics Board.

Delsys Trigno Wireless Electromyography (EMG) Surface Electrodes (Delsys Inc, Boston, MA) were attached to the tibialis anterior (TA), rectus femoris (RF), medial gastrocnemius (MG) and biceps femoris (BF) muscles of the right leg. A Delsys Wireless Force Sensitive Resistor (footswitch) was attached to the right foot with sensors placed on the base of the great toe, first and fifth metatarsal heads and heel respectively. Body weight was taken using an electronic scale after electrode and footswitch attachment was complete and with subjects wearing their shoes during the measurement. Finally, participants were fitted to the appropriate size ZeroG™ harness and were given the opportunity to practice walking at all of the various levels of BWS (0/20/40/60/80%); the ZeroG^{TM} BWS system has been described in detail previously [8].

EMGworks Acquisition software program (Delsys EMGworks 4.0) was used to collect muscle activity data from the randomized BWS trials. All participants initially completed baseline trials while walking at natural cadence along a 15 m track at 0% BWS. BWS in subsequent trials was randomized between four different levels (20/40/60/80%) while harnessed to the ZeroG^{TM}; 3 trials were performed at each level, for a total of 15 trials, all at the subject's respective natural cadence.

Statistical Analysis

EMGworks Analysis (Delsys EMGworks 4.0) and Statistica 8 software programs were used for analysis of acquired data; trials for each level were kept separate. To investigate change in EMG activity at different levels of BWS, the root mean square (RMS) of the filtered signal energy was determined over 3 gait cycles and normalized to the average signal at baseline (0% BWS). A similar approach was used to assess changes in gait cycle components (e.g. swing and stance phase) with increasing support, by using the footswitch signal to determine when the foot made contact with the ground. The effect of BWS on muscle activation and gait cycle components were analyzed with repeated measure ANOVAs using the filtered EMG signals. Tukey's HSD was used for post-hoc testing when necessary. Statistical significance was set at $p < 0.05$ for all analyses.

RESULTS

No significant differences were found between trials or genders therefore the described results represent the combined data from the second trial.

Muscle Activity During the Gait Cycle (Heel Strike vs. Toe-Off)

The EMG profiles of muscle groups were nearly identical despite variations in the amount of bodyweight offset; however, significant differences were observed with respect to amplitude (Fig 2). Averaged muscle activity across the gait cycle decreased significantly at ≥ 40% BWS with a mean maximum reduction of 23.7% at 80% BWS compared to baseline (Fig. 3). This difference was attributed to a decrease in heel strike muscular activity as BWS increased, with a mean maximum reduction of 33.4% at 80% BWS compared to baseline; no significant differences were found in muscle activity during toe-off with increasing levels of BWS. Looking at the individual muscle groups, MG activity significantly decreased at ≥ 40% BWS, with RF and TA demonstrating significant decreases at higher levels of BWS (60-80% BWS). The decline in BF activity with increasing levels of BWS was not significantly different from baseline. Although MG activity was the first muscle group at heel strike to be affected by increasing levels of BWS, RF was most affected by BWS with a 62.8% decrease in activity at the highest level of BWS (80%) relative to baseline (Fig. 4).

Figure 2: Pattern of hip flexor (RF) muscle activity during the different phases of gait with varying % BWS. The phases of gait were defined as follows: 1) beginning of heel strike, 2) max heel strike, 3) cross between heel strike and toe off (foot flat), 4) max toe off, 5) end toe off, 6) mid swing and 7) end of cycle. RMS represents root mean square. Values are mean ± SE.

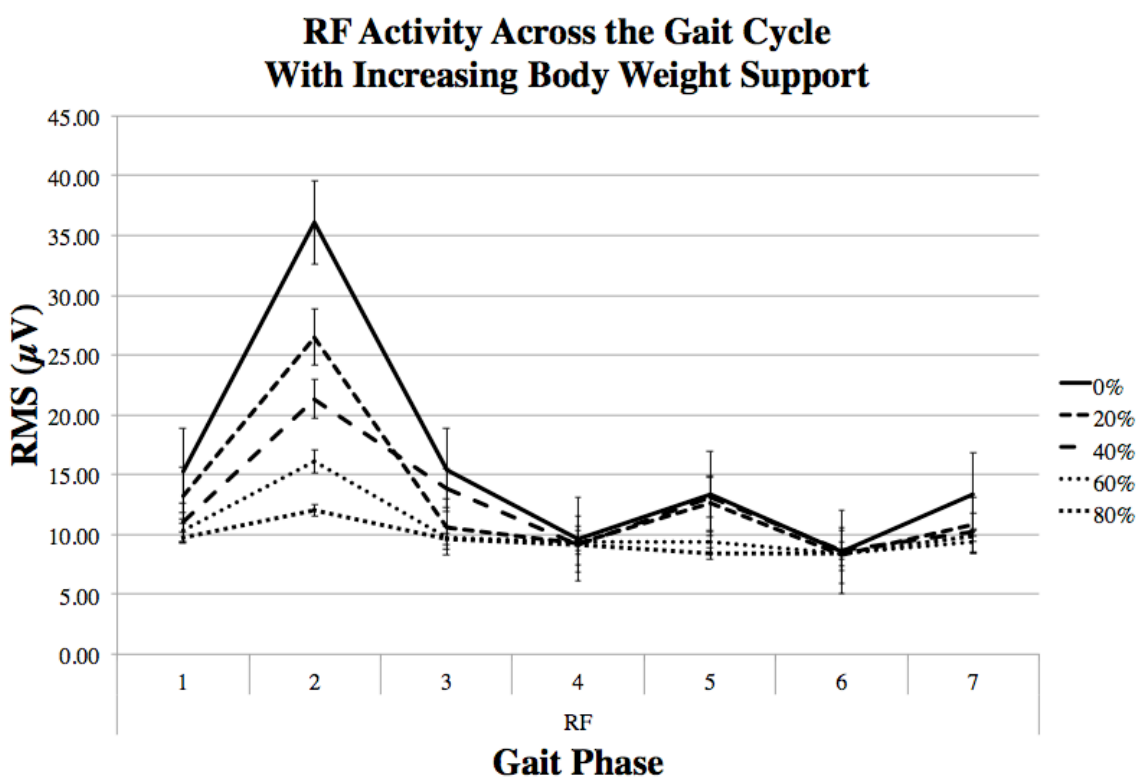


Figure 3: Overall muscle activity normalized to 0% BWS. RMS represents root mean square. Values are mean ± SE; *p < 0.05, **p<0.01.

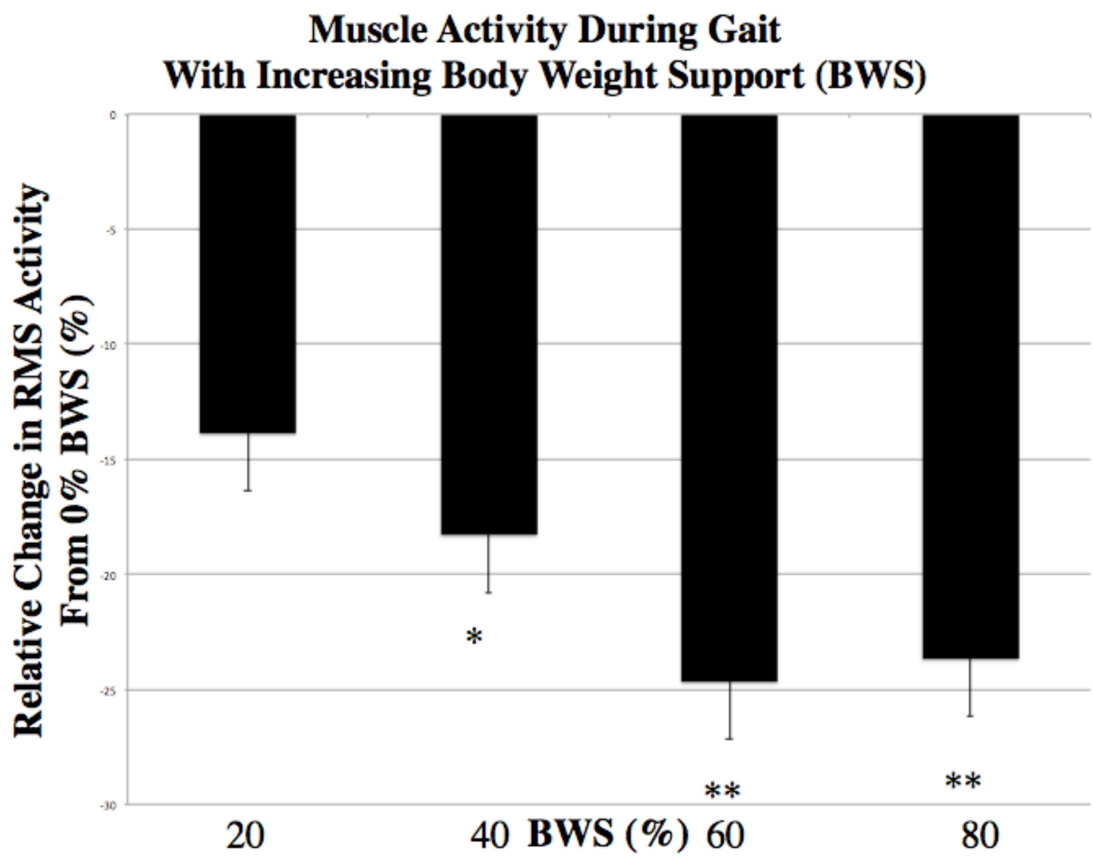
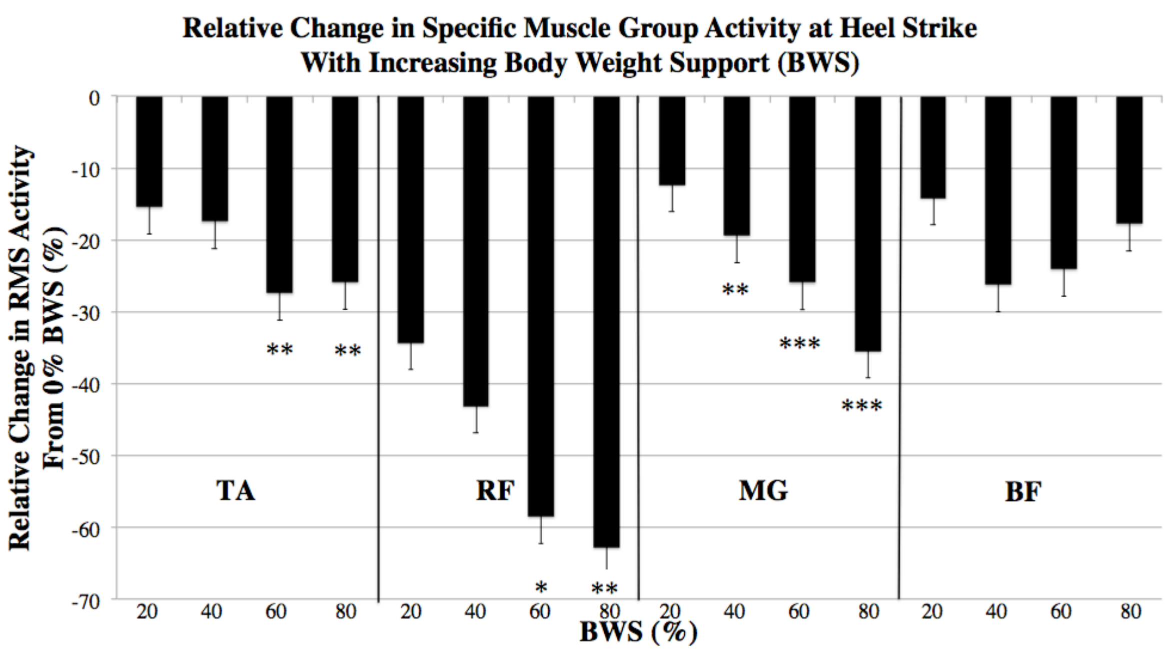


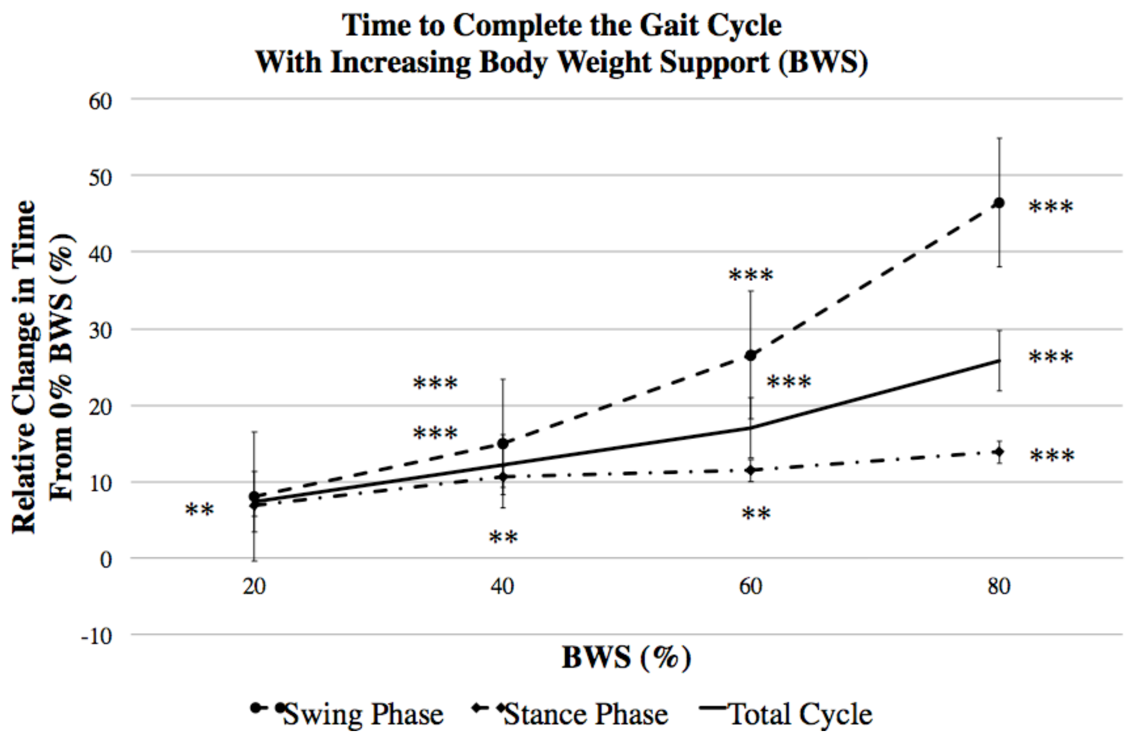
Figure 4: Heel strike muscle activity normalized to 0% BWS. Root mean square (RMS) of tibialis anterior (TA), rectus femoris (RF), medial gastrocnemius (MG), and biceps femoris (BF). Values are mean ± SE; *p < 0.05, **p < 0.01, ***p < 0.0001.



Time to Complete the Gait Cycle (Stance vs. Swing Phase)

The gait cycle took longer to complete with greater BWS, increasing by 25.8% at the highest level of BWS (Fig. 5). Swing and stance phase times were both significantly longer at ≥ 40% BWS, but the relative change was greater in the swing phase time (46.5% vs. 13.8% respectively).

Figure 5: Time to complete the gait cycle normalized to 0% BWS. Values are mean ± SE; **$p < 0.01$, ***$p < 0.0001$.



DISCUSSION

To our knowledge this is the first study to look at muscle activation patterns during gait with increasing levels of BWS using the ZeroG^{TM} training system.

Muscle Activity

Our data agrees with previous descriptions of “normal” patterns of muscle activation during walking with 0% BWS [9,10]. Some of these similarities include weight-accepting muscles (TA, RF, BF) peaking in activity during the first 15% of stride and distal supporting muscles (TA, MG) tending to be more active than proximal muscles (BF, RF). Differences in muscular activity during the walking trials in the present study were not significantly affected until 40% or more body weight was offset by the ZeroG™ training system. This is consistent with previous research evaluating body weight supported training, which demonstrated gait kinematics similar to baseline walking could be maintained when 30% or less support is provided [2].

In agreement with Hidler and colleagues, increasing BWS decreased muscular demand of the lower limbs during the walking trials thereby decreasing the intensity of the task [8]. In the present study, lower limb hip flexors and ankle plantarflexors were more affected by BWS (RF>MG) compared to ankle dorsiflexors and hip extensors (TA>BF). We hypothesized that the plantarflexor muscle group would be most affected by increasing levels of BWS; although MG activity was significantly affected at a lower level of offloading (40% BWS), the greatest decline in muscle activity with increasing BWS occurred in RF. RF activity was most affected at heel strike, and its activity decreased by as much as 63% at the highest level of BWS. Interestingly, at higher levels of BWS, RF's second activity burst, which is associated with forward acceleration of the limb in early swing, disappeared. Yang and Winter have attributed this disappearance to slower walking cadences where the acceleration required to swing the limb is low enough that the initial part of swing can be accomplished with minimal muscle activity, similar to a pendulum [10].

In this study, the non-significant change in BF activity may have been a result of large between-subject variability at the highest level of BWS possibly due to subjects resisting the vertical support from the ZeroG^{TM} training system in an attempt to maintain "normal" gait kinematics. The decrease in natural cadence at the highest levels of BWS may have also contributed to the increased variability in the signalling, a relationship highlighted in a study by van Hedel and colleagues [11]. The aforementioned study involved able-bodied subjects performing body weight supported treadmill walking at 0/25/50/75% BWS. Similar to the results in our study, the EMG patterns indicated RF and MG activity to be most influenced at the highest level of BWS, with only slight differences detected in TA activity. In contrast to our data, van Hedel and colleagues found large differences in lateral hamstring activation. Comparison of this data with the present study highlights the important differences that need to be considered when comparing overground and treadmill walking (e.g. hip range of motion) and suggests that knee kinematics may be more comparable than hip kinematics between the two modalities.

Gait Cycle Time

The slower natural cadence of participants in this study at ≥ 40% BWS supports previous research indicating that at higher levels of BWS individuals have difficulty moving their center of mass over their base of support [6]. It is also likely that subjects experienced reduced acceleration rates during locomotion at higher levels of BWS, which decreased walking velocity and step length, contributing to the prolonged gait cycle.

In the present study, increases in both swing and stance (ground contact) phase contributed to the slower gait cycle, with a greater contribution attributed to increased swing phase completion time by as much as 46.5%. This indicates that at higher levels of BWS individuals spend significantly more time balancing a portion of their body weight on one limb, while the opposite limb completes swing phase. This is in contrast to a previous overground BWS training study in persons with stroke, in which no changes in stance or swing period duration were evident with increasing BWS [5]. These differences may be attributed to the clinical population who potentially responded differently to the BWS, in addition to the different overground training devices used; the study by Sousa and colleagues used a device with an electric motor that relied on the individual for movement versus the current study in which the ZeroG™ uses a custom-series elastic actuator that moves independently of the participant.

Limitations

The fact that walking cadence was not controlled in the present study could be considered a limitation, however, all participants received the same instructions to walk at their natural cadence during the different trials. The sensitivity of EMG profiles to walking velocity is well known, resulting in decreased amplitude and increased variability of muscle activity at slower cadences. Changes in walking cadence affect the acceleration of the lower limb, which, according to Winter, impacts hip and knee musculature activity more than the ankle [12]. Future studies using the ZeroG™ which impose restrictions to step length (e.g. using floor tape markers) and walking velocity (e.g. using a metronome) would provide important information into the potential interaction effect of the above two factors with changing levels of BWS while using this device. These investigations will also provide a means of comparison to other studies including supported treadmill based walking studies, which have the ability to control for speed in addition to BWS. The clinical significance of the small differences observed in cadence and stride length still need to be determined. In addition, the difference between overground and treadmill walking needs to be acknowledged, as the treadmill belt, through its facilitation of hip extension, may alter the effect of offloading body weight compared to an overground device. The importance of these comparisons will better inform therapists in terms of modality selection, which may be modified based on the goal of the rehabilitation program for an individual.

The purpose of this study was to establish normative data (using able-bodied participants) describing the effect of BWS on muscle activation utilizing the ZeroG^{TM} overground training device. The potential generalizability of the presented results in clinical populations with mobility impairments is questionable, given the unique and wide-ranging functional limitations associated with impairments affecting gait. For example, it has been found that patients with neurological impairment lack an adequate push-off secondary to abnormal muscle activation during gait [2]. The important next step is to repeat this study using clinical populations in order to better understand the neuromuscular adaptations to BWS while using the ZeroG^{TM} device.

CONCLUSIONS

This is the first study to look at changes in gait patterns with increasing levels of BWS using the ZeroG^{TM} training system. The EMG profiles of muscle groups in this study were nearly identical despite variations in the amount of bodyweight offset; however, significant differences were observed with respect to amplitude. Thus, it can be concluded that the dynamic BWS provided by the ZeroG^{TM} decreases the muscular demand of the lower limbs without significantly altering muscle activation patterns during gait. Future research in clinical populations with mobility impairments (e.g. spinal cord injury, stroke) using this device should be conducted in order to compare the muscular activation response with the data acquired in the present study. This will help in the development of more effective and functional rehabilitation programs for these populations. The evidence from this study also emphasizes the importance of incorporating balance exercises into gait therapy as overground walking with ≥ 40% BWS forces individuals to support a portion of their body weight on one limb for a significantly longer period of time.

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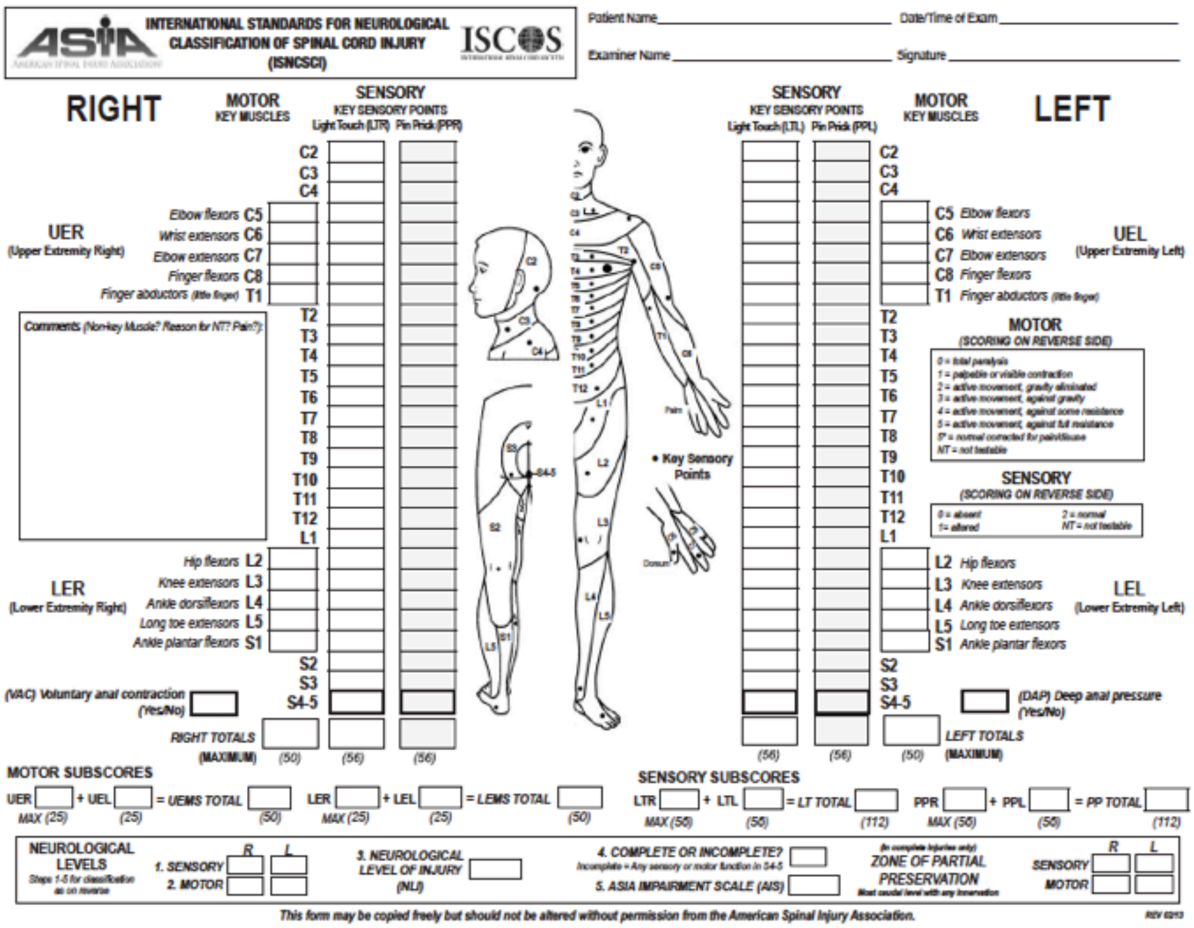
APPENDIX A: SPINAL CORD INJURY

International Standards for Neurological Classification of SCI Exam Worksheet.


INTERNATIONAL STANDARDS FOR NEUROLOGICAL CLASSIFICATION OF SPINAL CORD INJURY (ISNCSCI)

Patient Name ______ Date/Time of Exam ______

Examiner Name ______ Signature ______

RIGHT — MOTOR KEY MUSCLES — SENSORY KEY SENSORY POINTS Light Touch (LTR) Pin Prick (PPR)

C2, C3, C4

UER (Upper Extremity Right): Elbow flexors C5; Wrist extensors C6; Elbow extensors C7; Finger flexors C8; Finger abductors (little finger) T1

Comments (Non-key Muscle? Reason for NT? Pain?):

T2, T3, T4, T5, T6, T7, T8, T9, T10, T11, T12, L1

LER (Lower Extremity Right): Hip flexors L2; Knee extensors L3; Ankle dorsiflexors L4; Long toe extensors L5; Ankle plantar flexors S1

S2, S3, S4-5

(VAC) Voluntary anal contraction (Yes/No)

RIGHT TOTALS (MAXIMUM) (50) (56) (56)

• Key Sensory Points

SENSORY KEY SENSORY POINTS Light Touch (LTL) Pin Prick (PPL) — MOTOR KEY MUSCLES — LEFT

UEL (Upper Extremity Left): C5 Elbow flexors; C6 Wrist extensors; C7 Elbow extensors; C8 Finger flexors; T1 Finger abductors (little finger)

MOTOR (SCORING ON REVERSE SIDE)
0 = total paralysis
1 = palpable or visible contraction
2 = active movement, gravity eliminated
3 = active movement, against gravity
4 = active movement, against some resistance
5 = active movement, against full resistance
5* = normal corrected for pain/disuse
NT = not testable

SENSORY (SCORING ON REVERSE SIDE)
0 = absent 2 = normal
1 = altered NT = not testable

LEL (Lower Extremity Left): L2 Hip flexors; L3 Knee extensors; L4 Ankle dorsiflexors; L5 Long toe extensors; S1 Ankle plantar flexors

(DAP) Deep anal pressure (Yes/No)

LEFT TOTALS (MAXIMUM) (56) (56) (50)

MOTOR SUBSCORES
UER + UEL = UEMS TOTAL — MAX (25) (25) (50)
LER + LEL = LEMS TOTAL — MAX (25) (25) (50)

SENSORY SUBSCORES
LTR + LTL = LT TOTAL — MAX (56) (56) (112)
PPR + PPL = PP TOTAL — MAX (56) (56) (112)

NEUROLOGICAL LEVELS (Steps 1-5 for classification as on reverse) R L — 1. SENSORY 2. MOTOR

3. NEUROLOGICAL LEVEL OF INJURY (NLI)

4. COMPLETE OR INCOMPLETE? Incomplete = Any sensory or motor function in S4-5

5. ASIA IMPAIRMENT SCALE (AIS)

ZONE OF PARTIAL PRESERVATION (In complete injuries only) Most caudal level with any innervation — R L SENSORY MOTOR

This form may be copied freely but should not be altered without permission from the American Spinal Injury Association.

REV 02/13

Downloaded from the American Spinal Injury Association website: http://www.asia-spinalinjury.org/elearning/ISNCSCI.php. (American Spinal Injury Association, 2013).

APPENDIX B: BODY WEIGHT SUPPORTED LOCOMOTOR TRAINING

Body weight supported training devices used during walking sessions; from left to right – Manual Treadmill, Lokomat^{TM} and ZeroG^{TM}.

All devices provided body weight support (BWS) through a harness. The Andago GmbH treadmill (Loko) was used for robotic assisted (Lokomat^{TM}) and unassisted (Manual Treadmill) walking sessions. The ZeroG^{TM} had a custom series elastic actuator that traveled along an overhead trolley, which provided dynamic BWS while individuals performed overground walking sessions.

Manual Treadmill sessions completed with therapist assistance at both legs.

The Andago GmbH treadmill is used during these sessions without the robotic orthosis providing individuals with greater degrees of freedom during training.

Contraindications for body weight supported treadmill training with the Lokomat^{TM}.

The following contraindications must be observed in particular:

- Orthosis not adaptable to the patient's body (lower limbs)
- Body weight of more than 135 kg
- Severely fixed contractures
- Bone instability (non-consolidated fractures, unstable spinal column and severe osteoporosis)
- Open skin lesions in the area of the lower limbs and/or torso
- Circulatory problems
- Cardiac contraindications
- Uncooperative or (self-) aggressive behaviour, such as transitory psychotic syndrome
- Severe cognitive deficits
- Patients with (long-term) infusions
- Patients with extremely disproportionate growth of the legs and/or spinal column (e.g. bone or cartilage dysplasia)
- Severe vascular disorders of the lower limbs
- In general, patients who have been ordered to remain in bed or immobile due to, for instance, osteomyelitis or other inflammatory/infectious

disorders

- Hip, knee, ankle arthrodesis

Downloaded from the Hocoma website: http://www.hocoma.com/info/legal-notes/.

APPENDIX C: ETHICS AND RECRUITMENT

Body weight supported training devices used during walking sessions; from left to right – Manual Treadmill, Lokomat^{TM} and ZeroG^{TM}.

All devices provided body weight support (BWS) through a harness. The Andago GmbH treadmill (Loko) was used for robotic assisted (Lokomat^{TM}) and unassisted (Manual Treadmill) walking sessions. The ZeroG^{TM} had a custom series elastic actuator that traveled along an overhead trolley, which provided dynamic BWS while individuals performed overground walking sessions.

Poster used at the MacWheelers Spinal Cord Injury Rehabilitation Program at McMaster University and Regional Rehabilitation Centre at Hamilton General Hospital (Hamilton, ON).

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Interested in Participating in a Research Study?

PURPOSE:

To compare oxygen demand and muscle activation patterns in adults with incomplete spinal cord injury (SCI), while walking at the same level of body weight support using three different devices (manual treadmill, Lokomat - robotic treadmill, ZeroG - over-ground assistance).

MEASUREMENTS:

- Oxygen Demand
- Muscle Strength
- Muscle Activity
- User Preference

WHO CAN PARTICIPATE?

- Men and women
- Age 18 to 65 years
- 1+ year post-injury (incomplete SCI)

Your participation would involve 4 sessions (approximately 45 to 60 min per session) at the Robert Fitzhenry Specialized Rehabilitation and Exercise Lab (McMaster University, Hamilton).

For more information or to volunteer, please contact:

Alyssa Fenuta, HBSc
fenutaam@mcmaster.ca

Poster used by Spinal Cord Injury Ontario – Outspoken Magazine and E-Spoken.

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Participants Needed for a Walking Study

Adults with an incomplete spinal cord injury (1+ year post) are invited to participate in a study comparing oxygen demand and muscle activation patterns while walking using three different body weight supported devices. Participation involves 4 sessions, each approximately 1 hour, at McMaster University (Ivor Wynne Centre).

For additional information or to volunteer contact Alyssa Fenuta, HBSc:
fenutaam@mcmaster.ca

Sample letter of information/consent.

LETTER OF INFORMATION / CONSENT

A Study Comparing Oxygen Demands and Muscle Activity Patterns During Different Forms of Body Weight Supported Locomotion

Investigators:

Student Investigator:

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Purpose of the Study

Body weight supported training has created enthusiasm as a rehabilitation tool to enhance recovery of both ambulation and other health outcomes following a spinal cord injury (SCI). The latest technology now provides the ability for individuals to perform overground walking with dynamic bodyweight support (ZeroG^{TM}). It is not yet known how the bodyweight supported treadmill and ZeroG^{TM} walking track differ in terms of the oxygen demand or muscle activity during weight-supported walking in people with SCI. The purpose of this study is to compare oxygen demands and muscle activation patterns while participants with incomplete SCI (18-65 years old) complete a body weight supported training session using a) manual treadmill, b) robotic (Lokomat) treadmill, and c) overground (ZeroG^{TM}) assistance at the same level of dynamic bodyweight support. For comparison purposes, each participant will be matched with an able-bodied control (same sex, within 5 years of age of participant with incomplete SCI). Able-bodied individuals are considered as those who are free of musculoskeletal disease or injury (> 1 year post diagnosis).

Procedures Involved in the Research

You will be asked to come for four sessions at the Robert Fitzhenry Specialized Rehabilitation and Exercise Lab at McMaster University to participate in the study. On the first visit (60 min) you will complete a VO_2 peak test on an arm ergometer and be fitted to the ZeroG^{TM} to measure the minimal amount of body weight support required to complete a walking trial (Figure 1a). You will also perform lower limb strength testing (knee flexion/extension, hip flexion/extension) using the Lokomat (Figure 1b). The VO_2 peak test involves you exercising on an arm bike until you are exhausted and you will wear a fitted mask to your mouth so that we can collect the gases you breathe in and out while you perform this task. We will be gradually increasing the resistance on the arm bike until you feel you are unable to continue. The test will last between 6-10 minutes. The subsequent sessions (30 min/session) will involve completing a 10-15 minute walking session (achieving a minimum of 2 minutes of steady state walking) at a self-selected pace at the same level of body weight support using 1) manual and 2) robotic (Lokomat) assistance on the treadmill, as well as 3) overground using the ZeroG^{TM} (Figure 1). The skin on both legs will be lightly cleaned with an alcohol swab (and possibly shaved) in select areas to allow for special electrodes to be taped onto the major muscle groups of your right leg (quadriceps, hamstrings, calves, etc). These electrodes will monitor your muscle activation. All body weight supported devices will be used according to manufacturer instructions; you will be fitted to a harness that is connected to the Hokoma treadmill or ZeroG^{TM} overground training system and will be secured to the Lokomatic's robotic limbs. During each of the subsequent testing sessions, a mask will be fitted to your mouth to be worn to measure oxygen uptake and a heart rate monitor will be worn to ensure appropriate response to exercise.

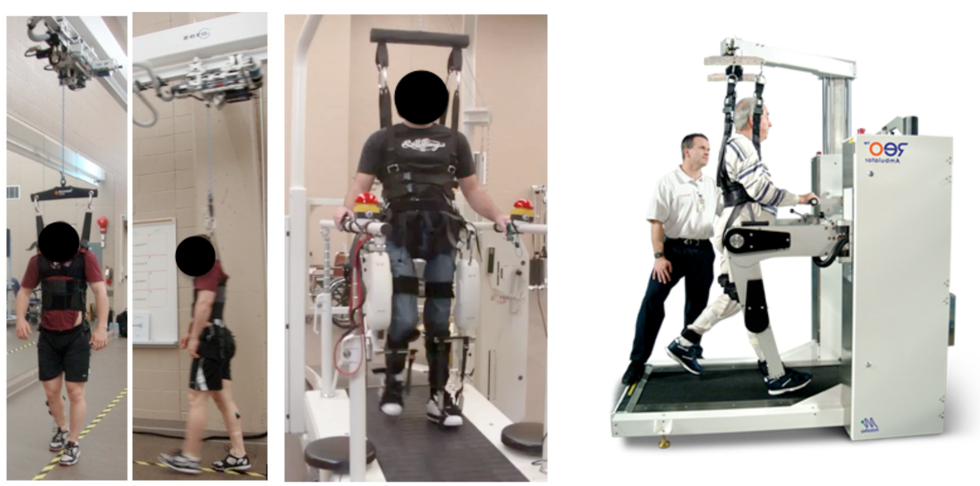


Figure 1: Body weight supported training devices to be used in the study: a) ZeroG^{TM} - overground training device, b) Lokomat - robotic assistance on Hocoma treadmill c) Hocoma treadmill without robotic assistance.

Potential Harms, Risks or Discomforts:

The risks involved in participating in this study are minimal; all sessions will take place in Robert Fitzhenry Specialized Rehabilitation and Exercise Lab and will be supervised by trained staff. There are some physical risks associated with completing the exercise bouts and the maximal exercise test. Your muscles might be tired and sore following your sessions and you may feel fatigued, but this fatigue will resolve within 15-30 minutes of completing the test. You will experience an increase in heart rate and may feel out of breath.

Discomfort that might result from the harness is likely greater than what you might encounter in every day life. You may feel uncomfortable with the tightness of the harness on your body, but padding will be used to minimize discomfort. The mouthpiece worn to measure oxygen demand during the walking sessions can also become uncomfortable, but it will only be worn for a short period (10-15 minutes) during each session.

Small areas on your leg muscles may need to be shaved to ensure a proper adhesion of the electrodes that measure muscle activation. There is a slight risk of cutting the skin or inducing skin irritation or a rash as a result of the shaving process. Although very rare, you may experience a temporary reaction to the adhesive from the surface electrodes.

It is important to note that you can withdraw (stop taking part) at any time. I describe below the steps I am taking to protect your privacy. Only researchers involved with the study will have access to your data (which will be coded) and all documentation will be securely stored within the Department of Kinesiology.

Potential Benefits

This study provides no direct or immediate benefit to the participants involved. The information acquired will be added to the knowledge base to benefit SCI rehabilitation procedures in the future. Currently there are different body weight supported locomotor devices available but to date there is no evidence of the best training stimulus with respect to oxygen uptake or muscle activity. By learning how muscle activation and oxygen uptake during locomotion changes when using these different training devices, despite providing the same level of body weight support, we can then see if these changes hold true in a training study. This will help us to tailor more effective and functional rehabilitation programs for individuals with SCI.

Confidentiality

Since this study is being conducted in a public setting (e.g. Robert Fitzhenry Specialized Rehabilitation and Exercise Lab, McMaster University) complete anonymity regarding your participation cannot be guaranteed. The data you provide however will remain confidential (e.g. coded) and will be kept in a locked cabinet where only my supervisor (Dr. Audrey Hicks) and I will have access to it. Information kept on a computer will be protected by a password. Once the study is complete, an archive of the data, without identifying information, will be deposited on a secure memory stick that only Dr. Hicks will have access to. Participants will be coded so no names will be associated with the acquired data. The data will be compiled into a Masters thesis document with intention to publish the findings in a rehabilitation journal. Any files used in the development of these publications will be left in the possession of Dr. Audrey Hicks where only she shall have access to it.

Participation and Withdrawal

Your participation in this study is voluntary. It is your choice to be part of the study or not. If you decide to be part of the study, you can decide to stop (withdraw), at any time, even after signing the consent form or part way through the study. If you decide to withdraw, there will be no consequences to you (e.g. continuation of services within the Department of Kinesiology and/or MacWheelers). In cases of withdrawal, any data you have provided will be destroyed unless you indicate otherwise. Upon completion of data analysis (April 2012) withdrawal is no longer possible.

Information About the Study Results

I expect to have this study completed by approximately June 2013. If you would like a brief summary of the results, please let me know how you would like it sent to you.

Questions About the Study

If you have questions or need more information about the study itself, please contact me at fenutaam@mcmaster.ca. This study has been reviewed by the McMaster University Research Ethics Board and received ethics clearance. If you have concerns or questions about your rights as a participant or about the way the study is conducted, please contact McMaster Research Ethics Secretariat at ethicsoffice@mcmaster.ca.

CONSENT

I have read the information presented in the information letter about a study being conducted by Alyssa Fenuta, of McMaster University. I have had the opportunity to ask questions about my involvement in this study and to receive additional details I requested. I understand that if I agree to participate in this study, I may withdraw from the study at any time. I also acknowledge that by consenting I have not waived any right to legal recourse in the event of research-related harm. I have been given a copy of this form. I agree to participate in the study.

□ Yes, I would like to receive a summary of the study's results.
Please send them to this e-mail address:
_____________________________________.

□ No, I do not want to receive a summary of the study's results.

Name of Participant (Printed): ___________________________________

Signature: ____

Date: ____

APPENDIX D: DATA COLLECTION

Sample pre-screening data collection sheet.

Demographics:

Name	
DOB (mm/dd/yy)	
Age	
Gender	
Injury (type/level)	Chronic (1+ post diagnosis) Incomplete ________________________________
Cause of injury	
Height (cm)	
Weight (kg)	
Lower limb dominance	
More affect side? (if any)	
Autonomic Dysreflexia	
Spasticity	
Medications	
Other	

Experience Using BWS Devices for Locomotion:

Device	Total Time	Days/Week	Time/Session
Manual Treadmill			
Lokomat™			
ZeroG™			

Isometric Lower Limb Strength:

Joint	Moment (Nm)
Hip right flex	
Hip right ext	
Hip left flex	
Hip left ext	
Knee right flex	
Knee right ext	
Knee left flex	
Knee left ext	

Lokomat™ Settings:

Guidance Force	100/100
Hip ROM (deg)	

	Offset (deg)	
Knee ROM (deg)		
	Offset (deg)	
Patient Co-efficient		
Speed (km/h)		
BWS (kg)		
	BWS (%)	

Patient record sheet: Adult gait orthosis settings

User:	Date:

Surname:	Height:
First name:	Weight:

Harness size:		Foot lifter (L/R):	
Leg strap size:		Parallel bar height:	

Back cushion depth:

Back cushion height:

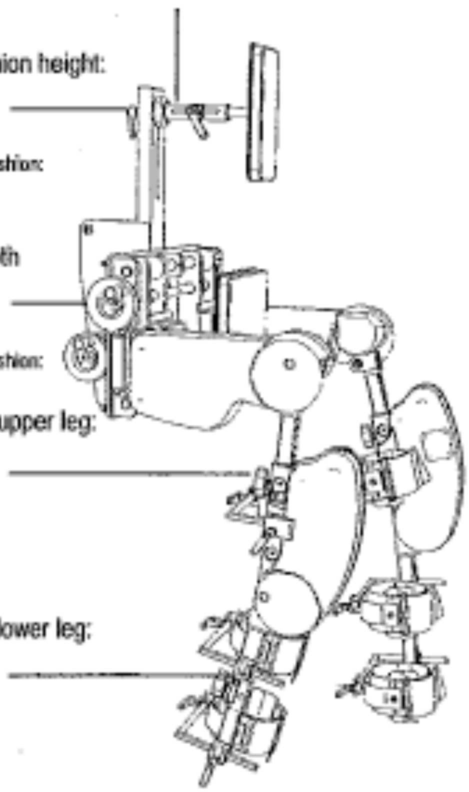
Thick/thin cushion:

Pelvic depth

Thick/thin cushion:

Length of upper leg:

Length of lower leg:



Size of **upper** cuff: L R

L R Lat. value

L R Lon. value

Size of **middle** cuff: L R

L R Lat. value

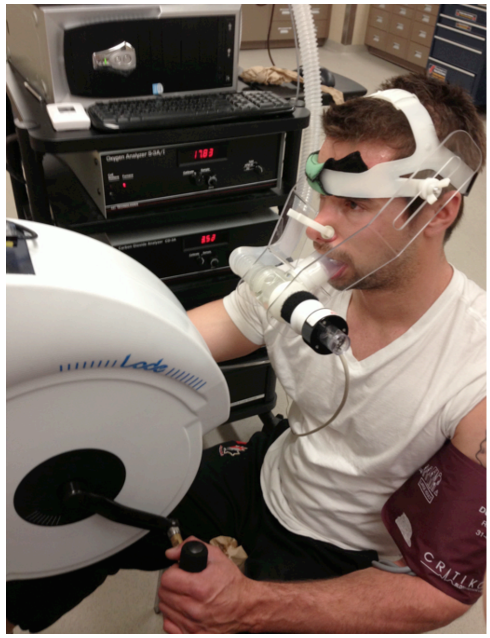
L R Lon. value

Size of **lower** cuff: L R

L R Lat. value

L R Lon. value

Participant set-up for VO_2 peak test on the arm ergometer.



Sample VO_2 peak testing data collection sheet.

Temp (°C): _ Humidity (%): ___ Barometric Pressure (mmHg): _ FiO_2 (fraction): _

Age Predicted Max Heart Rate (220-age) **paraplegic only: _______

Empty Bladder? ☐ Yes ☐ No	☐ Power ☐ Manual Chair

HR/BP @ Baseline: Cadence: _________ Cuff Size: ________

	HR	**BP**	**MAP**	**VO_2**
1:00				
3:00				
5:00				

VO_2 Peak Protocol: Cadence: _________

Time	**PO (W)**	**HR (BPM)**	**RER**	**VO_2 (ml/kg/min)**	**RPE (C/P)**
WU: ExpAir Time: ______________					

WU									
VO_2 Peak: ExpAir Start Time: ______________									
1:00									
2:00									
3:00									
4:00									
5:00									
6:00									
7:00									
8:00									
9:00									
10:00									
11:00									
12:00									
Cool Down Start - ExpAir Time: ________ Duration of Test: _______________									

HR/BP @ Recovery: Cadence: _________ Cuff Size: ________

	HR	BP	MAP	VO_2
1:00				
3:00				
5:00				

Summary Data:

	Resting	Max Achieved	Predicted	% Predicted
Freq (br/min)				
Vt (mL)				
Ve (L/min)				
VO2 (mL/min)				
VCO2 (mL/min)				
VO2/kg (mL/kg/min)				
RER				

HR (BPM)				
O2 Pulse (mL/b)				
Watts				
Mix O2 (%)				
Mix CO2 (%)				
VE/VO2 (L/mL)				
RPE Central				
RPE Peripheral				

Rating of perceived exertion (RPE) scale and anchor terminology.

0-10 Borg Rating of Perceived Exertion Scale	
0	Rest
1	Really Easy *Beginning of VO2 Peak Test*
2	Easy
3	Moderate
4	Sort of Hard
5	Hard
6	
7	Really Hard
8	
9	Really, Really Hard
10	Maximal *End of VO2 Peak Test*

Sample walking session data collection sheet.

Age Predicted Max Heart Rate (220-age) **paraplegic only: ______

Treadmill Speed (km/h): _______ BWS (%): _________ BWS (kg): _______

HR/BP @ Baseline:

	HR	**BP**	**MAP**	**VO_2**
1:00				
3:00				
5:00				

Start time for Warm-Up (WU): ________ Start time for $VO_{2steadystate}$: ___________

Time	**VO_2**	**HR (BPM)**	**$RPE_{Central}$**	**$RPE_{Peripheral}$**

	(ml/kg/min)			
WU				
0:30				
1:00				
1:30				
2:00				
2:30				
3:00				
3:30				
4:00				
4:30				
5:00				

HR/BP @ Recovery:

	HR	BP	MAP	VO_2
1:00				
3:00				
5:00				

Sample individual body weight supported training equipment evaluation form. [X = a) manual treadmill or b) treadmill with robotic assistance (Lokomat^{TM}) or c) ZeroG^{TM} (overground)]

Considering the exercise session you completed on the X please answer the following statements:

Have you used this piece of equipment before?

	Yes
	No

If yes, how often do you use this piece of equipment?

	Rarely
	Once per week
	Twice per week
	Once or twice per month
	More than two times per week

Please indicate the extent to which you agree with the following statements:

	Strongly Disagree						**Strongly Agree**
The exercise session intensity was appropriate for my ability	1	2	3	4	5	6	7
I was able to complete the exercise without any additional pain/discomfort	1	2	3	4	5	6	7
Overall, I enjoyed using this specific piece of exercise equipment	1	2	3	4	5	6	7
I would recommend exercise facilities purchase this specific piece of exercise equipment	1	2	3	4	5	6	7
This piece of equipment would be useful to improve my fitness and to help me perform my activities of daily living	1	2	3	4	5	6	7

Assuming that you are very motivated and fit, for **how many minutes** could you imagine yourself using this specified piece of exercise equipment without stopping?

	Less than 5 minutes
	5 minutes
	10 minutes
	15 minutes
	20 minutes
	25 minutes
	30 minutes
	Other (please specify):

Assuming that you are very motivated and fit, how many **times per week** could you imagine yourself using this specific piece of exercise equipment?

	1
	2
	3

	4
	More than 4

In your opinion, who could safely use this specific piece of exercise equipment (check all that apply)

Exercise Status		Injury Characteristics		Other	
	New Exercisers		ASIA A		
	Experienced Exercisers		ASIA B		
	Someone with low fitness levels		ASIA C		
	Someone with high fitness levels		ASIA D		
			Paraplegia		
			Tetraplegia		

Sample final comparative body weight supported training equipment evaluation form.

For the following questions please use a ranking system of **1 (“best”) to 3 (“worst”).**

Considering the three exercise sessions you have completed throughout the duration of this study, please rank the following body weight supported training devices based on

a) *your* **personal preference**:

	Manual treadmill assistance
	Robotic treadmill assistance (Lokomat^{TM})
	Overground assistance (ZeroG^{TM})

b) *your* **recommendation for exercise facilities to invest in the equipment**:

	Manual treadmill assistance
	Robotic treadmill assistance (Lokomat^{TM})
	Overground assistance (ZeroG^{TM})

c) *your* **belief that training sessions will help improve your fitness and ability to perform activities of daily living**:

	Manual treadmill assistance
	Robotic treadmill assistance (Lokomat™)
	Overground assistance (ZeroG™)