To my husband, Scott, and our future family
ACKNOWLEDGMENTS

As I complete this major milestone in my academic career, I am amazed at what I have experienced and accomplished over the last four and a half years. At the same time, I also am fully aware that I did not walk alone on this journey. Words cannot fully express the gratitude I have to so many for standing by my side during this marathon process: mentors, family, and friends.

I would like to express my sincerest gratitude to my primary mentor, Dr. Andrea Behrman. The undying passion she holds for her work both as a researcher and clinician inspires me every single day. She has shown me what it means to be a true “clinical” researcher; I now know it is possible to have a career based on what your heart guides you to do. In addition to Dr. Behrman, I have been extremely grateful to have one of the strongest graduate committees possible: Drs. Steve Kautz, Dena Howland, and Craig Velozo. Every graduate student should be so fortunate to have a committee who cares enough to find the perfect balance between support and challenge. It is that balance, which helped me find my potential as a “budding” clinical researcher.

I would like to extend a sincere thank you to the T32 Neuromuscular Plasticity Training program for providing several years of financial support and endless opportunities for growth. I am grateful to the Florida Brain and Spinal Cord Injury Program for the honor of being an Early Career Rehabilitation Research Award recipient. My appreciation also goes to the Department of Physical Therapy and Ms. Gloria Miller for developing a “creative” teaching assistantship position that continued to support me through the last semester of my doctoral training.
This journey would not have been as enjoyable as it was had it not been for the many friends I have made along the way. Those friends are the many who welcomed me to the university back in 2006 and others who arrived along the way. I am especially grateful to my “coffee-chat” friends for the great times when we just needed time to talk and people to listen.

My family also has been an incredible support team. This includes not only my immediate family, but also my extended family and in-laws. Even with most of them several states away, I always knew they were proud of me, no matter what I did or did not accomplish. In the end, they just wanted me to be happy and healthy.

Finally, I believe more than a lifetime would be needed to express to my husband, Scott, exactly what he has meant to me throughout this process. Every dream of mine, he has encouraged, and earning a doctorate has been no exception. In times of doubt, he showed me my capabilities. In times of frustration, he gave me laughter. When separated by almost a thousand miles for two years, he showed me what unconditional love and support really are. For everything, I am eternally grateful.
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6-2 Variability of outcome measures standardized to control data pre vs. post locomotor training for 10 SCI participants .........................................................158
Dynamic stability, defined as the ability to control one’s center-of-mass (CoM) within a moving base-of-support, is a co-requisite of functional walking. Following a spinal cord injury (SCI), dynamic stability is impaired. One clinical approach for dynamic stability deficits utilizes assistive devices (ADs) to compensate for functional losses post-SCI. However, SCI rehabilitation has recently begun a transition from compensatory therapies toward activity-based therapies, which target walking recovery. To remain consistent with this therapeutic shift, measurements should be conducted in this same recovery context. Therefore, the purpose of this dissertation was to examine movements that individuals with SCI employ to maintain dynamic stability using a framework that parallels the rehabilitation paradigm shift. Measurements utilized were based on scientific evidence of nervous system priorities to maintain stability.

The first experiment investigated head stability during walking with and without devices to understand the effect of ADs, as a conventional evaluation approach, on head stability. Additionally, this study aimed to determine how head stability differed between individuals after injury walking without ADs and healthy persons. This testing condition without devices remained through subsequent studies. The second
experiment examined dynamic stability via foot placement analysis relative to the 1) opposite foot, 2) CoM, and 3) CoM plus its velocity. Finally, the third experiment assessed the differential effects of manual-assisted and robotic-assisted locomotor training on dynamic stability. CoM trajectory length per stride was used as a primary indicator of stability control with trunk motion and spatial foot parameters examined as secondary outcomes. Collective findings across studies indicated that 1) ADs have a role in head stability, and when ADs are removed, individuals post-injury exhibit less stability than controls; 2) persons with SCI demonstrate greater variability in all measures of foot placement compared to controls and appear to maintain stability through a continuous pattern of corrective foot placements; and 3) both manual-assisted and robotic-assistive locomotor interventions have training benefits for dynamic stability, but the trunk and feet strategies used to maintain CoM control vary between interventions. This work is a first step in dynamic stability analysis post-SCI and lays the groundwork for further investigations in this field.
CHAPTER 1
LITERATURE REVIEW

Introduction

Spinal cord injury (SCI) is among the most devastating and incapacitating acquired medical conditions today. Every year, 12,000 new cases occur in the United States alone resulting in an estimated 255,700 persons chronically disabled by this condition currently (National Spinal Cord Injury Statistical Center 2008). Although young adult males remain the most frequently injured population, a recent trend shows an increasing number of injured adults over 60 years of age. Furthermore, motor vehicle crashes are the most common cause of injury, with falls rising as the second most frequent cause, above sports and violence. Despite the substantial amount of physical trauma often sustained by such incidents, the probability of sustaining a complete SCI with total paralysis and loss of sensory function has decreased in recent years secondary to improvements in emergency medical management (Bernhard et al. 2005). As a result, the percentage of persons discharged from the inpatient hospital with incomplete injuries, who demonstrate an emergence of sensory and/or motor function has increased. This clinical transition from complete to incomplete injuries has led to a higher number of individuals with a prognosis to recover functional abilities.

Both the extent and location of a spinal cord lesion influence the possible combination of physical and psychosocial impairments observed after an injury. Depending on the severity of these impairments, individuals post-SCI commonly experience reduced functional independence and quality of life as they are limited in the ability to fully return to their prior lifestyles. Physical consequences of SCI may include, but are not limited to, paralysis or paresis, diminished sensation and proprioception,
incoordination, and spasticity (Scivoletto et al. 2008). Functionally, these consequences may manifest as difficulties in standing and walking. Reportedly, recovery of walking ability is a highly coveted goal relative to recovery of other functions for persons who have sustained a SCI, regardless of injury level or chronicity (Ditunno et al. 2008). Additional research indicates that regaining walking ability is a first or second priority for approximately 38 percent of individuals with paraplegia and is of higher importance for persons with tetraplegia who have been injured less than three years (Anderson 2004). Thus, understanding how walking is generated and controlled as well as developing effective recovery-based walking interventions and outcome measures are critical for individuals to achieve their personal goals and to maintain health while avoiding adverse consequences of prolonged sitting versus standing or walking (e.g. compromised skin integrity, postural limitations, cardiovascular and respiratory co-morbidities, pain, osteoporosis) (Jacobs and Nash 2004).

Literature in both humans and animal models post-SCI highlights control of dynamic stability, or balance, as an essential component for successful walking recovery (Ladouceur et al. 1997; Deliagina et al. 2008; Scivoletto et al. 2008; Karayannidou et al. 2009). That is, an individual must acquire the capacity to walk upright, load through his/her lower extremities, and counteract the internally- and externally-generated forces created by the dynamics of the a moving body. However, because of the complexities of the task of walking, which involve integration of neural, muscular and biomechanical properties, in combination with the heterogeneity of each SCI, little attention has focused on the movement strategies or mechanisms necessary to achieve balance. Rather, walking balance control studies have investigated healthy
individuals (England and Granata 2007), elderly persons (Menz et al. 2003b; Kavanagh et al. 2005a), persons with musculoskeletal impairments (Hof et al. 2007), or individuals with other neurological disorders such as Parkinson’s disease, who generally comprise a more homogenous population, relative to persons with SCI, with characteristic movement patterns (Rochester et al. 2004; Oates et al. 2008). Only a few recent studies illustrate balance control in a sample of individuals with SCI evaluating either vestibulospinal integrity (Wydenkeller et al. 2006; Liechti et al. 2008) or automatic postural responses during standing perturbation tasks (Thigpen et al. 2009). Other studies in SCI that address walking evaluate stepping performance in the body weight support and treadmill environment, overground with assistive devices (Dobkin et al. 2006; Behrman et al. 2008; Bowden et al. 2008), and/or when adapting to environmental constraints (e.g. inclines) (Leroux et al. 1999; Pepin et al. 2003b).

Literature is needed, which directly investigates the mechanisms or movement strategies that persons with SCI employ to maintain walking balance when required to load through their lower extremities and support themselves without physical assistance. Such investigations would provide insight into the intrinsic capability of the nervous system to perform the specific task of walking and assist in reaching a consensus for how walking balance should be measured. Without an initiation of these investigations, advancement of walking interventions as well as measurement of patient progress and intervention effectiveness will be limited to compensatory strategies, which confine walking performance to external assistive devices.

Therefore, the purpose of this dissertation is to elucidate the motor control strategies that persons with incomplete SCI, who possess a heterogeneous ensemble
of impairments, implement to overcome deficits in dynamic stability during walking. Various biomechanical measurement techniques will be utilized to understand how individuals who exhibit any ability to self-generate at least a few steps post-injury individually solve the ubiquitous dilemma of balance control. The literature review will progressively outline research in the fields of locomotion, SCI rehabilitation, and balance control and its measurement to provide a foundation for the three studies to follow. More specifically, major sections of the literature review include 1) the neural control of locomotion with emphasis on the subtasks of walking and neurophysiological evidence of spinal pattern generation, 2) traditional compensatory approaches of walking rehabilitation post-SCI, evidence of activity-dependent plasticity, and the concurrent shift in walking rehabilitation towards recovery-based interventions, 3) balance control during standing and walking, its prerequisites, and contributory neural substrates, 4) the effect of SCI on balance control, and 5) current laboratory and clinical measures of balance. Subsequently, an explicit rationale and framework for the three complementary studies will be discussed, followed by methodological procedures and considerations in instituting each experiment.

**Neural Control of Locomotion**

**Subtasks of Walking**

Human walking is a complex task that requires the neural, muscular, and biomechanical integration of three main sub-components: 1) reciprocal stepping, including forward propulsion and hip extension at the stance-to-swing transition, 2) balance control, and 3) adaptability. Forssberg and colleagues first described this combination of fundamental tasks in the animal literature with locomotor studies in spinalized cats (Forssberg et al. 1980a; Forssberg et al. 1980b). Since that time, both
basic and clinical scientists have continued to demonstrate and express the necessity of these tasks for successful ambulation in animals and humans (Barbeau 2003).

The basic reciprocal stepping pattern in walking involves the alternating flexion and extension of the lower limbs. Additionally, this component of walking requires that the trailing limb propel the body forward for a continual translation overground. Moreover, hip extension as the lower extremities progresses from the stance to swing phases is a critical sensory cue to elicit this alternating interlimb pattern (Dietz et al. 2002). However, while the lower limbs are constantly moving, the upper body in contrast, requires a high amount of stability to keep the head, which houses essential visual and vestibular balance control apparatuses, relatively undisturbed. Therefore, balance control also is necessary to maintain an upright body posture and equilibrium in spite of the challenges that stepping presents. Finally, individuals rarely walk in isolation from the demands of the outside world, and their walking behavior typically is goal-directed. As a result, a person needs the ability to adapt successfully to the environment to achieve his/her personal goals (Shumway-Cook et al. 2002). In a healthy, uninjured nervous system, these subtasks integrate together to form a seemingly simple and automatic walking pattern. However, in reality, the production of walking requires a high degree of neural control to create such a smooth, rhythmic output.

Spinal (Central) Pattern Generation

The central nervous system (CNS) has an innate capacity to generate a rhythmical, motor output, such as the stepping behavior seen in walking, without afferent or supraspinal input. Interneuronal networks known as central pattern generators are responsible for producing that motor pattern (Grillner and Wallen 1985). Although these neural circuitries have been found throughout the CNS for the
production of vital functions such as breathing and mastication, central pattern
generators specific to locomotion are located in the spinal cord and known as spinal
pattern generators (SPGs) (Jordan et al. 1992; Marder and Calabrese 1996). Though
SPGs can self-sustain a stereotypical motor pattern, the locomotor output also can be
modulated and refined through incoming information from the periphery and supraspinal
commands to produce a relatively automatic behavior even in the presence of internal
and external demands (Dietz 2003; Edgerton et al. 2004). Following injury to the spinal
cord, reliance on afferent input to SPGs may greatly increase secondary to disruption of
descending neural flow (Edgerton et al. 2004) (Figure 1-1). Findings from animal
preparations of SCI described in the literature have been able to confirm the presence
of SPGs (Edgley et al. 1988; Juvin et al. 2007). However, in humans, while many
believe only indirect behavioral evidence exists secondary to the invasive procedures
required to isolate the SPG circuitry (Illis 1995; Duysens and Van de Crommert 1998),
others contend that direct evidence does exist (Dimitrijevic et al. 2005).

Evidence from Animal Literature

Early in the twentieth century, C.S. Sherrington and T.G. Brown published seminal
work on the locomotor capabilities of cats, which would later change the way the
scientific community viewed the intrinsic potential of the spinal cord and rehabilitation
post-SCI. Sherrington’s initial experiments demonstrated that cats decerebrated at the
brain stem level who received “direct stimulation of the cross section of the spinal axis”
exhibited a rhythmic hindlimb flexion-extension pattern that was of central origin
(Sherrington 1910). Subsequently, Brown altered this animal preparation, showing that
decerebrate, deafferented, T12 spinalized cats also were able to produce bilateral,
reciprocating flexion and extension contractions of the tibialis anterior and
gastrocnemius muscles. Based on these findings, Brown proposed a model of a “central mechanism consisting of antagonistic centres” in the lumbar spinal cord. This model theorized that each center was paired such that when flexor activity was generated in one center, extensor activity was inhibited. The halves allowed for oscillation between hindlimb movements, thus allowing the forward progression in locomotion. Furthermore, Brown regarded proprioceptive feedback to this central mechanism as highly important for adaptability in the environment, but that its presence was purely “regulative, not causative.” Despite the lack of intricate detail in this model relative to knowledge possessed today, the general concept triggered the production of vast amounts of animal locomotor research to understand SPG circuitry.

According to Burke et al., the features that best identify the existence of SPGs are “recognizable and reproducible patterns of rhythmic output in the absence of instructive external drive from other parts of the CNS or from peripheral sensory feedback” (Burke et al. 2001). Therefore, some of the strongest evidence of SPG networks in animals use experimental paradigms that induce fictive locomotion. Fictive locomotion involves blocking neuromuscular activity via pharmacological means or motor nerve transection, thus preventing movement-related sensory feedback. Following decerebration, spinalization, and/or spinal cord isolation as is noted in cat, rat and lamprey work (Edgley et al. 1988; Juvin et al. 2007; Mentel et al. 2008), sites within the CNS such as the mesencephalic locomotor region or spinal cord are electrically stimulated or perfused with neurotransmitters like dopamine or nialamide (Nielsen et al. 2005; Mentel et al. 2008). Subsequently, efferent neuronal recordings are obtained. The pattern of efferent activity seen has been consistent with the alternating rhythm of extensor or
flexor activity visible in intact vertebrate locomotion. Moreover, this activity has been observed in both the forelimbs and hindlimbs in spinal cats and rats to demonstrate the coordinated coupling of limb movements in quadrupedal locomotion (Yamaguchi 1992; Nielsen et al. 2005; Juvin et al. 2007).

One of the simplest vertebrate models of spinal pattern generation that also provides the greatest amount of mechanistic detail is the lamprey model. The lamprey has elicited proof of SPGs from the most basic, cellular level. Using the isolated spinal cord in vitro, N-methyl D-aspartate (NMDA)-receptor agonist baths have produced a cyclic pattern with a 1% phase lag of ventral root electrical bursts on either side of the spinal cord. This pattern has demonstrated the capability of rapidly activating 100 segments along the lamprey’s body to induce a highly coordinated swimming motion (Wallen and Williams 1984; Grillner et al. 1995). Moreover, mechanoceptors, known as edge cells, have been found to line the lamprey spinal cord and because of their sensitivity to small amounts of stretch can entrain the rhythmic motion and alter its frequency and timing (McClellan and Sigvardt 1988). The edge cells consist of two types of neurons: excitatory and inhibitory. When the notochord/spinal cord is bent, the cells inhibit contralateral side movement, while exciting ipsilateral movement, and vice versa. This activity is thought to propagate movement via several smaller oscillating units along the spinal cord (Grillner et al. 1995). When divided into pieces, the lamprey spinal cord has been shown to produce rhythmic activity in each piece separately, suggestive of pattern generating networks throughout the length of the spinal cord for continual generation of movement (Grillner et al. 1995).
In addition to the lamprey, both kittens and adult cats have been used as quadrupedal models of locomotion to demonstrate the existence of SPGs. In a two-study series, Forssberg et al. sought to determine how well the cat spinal cord could produce stereotypical, rhythmic stepping behavior (Forssberg et al. 1980a) and to what extent the hindlimbs could coordinate with one another (Forssberg et al. 1980b). Kittens spinalized between T10 and T12 initially exhibited alternating hindlimb patterns when the trunk was rotated to one side while lying on the floor. For some kittens, this pattern developed into the ability to reciprocally step while maintaining weight support on the treadmill. Despite deficits in equilibrium, muscle tone and step asymmetries (e.g. limping), distinct intralimb flexor (F) and extensor (E1, E2, and E3) phases were identified in the step cycles with EMG activity resembling that of intact cats (Forssberg et al. 1980a). Furthermore, increases in treadmill speed were shown to produce concomitant increases in hindlimb speeds through shortening of the extensor phases and to a lesser degree the flexor phases. Eventually, faster speeds altered the interlimb walking patterns from out-of-phase to in-phase manifested as galloping movements (Forssberg et al. 1980b). In split-belt treadmill situations where one belt was driven at a slow speed while the other was driven at a fast speed, the spinalized kittens also could immediately change the duration of the support and swing phases in each limb of the step cycle to stabilize the rhythm and maintain interlimb coordination. If the stepping was purely a product of belt speed, each limb would have continued to step at frequencies consistent with the belt speed. Based on these results of interlimb coordination, Forssberg and colleagues asserted that two separate, but interconnected spinal generators modified by peripheral feedback were accountable for the stepping
patterns observed (Forssberg et al. 1980b). This theory had previously been outlined based on cat fictive locomotion by Grillner and Zangger and later was reiterated by Howland et al. in T12 spinalized, neonatal kittens that developed a hindlimb stepping pattern on the treadmill (Grillner and Zangger 1979; Howland et al. 1995).

In contrast to both fictive and treadmill locomotion, another mode of detailing the presence of SPGs is through airstepping in spinalized animals. Airstepping eliminates limb loading or drive from treadmill motion and avoids the impact of chemicals sometimes used to induce stepping movements. Although initially requiring stimuli such as tailpinching to produce basic reflexes and airstepping, cats with complete low thoracic spinal lesions have shown the onset of a couple airstepping cycles immediately upon vertical lifting three weeks post-transection. At five weeks, this pattern developed into sustained, spontaneous airstepping with strong, appropriately-timed flexor and extensor EMG bursts consistent with both treadmill and fictive locomotion (Giuliani and Smith 1985). The airstepping model has been suggested to demonstrate better the “natural” behaviors of pattern generating circuitry.

In addition to the above animal models, motor behaviors in bipedal chicks have provided evidence for SPGs as well. Bekoff and colleagues (1987, 1989) have examined hatching and walking in post-hatchling chicks before and after lumbosacral deafferentation alone, cervical spinal transection alone, and both procedures together. In normal chicks, walking possessed cyclical, alternating interlimb coordination; whereas, hatching showed an episodic, synchronous coordination that was distinctly different from walking. However, after deafferentation, these motor patterns demonstrated a convergence suggesting that sensory input is the major modifier of the
same, or at least components of the same, SPG circuitry responsible for these individual behaviors (Bekoff and Sabichi 1987). Moreover, after C3 spinal transection alone, chicks again showed separate hatching and walking characteristics that resembled normal patterns (Bekoff et al. 1989). Furthermore, once the high cervical transection was combined with deafferentation, walking exhibited similar alternating patterns to those seen in only deafferented chicks. These findings suggested that the spinal circuitry did not require supraspinal or afferent inputs to produce the cyclical, rhythmic output. However, authors could not dismiss the possibility that lower cervical or thoracic proprioceptors may modulate activity of the SPGs in the absence of other ascending or descending inputs.

**Evidence from Human Literature**

As mentioned earlier, fictive locomotion provides potentially the strongest evidence for SPGs in animals. However, because no equivalent of this methodology exists for humans (i.e. anatomically complete lesions with deafferentation), questions still emerge as to whether the spinal circuitry that we detect in humans is the same as that in animals (Illis 1995; Duysens and Van de Crommert 1998; Dietz and Harkema 2004). Regardless, evidence from both training and spinal cord stimulation studies in humans is mounting to suggest strongly that SPGs are present in humans (Dietz and Harkema 2004; Dimitrijevic et al. 2005).

A hallmark study claiming to be “the first well-defined example of a central rhythm generator for stepping in humans” despite incompleteness of SCI was conducted by Calancie et al. (Calancie et al. 1994). This case report of an adult male, 17 years post-traumatic, incomplete, cervical SCI described the onset of involuntary, rhythmic, alternating lower extremity movements while in a supine position with hips and knees
extended. This pattern began with the initiation of an intense ambulation training program. Of note were the sensory inputs which increased the frequency of the pattern (e.g. hip/knee/cervical extension, toe dorsiflexion) and those which decreased or ceased the movement (e.g. hip/knee/cervical flexion, toe plantar flexion, involuntary bladder emptying, complete vertical unloading in harness). Additionally, anaesthesia to the right hip which upon x-ray was subluxed and sclerotic, attenuated the response. Furthermore, because of the predictability of electromyographic (EMG) activity and responses to sensory inputs indicating similarities with fictive locomotion studies in cats, the authors suggested that comparable SPGs were responsible for both. Since the Calancie et al. report, several other studies in humans have concurred that sensory input such as body loading and hip position influence locomotor pattern output via modulation of SPGs (Harkema et al. 1997; Dietz et al. 2002).

Perhaps the most convincing, and by some researchers described as direct (Dimitrijevic et al. 2005), evidence for human SPGs involves the epidural electrical stimulation of the dorsal spinal cord in individuals with complete SCI (Dimitrijevic et al. 1998; Minassian et al. 2004; Minassian et al. 2007). Completeness of injury for such studies has been defined based on the Brain Motor Control Assessment as absence of suprasegmental motor unit activation below the level of lesion (Dimitrijevic et al. 1998). Because of the approximation of this definition to an anatomically complete SCI relative to a clinically complete or discomplete SCI (Sherwood et al. 1992), humans with this type of injury are the best available models to compare to isolated spinal cord animal works. Tonic electrical stimulation of 25 to 60 Hz and 5-9 V via electrodes to the L2 epidural space produced a locomotor-like pattern with synchronized EMG activity in
these individuals (Dimitrijevic et al. 1998; Minassian et al. 2007). This optimal range of stimulation was based on experimentation of different stimulation parameters. Levels outside this frequency and amplitude range induced tonic extensor activity in the lower limbs. These optimal parameters demonstrated that the interneuronal network in the lumbar spinal cord has the capability of utilizing tonic trains of electrical stimuli to initiate oscillatory output. In turn, the rhythmic motor output generates patterned afferent feedback for the network to continue producing stepping.

Support for SPGs in humans and additionally that each limb has independent generators which cross-coordinate to produce optimal walking patterns under different conditions has been shown in split-belt treadmill paradigms, similar to the scenario discussed earlier in spinal kittens (Forssberg et al. 1980b). However, because of the inherent challenges presented by doing such a paradigm in persons with complete SCI, a healthy infant model has been used secondary to the limited cortical influence over stepping (Yang et al. 2005). When split treadmills are driven at different belt speeds or in opposite directions, infants have exhibited the ability to adopt a coordinated walking pattern. Each limb is thought to have an autonomous SPG because of the ability to modify stance phase durations and move legs in opposite directions concurrently, but SPGs on either side are believed to communicate in order to ensure only one limb is in swing at any given moment. These findings and conclusions are in agreement with literature in spinal cats and fictive locomotion in cats suggestive of SPG presence (Grillner and Zangger 1979; Forssberg et al. 1980b).
Spinal Cord Injury Rehabilitation

Traditional Rehabilitation of Walking

Until recently with the emergence of evidence supporting properties of SPGs in humans, rehabilitation of walking for individuals post-SCI reflected the hierarchical model of motor control (Shumway-Cook and Woollacott 2001). This theoretical model purported that the CNS was hardwired, irreparable, and non-malleable. Moreover, the spinal cord simply was nothing more than a conduit for information from the brain to the rest of the body. If an injury occurred to the spinal cord, the relay of information was suspended. Rehabilitation strategies for SCI emphasized compensation for the physical losses presumed to never return function, consistent with assumptions of the hierarchical model. In many cases, particularly those in which patients exhibited little to no voluntary muscle activity, relearning standing or walking was deemed impossible. Additionally, the common symptoms of spasticity and clonus were considered negative consequences of SCI, further limiting the probability of walking (Beres-Jones et al. 2003). Thus, therapists avoided any treatment regarding standing or walking and solely taught wheelchair propulsion, transfers, and other functional mobility skills. If standing or walking ever were attempted, therapists instructed patients in the use of assistive devices, supportive bracing, and alternative biomechanical strategies such as utilizing momentum to initiate movement or creating a wide base of support with the legs to increase stability. These strategies would assist the individuals in resuming some level of mobility using remaining physical strengths and external aids in light of the presumed irreparable spinal cord and inability to convey information from the brain to the muscles (Atrice 2005; Behrman and Harkema 2007).
Activity-Dependent Plasticity

Over recent years, evidence of activity-dependent plasticity in animals and humans has suggested a strong potential for the resumption of walking ability after SCI if the training conditions are optimized, thus providing the basis for rehabilitation strategies alternative to the traditional approach. Activity-dependent plasticity can be defined as persistent changes within the CNS that result from prior experiences and influence future motor behaviors (Wolpaw and Tennissen 2001). In essence, the CNS is exhibiting motor learning as a function of experience. Although activity-dependent plasticity traditionally is deemed to occur supraspinally (Kleim and Jones 2008), a great deal of animal and human literature also supports plasticity in the spinal cord due to peripheral inputs and/or descending influences from the brain (Barbeau and Rossignol 1987; Behrman et al. 2008). Repetition of specific types of sensory input as well as variability in activity intensity and challenging function through increasing postural and locomotor demands are essential to modify spinal cord neuronal structure and synapses and to induce the desired functional recovery after SCI (Barbeau 2003).

Evidence from Animal Literature

Research in adult cats and kittens with SCI have illustrated some of the most compelling evidence of locomotor recoveries consistent with activity-dependent plasticity. Through the use of locomotor training in the treadmill environment with appropriate sensory inputs provided by speed of the treadmill, bodyweight loading, and manual cues, spinalized cats have shown improved hindlimb stepping abilities over varying times periods post-transection (Lovely et al. 1986; Barbeau and Rossignol 1987; Hodgson et al. 1994). This is in contrast to work that has demonstrated poor,
uncoordinated stepping performance and difficulty maintaining weight support after two months training in mid-thoracic spinalized cats (Eidelberg et al. 1980).

Although plasticity is considered to be greater in younger, developing nervous systems as in the kitten (Smith et al. 1982), Barbeau and Rossignol demonstrated that appropriate training strategies can produce stepping recovery in adult cats as well (Barbeau and Rossignol 1987). Cats spinalized between T10 and T12 were trained in the treadmill environment two to three times per week. With use of a thoracic jacket and/or manual assistance at the tail, bodyweight support was gradually increased and treadmill speeds were varied according to the cats’ progress. Over periods of several weeks up to one year of training, deficits in balance and volitional movement remained. However, cats remarkably exhibited the ability to maintain hindquarter weight support and to generate rhythmic, coordinated stepping with overall similar joint excursions and EMG activity compared to both spinal kittens and intact cats.

As shown in Barbeau and Rossignol’s work (1987), providing correct afferent input can result in improved stepping performance. Additionally, one criterion associated with sensory input for activity-dependent plasticity to occur is task-specificity (Barbeau and Fung 2001). That is, learning of a particular task such as walking is observed when that task is specifically trained. However, walking training may not translate to the performance of other tasks such as standing, presumably because the sensory experience is different between the two tasks. Kang and Dingwell (2006) presented evidence consistent with this notion, identifying that the mechanisms controlling walking and standing balance are inherently different based on local dynamic stability analyses (Kang and Dingwell 2006). Therefore, rather than training standing tasks as a method
for improving walking, the repetitive afferent input from walking itself is thought to entrain that particular task into the CNS via neuroplastic changes. For example, the abilities of T12-13 spinalized cats randomized to either a non-trained, treadmill-trained, or stand-trained group one month post-transection have been explored (Hodgson et al. 1994). Following two to three months of training, all stand-trained cats could maintain standing for an extended time period with very little stimulation at the tail; however, they were unable to step with the exception of a few cats taking uncoordinated steps at extremely slow speeds under 0.2 m/s. In contrast, the step-trained cats required maximal stimulation at their tails to maintain standing but could step at an average of 0.62 m/s. Follow-up experiments initiated the stand or step training just one week after transection, continued for six to eight months, then crossed cats to the other training regimen. After the cross-over, the cats learned to perform the newest task in which they were trained, but they deteriorated in the ability to perform the original task; that is, if they originally learned to step before switching to stand training, then later they had difficulty stepping, but could stand well (Hodgson et al. 1994).

Furthermore, activity-dependent plasticity has been shown through comparisons of spontaneous recovery and recovery with locomotor training (Lovely et al. 1986; de Leon et al. 1998). In a sample of spinalized cats like those described by Hodgson et al. (1994), Lovely et al. randomly assigned cats to either a non-trained group or a treadmill-trained group. Five to seven months later, cats in the trained group exhibited a steeper rate of improvement in speed and achieved a significantly greater maximal treadmill speed with full weight-bearing steps compared to the non-trained group (Lovely et al. 1986). Subsequently, de Leon et al. found comparable results in a similar study, but
cats started training after one week and continued for three months. Kinematic and EMG characteristics were quantified and compared between groups and to activity before spinalization (de Leon et al. 1998). Several changes occurred consistent with greater stepping and speed-related improvements in the trained cats. For example, increased terminal hip extension in stance, higher vertical ankle displacement during swing, greater forward placement of the paw at initial touchdown, and increased tibialis anterior and iliopsoas EMG amplitudes all were noted.

In contrast to studies in the cat with complete SCI, evidence to support spinal cord plasticity with repetitive training has been demonstrated in the incomplete SCI chick model also (Muir and Steeves 1995). Since walking and swimming produce similar rhythmic, alternating movements, but differ in the amount of phasic sensory feedback available, recovery of these behaviors were evaluated in hatchling chicks following thoracic hemisections. After two weeks of both locomotor training on a runway and swim training, walking recovered to normal, but swimming motion remained poor relative to normal values. However, when phasic cutaneous input from a buoy was provided to the chicks as needed during limb extension (in the absence of a loading stimulus as in walking), swimming improved after five days; yet removal of the stimulus caused deterioration of the movement. By two weeks, however, even without the cutaneous input, swimming returned to normal (Muir and Steeves 1995). Therefore, task-specificity, by way of repeated sensory inputs, which are timed appropriately within the task, is thought to promote activity-dependent plasticity and locomotor recovery after SCI.
Finally, activity has illustrated the capacity to induce changes at the molecular level in animals as well (Vaynman and Gomez-Pinilla 2005). Exercise in rodents with hemisectioned spinal cords has revealed the upregulation of certain neurotrophins as activity increases. Specifically, brain-derived neurotrophic factor (BDNF) has shown elevated levels on the same side as the lesioned lumbar spinal cord. Additionally, activity has promoted expression of BDNF products in the CNS, such as synapsin I, involved in synaptic transmission and neurotransmitter release, as well as cyclic adenosine monophosphate (AMP) response element-binding protein (CREB), involved in gene transcription. These findings have suggested that the spinal cord has the potential to learn via rehabilitation post-SCI since these neural factors are essential components of synaptic plasticity for memory and learning (Vaynman and Gomez-Pinilla 2005).

**Evidence from Human Literature**

Based on knowledge gained from animal literature, human studies in SCI evolved using the training fundamentals that revealed positive functional changes in animals. Specifically, the concept of activity-dependent plasticity stimulated the implementation of activity-based therapies in humans (Behrman and Harkema 2007). Activity-based therapies utilize the principle of task-specificity to enhance the neurophysiological changes that promote functional gains. The goal is to retrain the nervous system below the level of the lesion and take advantage of any spared neural pathways that may interact with specialized circuitries above and below the lesion.

Evidence from the literature in humans has included both individuals with complete and incomplete SCI, cervical, thoracic and lumbar levels, adults and children. Participants have engaged in various locomotor training strategies from manual-
assisted to robotic-assisted training, strengthening exercises to functional electrical stimulation (Behrman and Harkema 2000; Field-Fote et al. 2005; Wirz et al. 2005; Gregory et al. 2007). However, to remain consistent with the definition of activity-dependent plasticity, only those interventions task-specific to walking are described here.

In persons with acute and chronic incomplete SCI of numerous etiologies and ability levels, four to twenty weeks of manual locomotor training were conducted. This training resulted in the majority of initially wheelchair-bound persons becoming able to walk without physical assistance. Furthermore, persons who were able to walk before training improved gait speed and endurance. Overall, these changes were maintained at follow-up, six months to six and half years later (Wernig et al. 1999). Such persistent changes in function indirectly corroborate activity-dependent plasticity in the CNS given the retention of training effects.

Similar findings have been described after manual locomotor training with translation to overground walking in persons with incomplete SCI (Behrman and Harkema 2000; Behrman et al. 2005). An adult who was non-ambulatory and initially had minimal voluntary activation in the lower extremities progressed to ambulating full-time with a straight cane and greatly improved volitional leg strength. Other individuals who were ambulatory with assistive devices prior to training advanced to walking with less restrictive devices and at faster gait speeds. Although these persons were all within a year of injury onset and spontaneous recovery may have influenced the outcomes, certain features of the training responses suggested that neural plasticity based on the task-specific rehabilitation strategy was at least partly responsible. For example, the
individual who had little voluntary leg movement prior to training demonstrated the ability to reciprocally step on the treadmill before any improvements in voluntary muscle control were observed. For other individuals who were able to walk before enrolling in training, improvements were seen in step symmetry and kinematics on the treadmill that did not immediately translate to the overground situation (Behrman and Harkema 2000; Behrman et al. 2005).

Similar to the adult with incomplete SCI who gained both strength and walking abilities despite initially having little leg activity, a non-ambulatory child with a chronic, severe, incomplete SCI also recovered walking function. However, after developing the ability to step and walk full-time with a walker following 76 manual locomotor training sessions, this child demonstrated no changes in voluntary muscle activity during that time (lower extremity motor score remained 4/50) (Behrman et al. 2008). Moreover, although stepping was present, equilibrium reactions were absent alluding to spinal pattern generated activity with minimal descending control for balance strategies. Additionally, the rhythmical, sustained stepping pattern was absent in a supine position, presumably because of the removal of afferent input due to loading while upright (similar rationale to the previously mentioned chick model of Muir & Steeves, 1995). This last finding, in addition to the evolution of walking with minimal, volitional lower extremity movement, is highly suggestive of activity-dependent plasticity in this child’s CNS as well as supraspinal, descending activation.

As discussed in spinalized cats, persons with clinically complete SCI also have exhibited the capacity step in the treadmill environment, albeit without transfer to the overground (Dietz et al. 1994; Harkema et al. 1997; Behrman and Harkema 2000).
Reciprocal stepping has been achieved using the same sensory inputs as those employed for persons with incomplete injuries (e.g. progressively increasing load, accentuating hip extension and upright trunk, triggering tibialis anterior activation for swing, altering speeds). The reduction in amount of assistance required, from full manual contact throughout the gait cycle to assistance for foot placement only or even a few independent steps, has been suggestive of plasticity in the neuromuscular system below the level of the spinal cord lesion. Moreover, EMG patterns over the course of a manual locomotor training program in patients with complete paraplegia demonstrate appropriate timing of the lower extremity flexors and extensors with increasing gastrocnemius and decreasing tibialis anterior amplitudes (Dietz et al. 1994; Dietz et al. 1995). The rate of change in gastrocnemius activity directly related to changes in loading during the program as well. Dietz et al. contend that these changes are not only indicative of a lumbosacral spinal circuitry, but also that the circuitry is capable of learning in response to sensory input such as loading (Dietz et al. 1997).

In contrast to stepping ability, preliminary data in both individuals with clinically complete and incomplete SCI report an improved ability to stand after stand and step training (Harkema 2001). Despite apparent lack of supraspinal input, persons with complete injuries gradually assumed standing with less bodyweight over time. One person was reported to have stood with only 10% bodyweight support (BWS) for 45 seconds. In comparison, those individuals with incomplete injuries have demonstrated independent standing (i.e. without BWS) for several minutes post-training.

Paradigm Shift toward Recovery

Although the conventional approach to SCI rehabilitation remains widely used presently, both evidence of activity-dependent plasticity and SPGs are changing the
way scientists and rehabilitation professionals view the capabilities of the CNS and rehabilitation post-SCI. A paradigm shift is emerging that transitions our rehabilitation mindset from one of compensation toward one of recovery (Behrman et al. 2006). Moreover, the principles of motor learning, which emphasize task-specificity and repetition of activity consistent with the task to be relearned, encompass this shift toward recovery of function. Specific sensory inputs influence the plasticity of both the brain and spinal cord (Kleim and Jones 2008). Thus, if recovery of walking is desired, the task of walking must be performed with appropriate sensory cues to the CNS. These cues limit the use of compensatory tools like assistive devices or bracing. Rather, recovery of walking promotes loading through the lower extremities, minimizing load through the upper extremities, optimizing kinematics and kinetics (e.g. trunk position, foot trajectory, hip position), and walking at a speed consistent with normal walking (Behrman and Harkema 2007).

Barbeau et al. (1999) presented a model of functional walking recovery consistent with this paradigm shift in rehabilitation, which could aid in directing clinicians towards appropriate evaluation and treatment strategies for patients after SCI. Figure 1-2 depicts a progression of walking recovery as a function of changes in “control” and “capacity.” Control indicates a person’s potential to alter four variables during walking: 1) generation/absorption of energy at specific points of the gait cycle, 2) trajectory of the foot during swing phase, 3) support of one’s own body weight, and 4) balance of the upper body. Capacity refers to the ability to maximize those variables as needed in order to meet environmental demands or personal goals. Image 1 of Figure 1-2 presents an individual in a permissive environment of BWS and a treadmill, which
allows that person to accomplish and optimize the necessary subtasks of walking. This individual possesses both low control and capacity for functional walking as he/she is unable to perform any component of walking overground, even with assistance of a walker, cane or crutches. Images 2 and 3 demonstrate varying degrees of increases in capacity and control as the individuals transition to an overground environment; however, they are only able to perform walking within the constraints of assistive devices. In Image 2, the person with a walker may be able to take a longer or higher step, for example, representing greater capacity of that variable, but at the expense of upper body control and lower extremity loading. That is, this individual could be compensating and increasing weight bearing through the upper extremities to give the lower extremity the biomechanical advantage of creating a maximal change in foot trajectory. In contrast, the individual with a cane in Image 3 demonstrates greater control of the head, arms and trunk as well as weight bearing through the legs, although it may be at the expense of achieving a maximal step length or height to step over an obstacle. Finally, Image 4 shows an overall increase in both control and capacity as a person demonstrates the ability to perform all the requisite subtasks of walking and modulate them appropriately for adaptability in the real world (e.g. no device, uneven terrain, stairs, etc.).

During rehabilitation, the BWS and treadmill environment may allow for thorough evaluation and optimal training of the control and capacity of variables essential to walking recovery. Since the emphasis of such an environment is on functional recovery rather than compensation, an individual’s true ability can be discerned in the absence of assistive devices and thus maximized. Furthermore, the shift in rehabilitation focus
towards recovery of walking behavior may enable individuals to experience and learn more appropriate motor control strategies for balance and adaptability without external devices that provide upper body support and minimize body loading.

**Determinants of Balance Control**

Human balance control is innately challenging, particularly in comparison to quadrupedral balance control requirements. Quadrupeds are naturally stable during standing and walking with a horizontal trunk position, low center-of-mass, and a broad base of support (BoS) created by three to four limbs in contact with the ground at all times. In contrast, the human body is an intrinsically unsteady ensemble. This instability results from a large mass consisting of the head, arms and trunk being placed in an upright position approximately two-thirds of body height above the ground. Furthermore, the only support structures are one or two feet on the ground forming a small BoS (Winter 1995). When humans perform tasks more demanding than static standing, such as reaching, turning, walking, or running, these functional skills continually create forces that challenge the structural organization of the body. Therefore, this multisegmental system requires balance control mechanisms that are capable of responding to the demands imposed by movement dynamics (Figure 1-3).

Maintaining balance concerns the integration of two essential control components: *postural control and equilibrium control* (Massion & Woollacott 1996). Posture refers to the orientation of body segments relative to the direction of gravity. Thus, postural control requires the body to combat Earth’s gravitational pull to prevent collapsing and to remain upright. Complementary to postural control, equilibrium control is the stabilization needed to counteract the linear and angular accelerations attempting to unbalance various segments of the body. Self-initiated movements such as walking or
external perturbations such as tripping on an obstacle create these destabilizing accelerations. To avoid falling, the magnitude of equilibrium control will depend upon movement speed and the inertial mass of the moving segment. That is, the faster the movement or the heavier the moving segment, the larger accelerations will be. Consequently, greater equilibrium control will be required to stabilize the interconnected segments against such forces (Massion & Woollacott 1996).

Ultimately, both postural and equilibrium control are achieved by two primary means: proactive and reactive mechanisms (Patla and Prentice 1995). Proactive balance mechanisms utilize both visual cues and prediction to prevent falls during voluntary movements. Initially, vision allows a person to scan the environment from a distance and evaluate impending challenges. Then, predictive mechanisms incorporate knowledge of prior experiences and calculations of potential forces that are likely to act on the body in order to select, plan and execute anticipatory postural adjustments (Huxham et al. 2001; MacLellan and Patla 2006; Misiaszek 2006). These feedforward responses “set” postural muscles in preparation for the expected destabilization induced by a voluntary task (Cordo and Nashner 1982). However, when proactive mechanisms under- or overestimate postural preparations or alternatively when perturbations are unexpected, the CNS responds to sensory feedback and employs reactive balance mechanisms. Unlike proactive mechanisms that have time to utilize higher executive processing, initial reactive mechanisms must occur in such a rapid manner that cortical control is not possible (Tang et al. 1998; MacLellan and Patla 2006; Misiaszek 2006). In those cases, the primary feedback comes from somatosensory and vestibular sources.
since the response times of those systems are much greater than vision (Woollacott and Tang 1997).

In addition to the balance control mechanisms intrinsic to the individual, the task performed and the environment in which the task occurs both affect the strategies utilized in balance control (Figure 1-3) (Huxham et al. 2001). Specifically, the task and environment may differentially dictate the biomechanical configurations of the body as well as the degree of information processing required to accomplish balance control successfully. For example, as a movement becomes more intricate (e.g. quiet standing to running), changes occur in the relationship between body segments and the accelerations that need to be controlled. Similarly, walking on a level linoleum floor will adjust the body's kinematics and kinetics differently and to a lesser extent than negotiating gravel terrain or avoiding moving obstacles. Furthermore, the complexity and familiarity of the task and environment both influence the amount of information processing (e.g. attention, prediction) that is necessary. Shumway-Cook and Woollacott (2001) have presented a systems approach to balance control that illustrates this interaction of the individual, task and environment via a convergence of musculoskeletal and neural components (Figure 1-4). More specifically, musculoskeletal components include the relationships among the various body segments including range of motion, flexibility, and strength. In contrast, neural components encompass sensory and motor strategies, sensory systems, internal representations of the body (body schema), and higher executive influences on motor responses through anticipatory or adaptive strategies.
Neural Systems Underlying Balance Control

Due to the intricate interrelationships between neural and muscular systems to coordinate motor output for functional balance, identifying the underlying neural substrates responsible for responses of balance control is difficult. In both animals and humans, suggestions of these substrates have been deduced from behavioral outcome studies primarily. Literature using animal models has analyzed postural and equilibrium control in lampreys, rabbits, and cats, to name a few (Deliagina et al. 2008). Moreover, research in humans traditionally has examined patient populations who have disorders related to various neural structures in order to understand how the postural responses of individuals with an abnormally functioning structure differ from responses of healthy individuals (Morton and Bastian 2004; Jacobs and Horak 2007a). Based on this kind of work, the brain stem, cerebellum, basal ganglia, cerebral cortex, spinal cord, and several descending neural pathways all have been suggested as critical for balance control; however, a consensus on their definitive contributions remains to be determined. Therefore, the following provides an overview of the possible roles of these neural substrates on different aspects of static and dynamic balance control.

Brain Stem

Many of the implications that the brain stem plays a critical role in balance control have evolved from literature of animal models. Specific brainstem sites in the cat model, such as the subthalamic and mesencephalic locomotor regions as well as the ventral and dorsal tegmental fields, show functions that integrate posture and walking when stimulated (Mori et al. 1992). Additionally, evidence has demonstrated that chronic, T6 spinalized cats lack the capacity to produce the necessary automatic postural responses to perturbations while standing, even though they exhibited hindlimb weight
support (Macpherson and Fung 1999). The maintained weight support was thought to result from muscular stiffness and tonic extensor activity elicited from spinal reflex mechanisms. However, the brainstem (and cerebellum) was implicated as a site required for balance control because it receives and integrates visual, vestibular and somatosensory input from the periphery (Horak and Macpherson 1996). Similarly, Torres-Oviedo et al. extracted a set of muscular synergy patterns from both quiet stance and automatic postural responses in cats exposed to platform perturbations (Torres-Oviedo et al. 2006). The brainstem and cerebellum again were suggested as integration sites for task-specific sensory information which ultimately simplified and organized the motor output into functional synergy patterns. Furthermore, the brainstem is thought to contain already established synergy patterns, which are selected and modified based on incoming afferent input about the context of a situation (Jacobs and Horak 2007b).

**Cerebellum**

In addition to the above proposed roles of the cerebellum in conjunction with the brainstem, several other theories have been presented regarding its function. Early work by Nashner and Grimm found delayed and unorganized muscle responses to standing surface perturbations in patients with cerebellar disorders (1978). Based on these findings, the coordination of muscle synergies in postural responses was hypothesized as one major role of the cerebellum. However, more recent investigations have posited that the cerebellum is involved in predictive balance control primarily (Bastian 2006); that is, using prior experiences to shape motor output. Evidence from individuals with cerebellar damage supported this hypothesis, showing hypermetric postural responses to both unexpected and expected platform perturbations. Moreover, this patient group also demonstrated the inability to scale response magnitudes to
expected perturbations, regardless of having previous exposures (Horak and Diener 1994). Other studies also have noted deficits in scaling magnitude of automatic postural responses; however, they indicated more specifically that the cerebellum had greater involvement in gain control of response magnitudes to changes in conditions (Timmann and Horak 1997; Mummel et al. 1998).

**Basal Ganglia**

Research aiming to discover the basal ganglia's role in balance control has largely utilized patients with Parkinson's disease (PD) as models. Chong et al. gained support for their hypothesis that the basal ganglia is essential for the ability to quickly adapt postural set to a sudden change in task context (Chong et al. 2000). Although reactive muscle response onsets and synergy pattern organization were normal in persons with PD, multiple trials were required to switch postural set and select a different synergy pattern. In contrast, delayed muscle onsets and relative timing as well as decreased muscular amplitudes were shown in a voluntary task of rising on toes (Frank et al. 2000). Nevertheless, the scaling amplitude of postural responses to changes in velocity and excursion of movement was appropriately produced. These findings from the Frank et al. study and similar results from additional studies led to the conclusion that the basal ganglia contribute to the ability to initiate and generate quick, sufficient force to control center of mass movement in both voluntary and externally-cued perturbation tasks (Horak et al. 1996; Burleigh-Jacobs et al. 1997). Furthermore, functional imaging studies in animals and humans also have supported the role of the basal ganglia in activating necessary postural muscles while inhibiting unnecessary antagonists when preparing for and initiating responses to centrally-driven and externally-triggered movements (Brooks 2001). For example, when basal ganglia dysfunction is present, an
overall increase in background postural EMG (co-contraction or inability to inhibit antagonistic muscles) has been observed during quiet standing, manifesting itself as muscular rigidity (Horak et al. 1996).

**Cerebral Cortex**

Involvement of the cerebral cortex in balance control remains a debatable issue, particularly for reactive mechanisms. Some researchers have cited that postural response latencies are too short to gain cortical input, thereby deeming subcortical mechanisms responsible (Diener et al. 1984). However, reactive postural responses may actually last longer than the initial, automatic response mediated by brainstem or spinal mechanisms. As a result, the later phases involving feedback loops to correct for errors, likely require cortical influence (Jacobs and Horak 2007a). Later phases include compensatory strategies, such as stepping to form a wider base of support or reaching for external support, beyond the potentially ineffective ankle or hip strategies produced by short- or medium-latency responses. Moreover, cerebral cortex contributions have been observed via electroencephalography potentials in studies of anticipatory responses to expected perturbations and voluntary movement as well (Slobounov et al. 2005; Jacobs and Horak 2007a; Jacobs et al. 2008).

**Spinal Cord and Descending Pathways**

In addition to the initial reactive myotatic stretch reflex induced by standing perturbations, evidence asserts the ability of descending pathways to modify segmental spinal reflexes. Specifically, vestibulospinal and reticulospinal neuron recordings have shown increased responses in cats walking on an inclined treadmill angle (Matsuyama and Drew 2000). Reticulospinal drive also has revealed increased antigravity extensor tone during locomotor activities, likely due to the excitation of the mesencephalic
locomotor region (Mori 1987). Furthermore, enhanced corticospinal pathway activity has
been detected and well-correlated with cats’ balance corrections to standing platform
tilts (Beloozerova et al. 2005). Additionally, studies in both adult humans, infants, and
spinal cats have implicated spinal networks and reflexive pathways for the quick,
stumbling-corrective reaction that occurs in response to stimuli contacting specific
extremity locations (e.g. dorsum of paw/foot) and during certain phases of the gait cycle
(Forssberg et al. 1975; Forssberg 1979; Zehr et al. 1998; Lam et al. 2003). These
pathways, as well as the interaction of pathways linking the above mentioned major
neural structures, like the basal ganglia-brainstem system, have been suggested as
participants in the regulation of balance control (Drew et al. 2004; Takakusaki et al.
2004).

**Balance Control in SCI**

For individuals post-SCI who regain any degree of walking ability, the challenge
remains to identify which components in the triad of walking subtasks (stepping,
balance, and/or adaptability) continue to limit their return to complete independence.
Although difficult to determine secondary to the multitude of factors comprising each
subtask, Brotherton et al. noted that one of the top three perceived factors associated
with the high falls incidence in this population was loss of balance (Brotherton et al.
2007a). Anderson (2004) further suggested that balance is a limiting factor for this
population by indicating that recovery of trunk stability was an important quality of life
factor for persons with paraplegia, particularly those less than three years since injury.
As mentioned previously, human walking requires one to balance the large trunk mass
above a continually moving base-of-support (Winter 1990). Therefore, trunk instability
would create yet an additional challenge to accomplish the task of walking successfully.
Moreover, another study showed a significant correlation between balance and walking performance in SCI (Scivoletto et al. 2008). These authors recommended that balance become a highly emphasized component of walking rehabilitation in SCI, commenting that the SCI literature only discussed balance with respect to trunk activity in the wheelchair. A more recent search of the literature revealed that a few studies do exist which examine standing balance control in this population (Wydenkeller et al. 2006; Liechti et al. 2008; Thigpen et al. 2009). Although walking balance control in SCI awaits investigation, studies provide evidence of sensorimotor complications that may contribute to balance dysfunction during walking and further inhibit recovery of this task.

The systems approach to balance control by Shumway-Cook and Woollacott (2001) presented earlier (Figure 1-4) supports the possible impact of these complications on walking balance control, inclusive of the proactive and reactive strategies necessary for participation in the any real world environment. Additionally, Figure 1-5 models a bi-directional interaction of example sensory, motor, and postural deficits including balance following SCI, further illustrating the potential influence and interference of these problems on walking recovery (Barbeau et al. 1999).

As Barbeau et al. (1999) suggest, the various complications from SCI are not mutually exclusive and have the potential to impact the functioning of other biological systems. Noreau et al. (2000) presented the prevalence of secondary impairments resulting from SCI and highlighted the multiple body systems affected by these injuries that could disrupt function and quality of life. After surveying almost 500 individuals post-SCI of varying age, gender and injury severity, this investigation confirmed that in addition to nervous system involvement, the cardiovascular, respiratory,
musculoskeletal, intestinal, cutaneous/integumentary, and urinary/renal systems also were compromised (Noreau et al. 2000). Although some of these systems appear removed from the control of walking, many conceivably could disrupt this task and its subtask of balance, either directly or indirectly.

On a musculoskeletal level, primary complaints following SCI are atrophy and weakness in muscle groups that have diminished neural innervations secondary to the level of injury (Jayaraman et al. 2006; Shah et al. 2006). These deficits often include decrements in trunk muscle activation (e.g. paraspinals and/or abdominals), which are key muscles in trunk stabilization during unsupported functional tasks (Borghuis et al. 2008; Bjerkefors et al. 2009). Residual, isolated muscle strength using the AIS evaluation after acute injury reportedly strongly predicts motor recovery (Waters et al. 1994), although more recent evidence argues that changes in lower extremity motor scores on the American Spinal Injury Association Impairment Scale (AIS) do not always accompany changes in walking function, acutely or chronically (Wirz et al. 2005; Wirz et al. 2006; Behrman et al. 2008). Other skeletal muscle properties deemed necessary for skilled movement, such as the ability to alter timing and power of motor output, have been implicated as limiting factors in walking recovery. Specifically, persons with SCI demonstrate drastically reduced rates of torque production as well as voluntary peak torque in affected muscles such as the plantar flexors and quadriceps (Jayaraman et al. 2006; Gregory et al. 2007). In healthy individuals, energy generated from plantar flexor torque and power is transferred up the lower extremity to provide trunk support and forward progression during walking (Zajac et al. 2003). When disrupted, persons with
SCI may require alternative movement strategies to generate an upright walking pattern while fully-loaded through their lower extremities.

Thigpen et al. (2009) additionally showed that individuals with mild to moderate sensorimotor impairments (e.g. deficits in light touch, pin prick sensation, lower extremity strength, and proprioception) secondary to SCI are able to adapt automatic postural responses appropriately to expected and unexpected changes in standing perturbation conditions. However, muscle onset times are delayed and activation magnitudes are decreased compared to control subjects (Thigpen et al. 2009). Furthermore, persons with motor incomplete SCI display cocontraction of antagonistic lower extremity muscle groups (tibialis anterior/soleus, quadriceps/hamstrings) during treadmill walking with BWS (Gorassini et al. 2009). Some of these individuals also develop three to four times greater activation in proximal leg muscles compared to healthy individuals; presumably, this abnormal increase may be an effort to maintain postural stability in the absence of adequate strength in other muscles.

In the presence of lower extremity weakness, spasticity is speculated clinically to be a partial compensatory mechanism for loss of supraspinal drive by providing extensor support during stance post-SCI (Dietz 2002). Yet, Scivoletto et al. (2008) detected a strong negative relationship between spasticity and walking performance, promoting spasticity as one of the best negative predictors of walking ability. Reportedly, spasticity has potential to fragment walking motion, thus diminishing its smoothness. Spasticity develops as a sequela of disinhibition and hyperexcitation of monosynaptic stretch reflexes as well as loss of long-latency reflexes resulting from lack of feedback control (Dietz 2002). Proprioceptive feedback from joint movement, muscle
stretch, skin pressure or noxious stimuli provides information about the location of the body in space and contributes to the regulation of reflex activity. Often after SCI, proprioception is diminished or absent. When modulation of these reflex loops is reduced secondary to proprioceptive deficits, walking control is impaired. As a result, the nervous system relies more heavily on the vestibular system. However, evidence of compromised vestibulospinal tract integrity in samples of individuals post-motor incomplete SCI has emerged recently (Wydenkeller et al. 2006; Liechti et al. 2008).

While undergoing galvanic vestibular stimulation during different standing sensory conditions, persons with SCI demonstrate longer muscle and center-of-pressure movement onset latencies suggesting the vestibular system has difficulty substituting for sensory impairments and maintaining balance in the same manner a healthy individual’s vestibular system does.

Multiple impairments secondary to SCI exist, which may be contributing to balance dysfunction and the aforementioned high falls incidence in this population. These impairments create functional deficits that necessitate reliance on assistive devices for stability and mobility and further limit maneuverability in a variety of real world contexts (e.g. up and down stairs, inclines, uneven terrain) (Pepin et al. 2003a; Leroux et al. 2006; Musselman and Yang 2007). Instability post-SCI also imposes altered movement strategies where the trunk and arms may have to compensate for impairments in the legs (Melis et al. 1999). Such biomechanical compensations may further limit the physiological activity tolerance required for functional adaptations like altering walking speeds or walking for long distances in the community (Waters and Mulroy 1999; Jacobs and Mahoney 2002). Additionally, individuals post-SCI have reported several
secondary injuries due to loss of balance and falls, including fractures, joint dislocation, muscle/ligament strain or sprain, and loss of consciousness (Brotherton et al. 2007a), which could further postpone and limit walking recovery. Krause (2004) asserted that those individuals with less severe SCIs, classified as AIS D, are more likely to sustain subsequent injuries. Perhaps this is because they demonstrate some recovery of function and ability to ambulate, but may have difficulty with the task (Krause 2004). Thus, the study of balance control in the spinal cord injured population is essential to minimize risk of falls and maximize recovery potential for one’s ultimate return to functional community walking and participation.

**Outcome Measures of Walking Balance**

**Current Laboratory-Based Measures of Balance Control**

Balance control has been measured objectively in several ways utilizing different modes of technology in the laboratory setting. The reason for the development of various measurement strategies is that studies aim to investigate diverse components of a person’s ability to balance. Because of the abstract and vast nature of “balance” as well as the infinite number of everyday tasks available to explore, researchers often consider individual variables that contribute to balance as a whole rather than objectively measuring the construct of balance. For example, a researcher may be seeking to understand possible neural mechanisms involved in the production of balance strategies while walking or rather to investigate the ability of muscles to coordinate their timing appropriately in order to achieve those strategies. Alternatively, a study’s goal might be to determine which sensory systems are relied upon more or less for maintaining upright standing balance in specific patient populations. Therefore, measurement selections differ depending on whether one is investigating the neural,
biomechanical, or muscular mechanisms underlying balance control, and furthermore which type of task (standing versus walking) is of interest. A review of common objective balance measurements are provided here with the disclaimer and understanding that researchers frequently alter or combine information from basic measurements to fit their particular scientific inquiries. As a result, a large number of measurement variations exist.

One central, well-established concept in balance control that has pervaded the literature is based on the inverted pendulum model (Figure 1-6) (Hof 2008). This model assumes the body is a single mass located at the center of mass (CoM), supported by a leg that is in contact with the ground at the center of pressure (CoP). The whole body CoM is often considered the balance point of the body and is the weighted average of all the body segments’ individual centers of mass. In contrast, CoP is the weighted average of all the contact areas of the feet with the ground. When the CoM falls to one side of the CoP, the CoP must shift to prevent falling. Therefore, based on this model, a person’s CoM must remain within the CoP or base of support during standing or must transition along the trajectory of the swinging limb towards a new base of support in walking in order to remain stable. This relationship between the CoM and CoP has become the underlying theme in the development of many laboratory-based balance measurement tools.

Both the static and dynamic relationships of the whole body CoM to CoP have been used to examine balance in standing and walking tasks (MacKinnon and Winter 1993; Pai and Patton 1997; Hof et al. 2005). Biomechanical models from motion analysis data can be created to determine the CoM position and velocity. Subsequently,
CoM data may be compared to the CoP obtained from ground reaction forces on force plates. A static relationship which evaluates the horizontal distance between the vertical projection of the CoM location and the CoP remains a frequent measurement tool (Martin et al. 2002; Chou et al. 2004; Hughey and Fung 2005). As the distance between CoM and CoP becomes smaller, balance becomes compromised. However, given the dynamic nature of tasks such as walking where time and movement are important components, several researchers have recognized the contribution of CoM velocity in this relationship. Measures such as the margin of stability (aka. dynamic stability margin) have been proposed to account for velocity’s influence (Pai and Patton 1997; Hof et al. 2005; Hof 2008).

In addition to CoP and CoM relationships, location and variability in CoP patterns alone (Goldie et al. 1989; Maki and McIlroy 1996; Doyle et al. 2007) are measures commonly used in quiet and perturbed standing assessments. Single or dual force platforms are capable of providing CoP information averaged under both feet or for each foot individually. Within testing protocols, individuals are often instructed to stand statically with eyes opened or closed, on a compliant surface, and/or in single limb or tandem stance (Jonsson et al. 2004). The primary goal of these experimental manipulations is to induce sensory conflict, stressing the visual, vestibular, or somatosensory systems separately or in combination. The direction and amplitude of CoP change (i.e. postural sway) that results from ankle strategies typically, but also hip, stepping, and suspensory strategies, is measured. In a similar manner, CoP patterns have been recognized as a means to assess postural sway secondary to unilateral galvanic vestibular stimulation techniques as well (Latt et al. 2003). This information
provides details regarding how the vestibular system interacts with the other sensory systems involved in postural control. As a result, insight into the neural mechanisms of balance can be obtained.

In contrast to CoM and CoP quantities, electromyography (EMG) of the limb and paraspinal muscles is utilized to detect whether or not appropriate muscles are activated during tasks and the timing of their activation. Specifically, *muscular onset times and response latencies* have been used extensively to determine the initiation of reactive balance strategies after external perturbations are imposed to an individual while quietly standing or walking (Nashner 1980; Horak and Nashner 1986; Bakker et al. 2006). A great deal of the balance literature has adopted or adapted the classic perturbation paradigm developed by Nashner, which introduces unexpected platform rotations and translations (Nashner 1976). Muscle response times after the initiation of platform movement can provide information about the neuromuscular, coordinative mechanisms of a person’s response; that is, determining if a delay in response is present or if additional muscle groups are compensating for deficits in normally activated muscles. Similarly, muscle onset times can also offer insight into anticipatory postural strategies to internal perturbations in which muscles are activated in preparation for a task like initiating gait or reaching with the arms (Timmann and Horak 2001; Mochizuki et al. 2004). In such tasks, visual input as well as prior experience has instructed the central nervous system to “set” postural muscles in advance of a dynamic task to avoid an uncontrolled CoM displacement.

Additionally, EMG has been used to develop another measure for balance control. Applying non-negative matrix factorization techniques to EMG data, the specific set of
synergies responsible for a balance response to a perturbation task during standing can be identified. These synergies are interpreted as the way in which the nervous system simplifies the many degrees of freedom available in human movement and coordinates the motor output into stereotypical response patterns (Ting and Macpherson 2005).

Alternatively, kinematic changes primarily in the head, trunk, pelvis, hip, knee, or ankle angles have been measured via motion analysis to indicate normal and abnormal balance patterns during standing and walking (Krebs et al. 1992; Leroux et al. 2002; Nadeau et al. 2003; Grabiner et al. 2008). Angles of trunk or CoM inclination and angular dispersions or mediolateral linear excursions of head and trunk segments are examples of objective measures utilized (Hahn and Chou 2003; Nadeau et al. 2003; Lee and Chou 2006). Other temporal-spatial parameters derived from kinematics including step width (interfoot distance) and step velocity have likewise been considered indicators of balance control during gait (Timmann and Horak 2001; Krebs et al. 2002).

Unlike the above measures, non-linear measurements using kinematics have been devised over recent years for dynamic tasks specifically (Dingwell and Cusumano 2000; England and Granata 2007; Segal et al. 2008). These measurements, such as the maximum finite-time Lyapunov exponent, are calculated based on the rate of kinematic variability at various joints during a movement. These tools provide an indicator of “local dynamic stability” as a result of internal disturbances, such as neuromuscular errors within the body.

Finally, acceleration patterns of the upper body from the head to the pelvis have been measured via accelerometers or double differentiation of motion analysis position data to illustrate the way the postural control system coordinates and attenuates the
oscillations imposed by the dynamics of movement. Segmental coupling, peak amplitude, frequency, regularity, and smoothness of the patterns are all outcomes that have been described (Menz et al. 2003a; Kavanagh et al. 2004; Kavanagh et al. 2005a; Kavanagh et al. 2006). Given the influence of major sensory systems on balance control as well as studies in healthy individuals suggesting that the head is well stabilized during various locomotor tasks (Pozzo et al. 1990; Nadeau et al. 2003), acceleration may provide information indirectly about the ability of the body to gain visual stabilization and accurate vestibular inputs at various gait speeds and over uneven terrain.

**Limitations of Laboratory-Based Measures**

Measurements of balance gained within the laboratory setting provide extensive knowledge and generate further inquiries into the complexities of the balance control system. Nevertheless, limitations exist in these measures. First, the amount of time required to collect, process, and analyze data is substantially greater than the time allotted to conduct a therapeutic examination in clinical settings. Second, the equipment needed to obtain these measures is costly. When a major goal of creating laboratory measures is to eventually translate findings to the clinic for use with patient populations, time and expense must be factors considered (Allum and Carpenter 2005). Additionally, the currently developed measures necessitate integration with one another to achieve insight into the cause and effect of balance maintenance and loss through a host of contributing variables. Optimally, only a few measures would be able to offer that depth of information.
Current Clinical Measures of Balance Control

Unlike the highly technical measures acquired in laboratory settings, clinical measures of balance typically use little equipment and can be conducted and interpreted in a short period of time. With the exception of more specialized balance clinics, which utilize Computerized Dynamic Posturography systems such as the Balance Master® or Equitest®, the objective clinical measures most frequently used in the clinic are standardized assessments based on observational analysis. Numerous measures have been developed over the years with several redundant balance items across tests (e.g. inclusion of the Functional Reach in the Berg Balance Scale). As a result, only the most commonly used assessments in the clinic currently with balance as the principal construct are presented here: Berg Balance Scale, Dynamic Gait Index, Tinetti Performance-Oriented Mobility Assessment, and the Clinical Test of Sensory Integration and Balance (Huxham et al. 2001). Additionally, the most recently developed and still evolving balance tool, the Balance Evaluation Systems Test, which aims to be a comprehensive evaluation of balance will be described (Horak et al. 2009).

The Berg Balance Scale (BBS) consists of 14 items, each scored on a scale of 0 to 4 (Berg et al. 1989). Scores are based on the amount of assistance required to complete the tasks as well as ability to maintain positions for specified time periods or ability to perform a task within a time constraint. All items on the scale prohibit the use of any assistive devices. Tasks range from static sitting and transferring over level surfaces to a functional reach test, turning in a circle, and picking up an object from the floor. The BBS has been shown to have excellent interrater (intraclass correlation coefficient (ICC)=.97) and intrarater (ICC=.98) reliability in individuals with acute stroke.
and the elderly (Berg et al. 1995) as well as high test-retest (ICC=.96) and interrater reliability (ICC=.96) in persons with multiple sclerosis (Cattaneo et al. 2007).

In contrast to the BBS, the Dynamic Gait Index (DGI) assesses an individual’s ability to perform eight gait-related tasks with his/her usual assistive device(s) as needed (Shumway-Cook and Woollacott 2001). Each item on the DGI is scored from 0 to 3, ranging from severe impairment to normal. Differences in scores are based on use of an assistive device, physical assistance needed, imbalance, gait deviations and/or time required to complete a task. Examples of tasks on this assessment include changing gait speeds, stepping over an obstacle, gait with vertical head turns, and stairs. In persons with chronic stroke, the DGI exhibits high test-retest and interrater reliability (ICC=.96, .96 respectively) and concurrent construct validity with the BBS (r=.83) (Jonsdottir and Cattaneo 2007). High reliability has also been found for persons with vestibular dysfunction (kappa=.95) (Wrisley et al. 2003) and multiple sclerosis (ICC=.98) (McConvey and Bennett 2005).

Next, the Tinetti Performance-Oriented Mobility Assessment (POMA), originally designed to predict falls in an institutionalized population, has two subscales: a gait assessment (G-POMA) and a balance assessment (B-POMA) (Tinetti 1986; Tinetti et al. 1986). G-POMA instructs an individual to walk at a preferred pace with an assistive device as needed, while nine qualities of performance are evaluated as 1 (normal) or 2 (abnormal/compensatory). Performance qualities observed include initiation of gait, step length, path deviation, and trunk stability. In contrast, B-POMA consists of 13 tasks that are scored as 1 (normal), 2 (adaptive response), and 3 (abnormal). Although tasks are similar to those on the BBS, a reactive balance task in which a nudge is applied to the
sternum while standing is also assessed. Both subscales have demonstrated good interrater (R=.80 to .93) and test-retest reliability (R=.72-.86) in a sample of elderly participants (Faber et al. 2006).

Although not measured along an ordinal scale as with the above measures, the *Clinical Test of Sensory Integration and Balance* (CTSIB) developed by Shumway-Cook and Horak is frequently administered in the clinic (Shumway-Cook and Horak 1986). The CTSIB tests a person’s ability to maintain static standing for 30 seconds while compensating for altered sensory inputs in six different conditions. The conditions are 1) eyes open while standing on a firm surface, 2) eyes closed on a firm surface, 3) eyes open with a visual conflict dome on a firm surface, 4) eyes open on a compliant surface, 5) eyes closed on a compliant surface, and 6) visual conflict on a compliant surface. Maintenance of normal balance, increased postural sway, or loss of balance in each condition indicates which sensory system (visual, somatosensory or vestibular) a person relies upon primarily or more so than other systems.

Finally, the *Balance Evaluation Systems Test* (BESTest) was developed recently in an attempt to identify the underlying systems contributing to balance control deficits and emphasize the importance of the systems approach to motor control evaluation (Figure 1-4) (Horak et al. 2009). Unlike other tools, which were created for determining balance dysfunction in older adults in particular, this assessment aims to more broadly apply to a variety of patient populations in order to approach rehabilitation most effectively. The BESTest assesses six balance control systems (biomechanical constraints, stability limits/verticality, anticipatory postural adjustments, postural responses, sensory orientation, and stability in gait) across 36 test items. Some items
are newly created if unavailable on current tests and some items are borrowed from already published standardized tests (e.g. BBS, DGI, CTSIB) Although still in development, this tool thus far demonstrates excellent interrater reliability overall (ICC=.91).

**Limitations of Clinical Measures**

Therapists select balance measures with the intention of revealing a patient’s risk of falls and areas of balance deficits and of guiding a plan for treatment. Additionally, these measures are used to assess progression over time and recovery after injury, following surgery, or from a disease process. “Recovery” as described here is defined as Levin et al. (2009) proposed: “restoring the ability to perform a movement in the same manner as it was performed before injury.” However, the current measures of balance present limitations for therapists in accurately obtaining this information.

Although most falls occur during walking tasks (Berg et al. 1997), a limited number of measures assess balance during walking. Moreover, the assessments that have been developed allow an individual to perform tasks with the use of an assistive device such as walkers, canes or crutches (e.g. DGI, BESTest: Stability of Gait). Given the literature emphasizing critical differences between recovery of walking function versus functional compensation (e.g. with assistive devices) (Behman and Harkema 2007) and research showing altered muscle activation or balance strategies with upper extremity support (Visintin and Barbeau 1994) or even light surface contact (Jeka and Lackner 1994), our current assessments for walking balance appear to place emphasis on compensatory balance only. Obtaining information regarding compensatory walking balance may be misguiding therapists as to the true underlying balance deficits in their
patients, thus preventing implementation of appropriate treatment strategies for progression toward recovery of function.

Furthermore, the balance measures that do emphasize recovery by eliminating use of assistive devices (e.g. BBS) neglect assessment of walking and primarily examine sitting, standing, and transitional movements like transfers. However, with the principle of task-specificity for motor learning infused throughout the literature (Lovely et al. 1986; Barbeau 2003), if information regarding walking balance is needed, then walking should be the task assessed. Nevertheless, current measures like the BBS still are used to globally assess fall risk and balance for all tasks secondary to strong psychometric properties and lack of other appropriate measures (Patterson et al. 2007; van de Port et al. 2008).

Finally, balance is comprised of many sub-components of control via proactive and reactive mechanisms. Yet, current clinical balance measures primarily test proactive strategies and minimally assess reactive, which occur quite frequently in everyday life (Huxham et al. 2001). Although individuals often have to motor plan in advance how they are going to approach certain obstacles, they also need to respond suddenly to unexpected perturbations on a regular basis. Laboratory investigations are capable of examining factors such as these to provide a more in-depth understanding of the construct of balance for eventual translation to the clinic.
Figure 1-1. Modulation of central/spinal pattern generator (CPG) circuitry in (A) healthy and (B) complete spinal cord injured nervous systems through afferent inputs, descending neural pathways, and/or therapeutic interventions. Adapted from Edgerton VR, Tillakaratne NJ, Bigbee AJ, de Leon RD, Roy RR (2004) Plasticity of the spinal neural circuitry after injury. Annu Rev Neurosci 27: 145-167.
Figure 1-4. Interaction of the multiple systems contributing to balance control. Adapted from Shumway-Cook A, Woollacott M (2001) Motor Control: Theory and Practical Applications. Lippincott Williams and Wilkins, Philadelphia.
Figure 1-6. Diagram of the inverted pendulum model. $mg = \text{mass} \times \text{gravity}$, $l = \text{length}$, $\text{CoM} = \text{center of mass}$, $\text{CoM}'(x) = \text{vertical projection of center of mass onto ground}$, $\text{CoP}(u_x) = \text{center of pressure}$, $-F_y = \text{vertical ground reaction force at CoP}$. Adapted from Hof AL (2008) The 'extrapolated center of mass' concept suggests a simple control of balance in walking. Hum Mov Sci 27: 112-125.
CHAPTER 2
CLINICAL RELEVANCE AND RATIONALE

Of the continuum of individuals who sustain a SCI, approximately 52 percent are diagnosed with incomplete injuries at the time of hospital discharge with a greater potential to regain walking ability than those individuals with complete injuries (National Spinal Cord Injury Statistical Center 2008). A smaller subset of persons with motor incomplete injuries regains some degree of walking function with or without assistance of an assistive device and/or brace. Of those who demonstrate any walking ability, an even smaller, unknown percentage is capable of walking without a device or bodyweight support (BWS) and avoiding falling simultaneously (Figure 2-1). This cohort exhibits a level of neural compensation and functional balance recovery (Levin et al. 2009) post-injury that others are not able to achieve. Even more importantly, through their functional recovery, we may begin to understand the way they accomplish this task and mechanisms underlying such recovery. Ultimately, this knowledge will allow researchers and clinicians the opportunity to determine how to progress others with SCI towards functional balance and community ambulation.

As mentioned in previous sections, current measures of balance possess limitations. Specifically, when clinicians wish to evaluate balance control for the task of walking post-SCI, standardized tests that use assistive devices such as the Dynamic Gait Index, mask a person’s actual capacity to maintain their posture and equilibrium secondary to compensation through the upper extremities. In addition to measures of balance, the primary clinical prognostic indicator for overall walking potential, which globally includes the ability to balance as a pre-requisite, is the American Spinal Injury Association (ASIA) Impairment Scale (AIS) (Waters et al. 1994). Through an evaluation
of voluntary, isolated, limb strength and sensory integrity, persons who regain some to all isolated movement have injuries classified into three main categories: AIS C through E. However, given the anatomical, physiological, and functional heterogeneity present in persons with SCI, the ability to maintain balance during walking varies greatly within each category. Thus, clinicians still await adequate tools to detect the underlying balance control that reflects task-specific functional recovery rather than compensation. In order to develop appropriate clinical tools, laboratory investigations are critical to understand the construct of “balance” and all its relevant neural, muscular, and biomechanical components. This will enable the determination of clinical correlates to laboratory-based measures such that efficient and true assessments of balance can occur in the clinical environment.

With the intention of measuring task-specific functional recovery, the studies in the following chapters were developed within the framework in Figure 2-2. This framework illustrates a linear progression through the paradigm shift occurring in neurological rehabilitation today: one that started as a “compensatory” approach and is now shifting toward a “recovery-based” approach. The first study begins with an examination of dynamic stability in the context of traditional rehabilitation (i.e. with assistive devices). In the second study, that layer of compensation with external devices is removed to evaluate stability during a functionally uncompensated walking experience. Finally, the third study remains within the paradigm shift to investigate the effects of two recovery-based locomotor training interventions on dynamic stability.

A dynamic stability measurement model also is presented in Figure 2-2 to demonstrate various avenues in which to evaluate the framework above. These
measurement strategies elucidate the interplay of body segment dynamics along the entire body axis from the head to the interaction of the feet with the ground. The rationale for this “top-down, bottom-up” model of dynamic stability is to illustrate that the human body is series of interconnected elements that all possess a role in balance control. According to the systems approach to balance control, these elements unite on a multiple system level when considering the intersegmental biomechanics as well as the many sensory inputs, their integration, and motor outputs along the neural axis (Horak et al. 2009). However, the studies presented here represent the first time a biomechanical investigation has evaluated several elements along the entire body in each person as well as conducted assessments during walking conditions without assistance of any type in individuals who regularly walk with assistive devices. The specific measurement tools selected capture what are purported to be critical components or priorities for maintaining balance while walking. Motor strategies observed in persons with SCI across these studies may be able to provide insights into potentially affected balance control systems for examination in future research as well as the impact of recovery-based locomotor interventions on dynamic stability of individuals with particular motor strategies. One motor strategy (e.g. head stabilization, foot placement to capture the moving center-of-mass) may compensate for insufficiencies in other typically “normal” strategies or locomotor training may influence the development of certain motor strategies (e.g. control of center-of-mass, trunk movement, spatial parameters). Details regarding specific measurement tools mentioned in the developed model are described in Chapter 3, “Methodology: Procedures and Considerations.”
Figure 2-1. Theoretical continuum of walking ability in individuals post-SCI.
Figure 2-2. Dynamic stability measurement framework.
Figure 2-3. Rationale and progression of three experiments.

**Experiment #1:** Examine traditional rehabilitation approach to dynamic stability using devices

**Experiment #2:** Evaluate dynamic stability during walking without devices

**Experiment #3:** Compare effects of two recovery-based locomotor interventions on dynamic stability
CHAPTER 3
METHODOLOGY: PROCEDURES AND CONSIDERATIONS

Participants

Ten individuals with chronic, incomplete SCI and ten healthy persons comprising a control group participated in each study. Those with SCI were involved in a larger randomized clinical trial (RCT) and healthy controls were included as part of a larger, ongoing cross-sectional study. By including the same individuals across studies, findings from each could be assimilated to evaluate dynamic stability collectively in each person with SCI according to various measurement tools and within the overall framework presented. As part of the RCT, participants with SCI were enrolled based on the following inclusion criteria: 1) at least 18 years of age, 2) injury sustained at least 6 months prior to the study, 3) upper motor neuron, motor incomplete spinal cord lesion, 4) ability to ambulate at least 10m with or without an assistive device, and 5) injury of traumatic or non-traumatic origin, excluding those of congenital etiology. From those enrolled in the RCT, a sub-sample was included in the three studies presented here, based on one’s ability to generate at least 3 steps without an assistive device, bodyweight support or physical assistance. Healthy adult controls also were 18 years of age or older; however, they ambulated full-time without assistive devices or physical assistance. All experimental procedures were conducted at the Brain Rehabilitation Research Center, Malcom G. Randall Veteran Affairs Medical Center in Gainesville, Florida. Each participant signed a written informed consent approved by both the VA Subcommittee for Clinical Investigation and the University of Florida Health Science Center Institutional Review Board.
Clinical Assessments

A licensed physical therapist assessed bilateral upper and lower extremity motor and sensory function in persons with SCI based on the American Spinal Injury Association (ASIA) International Standards for Neurological and Functional Classification of Spinal Cord Injury (American Spinal Injury Association 2002). This assessment established SCI severity and categorized injuries according to the ASIA Impairment Scale (AIS). A physical therapist also assessed performance on standardized balance assessments, the Berg Balance Scale (BBS) (Berg et al. 1992) and the Dynamic Gait Index (DGI) (Shumway-Cook and Woollacott 2001), to provide further clinical descriptions of each individual’s balance ability.

Biomechanical Data Collection Procedures

For each study, the following paragraphs detail the overall data collection process. This process generated all data required for the entire dissertation. Therefore, not every component described here is included in each study.

Both participants with SCI and healthy controls had reflective marker balls and rigid body clusters positioned on specified body landmarks to acquire three-dimensional (3D) motion data. Marker positions were based on the Vicon PlugInGait marker set (modified Helen Hayes set). All individuals wore a safety harness attached to an overhead cable and track system that was suspended from the laboratory ceiling. Walking trials lasted a maximum of 30 seconds and were performed over a split-belt instrumented treadmill (Tecmachine, Inc.) (Figure 3-1). Walking practice was permitted to become accustomed to walking on the treadmill and to obtain the best possible steady-state walking speed. When comfortable, data collections commenced. Individuals with SCI performed two trials: the first with their customary assistive device,
the second without any device. Although participants wore a safety harness should they stumble or fall, neither BWS nor manual assistance was provided during any trial. All individuals were instructed to walk to the best of their abilities at self-selected (SS) treadmill speeds during both trials. Trials conducted without assistive devices were intended to reveal the underlying capability of each individual to walk without external assistance and the way each person’s nervous system solved the problem of balance control. Either sitting or standing rest periods were provided as needed between bouts of activity. Healthy control participants performed four separate walking trials. Three trials involved randomly-introduced speeds at 0.3, 0.6, and 0.9 m/s. These slow speeds were selected a priori to represent a likely range within which persons with SCI would elect to walk. During the fourth trial, healthy individuals walked at their SS treadmill speeds.

Although community walking occurs overground rather than over a treadmill and debate continues about the similarities and differences in biomechanics and muscle activity between the two environments, particularly in neurologically-impaired populations (Harris-Love et al. 2001; Parvataneni et al. 2009), measurement of walking in the treadmill environment was necessary for several reasons. First, given the continuum of functional abilities across persons with SCI selected for these studies, some were able to generate only a limited number of steps without an assistive device. Walking overground without a device and without a safety harness would have created an unsafe environment in which to learn about balance strategies in this subset of individuals. Second, since Experiment 2 required 3D ground reaction forces under each foot separately for outcome measurement, walking needed to occur over forceplates.
However, walkways that require subjects to place their feet on two or three individual forceplates, also require several passes across the walkway to capture enough footfalls and to ensure that subjects landed the whole foot on each forceplate. Many individuals with SCI would be unable to complete multiple passes, primarily due to fatigue and balance deficits. In contrast to a walkway, an instrumented split-belt treadmill allows one to capture kinematic and kinetic data from a continuous number of steps in one walking trial. The treadmill can record every footfall without concern that the entire foot was captured. Additionally, if persons with SCI were limited to walking on a treadmill, then healthy controls were required to as well to maintain a consistent environment and equal comparison between both groups. Finally, the treadmill also allowed for a controlled comparison of walking speeds, one of the few variables capable of being controlled in such a heterogeneous population.

**Data Acquisition and Processing**

Twelve camera passive motion analysis (Vicon Motion Systems) and 3D ground reaction forces (GRFs) from four piezoelectric force transducers (Advanced Medical Technology, Inc.) located beneath each half of the treadmill were acquired continuously during walking trials. The split-belt treadmill system allowed collection of GRFs for each stance phase over multiple steps of the gait cycle. Raw kinematic data were collected at 100 Hz, then low-pass filtered using a fourth-order, zero-lag Butterworth filter with a 6 Hz cut-off frequency. GRFs were acquired at a sampling rate of 2000 Hz, and low-pass filtered using a fourth-order, zero-lag Butterworth filter with a 20 Hz cut-off frequency. A 13-segment musculoskeletal model was created using Visual 3D (V3D) processing that fit the model to marker trajectories. Anthropometric and inertial values defined within V3D were applied for segment modeling and segmental center-of-mass (CoM)
calculations. V3D models were used to conduct inverse dynamics analyses for calculation of intersegmental joint kinetics. Custom Matlab programming (Mathworks, Inc.) was developed to calculate the outcome measures.

Excellent absolute agreement has been reported between Vicon motion analysis systems and GaitRite instrumented walkways for measurement of spatiotemporal parameters (ICC= 0.92-0.99) (Webster et al. 2005). In addition, center-of-mass estimations calculated via the Vicon system and a Crossbow accelerometry system demonstrated excellent concurrent validity (r=0.87) (Lee et al. 2007). Furthermore, the split-belt instrumented treadmill system utilized in these three studies has been validated as an instrument for gait analysis as well (Tesio and Rota 2008).

**Dynamic Stability Outcome Measures and Rationale**

**Experiment #1**

**Head and pelvic motion.** A literature base is mounting which evaluates dynamic stability based on the ability of the normally functioning human body to reduce perturbations to the head, which houses critical sensory apparatuses for balance. This function allows for enhanced gaze stabilization and integration of accurate vestibular inputs for balance control while moving. Recently, common measurement techniques have examined the degree of head acceleration attenuation relative to the pelvis (Menz et al. 2003a; Kavanagh et al. 2004; Mazza et al. 2008). Using this literature as a foundation, both displacement and acceleration patterns of the upper body were evaluated. Segmental CoM positions were used to calculate 3D displacements of head and pelvis CoMs were determined for each step in a trial. Subsequently, position data were double-differentiated to acquire 3D linear accelerations for the same steps. Visual inspection of acceleration profiles located abnormally high and low spikes, which were
interpreted as non-physiological noise. Therefore, maximal and minimal thresholds were manually-selected on the profiles via customized Matlab coding, and steps with accelerations exceeding the established thresholds were eliminated.

**Accelerations.** Root mean squares (RMS) of frame-by-frame acceleration values for both the head and pelvis CoMs in each direction were calculated from valid steps. RMS values then were used to calculate attenuation coefficients (AC) for the entire trial as described by Mazza et al. (2008; 2009).

\[ \text{AC} = (1 - (\text{RMS}_H/\text{RMS}_P)) \times 100, \text{ where } H = \text{head}, \ P = \text{pelvis} \]

Attenuation coefficients, expressed as percentages, represent the ability of the individual to reduce accelerations from the pelvis to the head. A positive, larger coefficient indicates greater head stability relative to the pelvis. In contrast, negative coefficients represent larger head accelerations (less stability) relative to the pelvis.

**Displacements.** The means and standard deviations of 3D head and pelvis CoM displacements were calculated. Holt et al. (1999) quantified vertical (VT) head displacements in the sagittal plane using a single head marker as an indicator of head stability. We expanded our methodology to include the anteroposterior (AP) and mediolateral (ML) directions in order to parallel our 3D acceleration outcomes and additionally selected the segmental head CoM as a more robust measurement of head motion than a single marker. Using similar methodology to Hong and Earhart (2010) for the relative motion between two segments, the difference between pelvis and head mean displacements (“mean difference,” \( \text{mean}_{\text{pelvis}} - \text{mean}_{\text{head}} \)) was an outcome indicating which segment produced greater displacements over the walking trial. The difference between pelvis and head displacement standard deviations (SD) (“variability
difference, \( \text{SD}_{\text{pelvis}}-\text{SD}_{\text{head}} \) was an outcome indicating whether the head or pelvis produced more variable displacements over the course of a walking trial. Similar to the acceleration attenuation coefficients, positive values for difference calculations represent less head motion relative to the pelvis, while negative values represent greater head motion relative to the pelvis.

**Experiment #2**

**Spatial foot parameters and foot placement relative to pelvis center-of-mass.** Since walking creates a situation in which the CoM is outside the base of support for 80 percent of the gait cycle (Winter 1995), the CoM is continually falling forward with each step. Thus, awareness of where a foot lands at initial contact provides insight into how the CoM is “restabilized” with each step, and consequently, the way in which a person maintains balance over a series of steps. The placement of one foot relative to another during double support (e.g. step width, step length) is commonly documented in the clinic using observational analyses (Krebs et al. 2002). Yet that method may or may not provide insight into the control of the CoM as the balance point of the body. Additional measures of foot placement indicating the location of the foot relative to the “falling” body might provide another level of information (Redfern and Schumann 1994; Pai and Patton 1997; Balasubramanian et al. 2010). Overall, four measures were defined as the following distances at each initial contact: 1) leading foot CoM to trailing foot CoM in the ML direction (“Step width”) and in the AP direction (“Step length”) (Figure 3-2) and 2) leading foot CoM to pelvis CoM in the ML direction (“ML foot placement”) and in the AP direction (“AP foot placement”) (Figure 3-3).

**Margin of stability (MoS).** MoS, a measure developed by Hof et al. (2005), estimates dynamic stability using a method derived from the conventional comparison of
CoM position to center-of-pressure (CoP). In contrast to the conventional comparison, Hof and colleagues accounted for the influence of velocity on the CoM position (2005). As such, MoS compares the shortest perpendicular, mediolateral distance between the CoP and the extrapolated center of mass (XcoM) during single limb stance of gait (Figure 3-4). The XcoM is the vertical projection of the CoM in the direction of its velocity.

1. \[ \text{XcoM} = \left| x + \left( \frac{\nu}{\omega} \right) \right| \], where \( x \) = CoM position, \( \nu \) = CoM velocity, \( \omega = \sqrt{\text{gravity/leg length}} \)

2. \[ \text{MoS} = \left| \text{umax} - \text{XcoM} \right| \], where \( \text{umax} \) = boundary of CoP

A larger MoS is consistent with increased dynamic stability as a greater margin exists between the XcoM and the CoP. Furthermore, the size of the margin is proportional to the impulse that would be required to destabilize or unbalance a person. That is, by “adding” a critical amount of CoM velocity (i.e. impulse) in the direction of the nearest CoP, an unstable situation can be created.

**Experiment #3**

**Center-of-mass (CoM) trajectory.** Critical biomechanical control elements are responsible for preventing collapse of the body as it produces a continuous, smooth forward progression during walking. Research of dynamic stability during walking suggests that the end-product of these control elements is motion of the body’s CoM. During walking, the upper and lower body must complement one another and shift in a coordinated effort to recapture the CoM within the CoP during each step since the CoM lies outside the CoP for most of the gait cycle (Winter 1995). Therefore, trajectory of the CoM indirectly exemplifies the compensatory strategies of all other body segments to avoid falling.
For each stride in a walking trial, the length of the whole body CoM trajectory normalized to stride length was calculated in the transverse plane (Figure 3-4). Stride length was defined as the anteroposterior distance from the initial contact of one foot to the next initial contact of that same foot (heel marker to heel marker) (Figure 3-5) (Perry 1992). Length of the CoM trajectory may illustrate whether an individual progresses normally in a smooth sinusoidal nature or whether the trajectory deviates in alternate directions in an attempt to control the CoM during a given step. The latter situation would create a longer CoM trajectory. Alternatively, a shorter than normal CoM trajectory may also be possible should other biomechanical characteristics such as step width, or perhaps neuromuscular properties such as hypertonicity or muscle tightness, limit CoM motion.

**Trunk excursions.** Literature investigating falls in the elderly population has reported that a primary factor, which distinguishes older adults who fall from those who do not, is the ability to limit trunk movements after a perturbation during walking (Grabiner et al. 2008). Additionally, when the trunk is in a flexed position, other body segments, such as the pelvis and lower extremities, compensate by altering their alignment to restore control of the CoM (Saha et al. 2008). Therefore, trunk excursions were used as an adjunctive measure of balance to complement the CoM trajectory length in Experiment 3. Three-dimensional trunk angular displacement ranges were determined for each step in a walking trial. Trunk displacements were defined as the angles of rotation (degrees) about each axis. The established reference axes were positioned orthogonal to one another with the x-axis directed mediolaterally, the y-axis anteroposteriorly, and the z-axis vertically. Proximal and distal ends of the modeled
trunk segment were used to calculate movement about the x- and y-axes, while markers positioned on bilateral acromion processes were used to calculate rotation about the z-axis.

**Spatial foot parameters.** As mentioned previously in Experiment 2, foot location relative to the contralateral foot is a common clinical measurement. The ease of describing spatial parameters ("wide base-of-support", “variable step lengths”) (Krebs et al. 2002) or even collecting spatial measurements using a low-cost instrumented walkway makes these parameters particularly clinic-friendly (Brach et al. 2001). Unlike the spatial foot parameters described earlier in Experiment 2, which utilize foot CoM locations and require complex motion analysis equipment, step width and step length also were measured using body landmarks more applicable to the way in which these parameters could be gathered in the clinic environment. Step width was defined as the mediolateral distance between the heel marker of the leading foot at initial contact and the heel marker of the contralateral, trailing limb at that same point in time. Step length was defined as the anteroposterior distance between the heel marker of the leading foot at initial contact and the heel marker of the contralateral, trailing limb at that same point in time (Figure 3-5) (Perry 1992).

**Locomotor Training Intervention**

The locomotor training (LT) intervention for individuals with SCI in Experiment 3 occurred in either a manual or robotic environment, both activity-based therapies (Figure 3-6) (Dromerick et al. 2006). While training in the manual environment, a harness suspended participants with varying amounts of BWS over a moving treadmill as human trainers physically facilitated walking biomechanics. Three to four trainers placed their hands at the pelvis/trunk for rotation, shifting, and upright posture, at
bilateral lower extremities for optimal stepping kinematics with appropriately-timed gait events, and at bilateral upper extremities to promote arm swing, as needed (Behrman and Harkema 2000). Although training in the robotic environment (Lokomot, Hocoma) also occurred with a harness, BWS, and a treadmill, this training scenario immobilized the pelvis and legs in an exoskeleton and utilized a computerized system to drive stepping in lieu of human trainers. The robotic environment possessed an additional feature called guidance force, which systematically reduced the percentage of stepping assistance provided. This feature intended to mimic the progressive decrease of human trainer assistance in manual training as participants exhibited greater independence (Colombo et al. 2000; Reinkensmeyer et al. 2004; Galvez et al. 2007).

Participants involved in Experiment 3 were randomized to one of these two environments as part of a larger randomized clinical trial. Training occurred three to five days per week for 45 sessions total. At least one licensed physical therapist was present always during training. The treatment goal was for participants to achieve 30 minutes of quality stepping per session; however, the number and length of individual bouts necessary to obtain this goal varied daily depending on each participant’s functional ability and/or trainer fatigue. Over the 45 sessions, participants were challenged continually as they demonstrated greater independence while exhibiting appropriate stepping patterns, including an upright trunk. Specifically, BWS decreased, treadmill speed increased and guidance force or trainer manual assistance decreased. BWS began at approximately 40% unloading with the ultimate goal of attaining 0%. Treadmill speed was initiated slower than normal walking speed (normal ≥ 0.8 m/s) (Perry et al. 1995), then increased as soon as possible to normal and beyond according to
participant and/or trainer abilities. Guidance force on the Lokomat began at 100% and reduced toward 0% once treadmill speed was at least in a normal range and BWS was below 20%. Rest periods, typically standing with assistance for posture and BWS lowered to promote loading, were provided between stepping bouts. Blood pressure and heart rate were obtained before, at regular intervals during and after training sessions to assess participants’ physiological responses to exercise and to ensure episodes of autonomic dysreflexia or alternatively postural hypotension were avoided or addressed as necessary. Borg Ratings of Perceived Exertion also were recorded regularly according to the participant’s report.
Figure 3-1. Split-belt instrumented treadmill, harness and instrumentation.
Figure 3-2. Spatial foot parameters using feet center-of-mass (CoM) as body reference points. White arrows represent (A) step width and (B) step length. This representative motion analysis graphic depicts an individual with SCI walking with a left forefoot initial contact.

Figure 3-3. Foot placement relative to center-of-mass (CoM) distances. Body reference points are the leading foot CoM and the pelvis CoM. White arrows represent (A) mediolateral foot placement, (B) anteroposterior foot placement. The same individual with SCI seen in Figure 3-2 is depicted here.
Figure 3-4. Example of center of pressure (CoP), center of mass (CoM), and extrapolated center of mass (XcoM) trajectories during walking with margin of stability (MoS) represented at the transition from double limb support to single limb support. CoM trajectory length also demarcated between dotted lines for a single step (1/2 stride) in the transverse plane.

Figure 3-5. Step length (a) and step width (b) defined for a given gait cycle. Gray circles represent reflective heel markers.
Figure 3-6. (A) Manual-assisted locomotor training, (B) Robotic-assisted locomotor training
Following a spinal cord injury (SCI), individuals possess altered, and what could be described as “new,” nervous systems (Edgerton et al. 2004). This injured nervous system is particularly plastic and inclined to learning based on the task-specific sensorimotor experiences in which a person engages. Therefore, as an individual attempts to relearn walking, the experiences introduced influence neural changes which manifest in 1) walking recovery in the way one walked prior to injury, or alternatively, 2) compensatory strategies that cope with losses and use only residual motor functions (Barbeau 2003).

Traditionally in walking rehabilitation post-SCI, clinicians introduce assistive devices (ADs) to patients. The intention is to provide support through the upper extremities and compensate for impairments contributing to balance dysfunction. However, these ADs provide externally-controlled stability that may be preventing individuals from exploring normal balance strategies (Bateni and Maki 2005). Moreover, ADs may be teaching a task with sensory and biomechanical attributes different from normal walking (Jeka 1997; Dromerick et al. 2006; Ivanenko et al. 2009). Thus, a person likely is learning balance strategies in the context of having an AD present, rather than relearning how to develop balance strategies in the absence of external assistance. Evaluating an individual while walking without his/her AD, therefore, may reveal the way the person’s nervous system is coping with SCI-mediated deficits and the solutions found in an attempt to recover balance.

In contrast to a person with an injured central nervous system, a person with a healthy nervous system has an array of possibilities for controlling walking balance. An
individual’s body segments coordinate together to create an apparent effortless and smooth walking trajectory. Research suggests that this smoothness is necessary in order to stabilize the position of the head (Menz et al. 2003a; Kavanagh et al. 2004; Kavanagh et al. 2005b). Stabilization of the head is essential for acquisition of accurate sensory inputs to the visual and vestibular systems, since these two systems have primary roles in maintaining balance control (Shumway-Cook and Woollacott 2001). In addition, when the feet initially contact the ground during walking, high linear accelerations at the legs transfer perturbations from body segment to body segment (e.g. from shank to thigh to trunk) (Kavanagh et al. 2006). In order to optimize head stability and maintain balance control under these conditions, linear accelerations decrease in a proximal-to-distal manner from the pelvis to the head (Winter et al. 1990; Menz et al. 2003a).

The variable upper body movements often observed in persons post-SCI have potential to inundate the nervous system with conflicting visual and vestibular information, thus further propagating balance dysfunction. The conventional implementation of ADs to provide leg and trunk support may also reduce upper body motion and afford head stabilization. This result may further enhance compensated balance. Additionally, visual field stabilization and balance control can improve through a self-elected reduction in walking speeds, resulting in enhanced vestibulo-ocular reflexes (Mamoto et al. 2002). Melis et al. (1999) observed slow walking speeds in persons with SCI when walking with ADs. Speeds appeared to be at least partially attributed to type of device used with walker users exhibiting slower speeds than both crutch and cane users. Consequently, given the clinical utilization of ADs to provide
postural stability during walking as well as the associated compromised walking speeds while using devices, it is plausible to suggest that ADs may assist in achieving head stabilization, thus affording externally-derived balance control for the SCI population.

Therefore, the main purposes of this study were to investigate 1) the effect of ADs on head stability post-SCI and 2) the effect of SCI on head stability during walking without ADs compared to healthy individuals. The first part of our investigation will provide insights into whether ADs have a role in stabilizing the head relative to the pelvis for balance control. The second part of the study will assist in understanding the abilities of individuals to maintain normal levels of head stability after injury when tested in a recovery-based environment without ADs. This particular assessment also could elucidate the possible long-term influences of not only the SCI but also chronic AD usage. Based on clinical intentions for ADs to stabilize and support the upper body, we hypothesized that persons with SCI would demonstrate greater head stability while walking with ADs compared to without. Specifically, head accelerations would attenuate to a greater degree relative to the pelvis, and head displacements would possess less magnitude and variability relative to the pelvis when walking with ADs. Additionally, because the individuals with SCI will have experienced only walking conditions with ADs since injury, we anticipated they immediately would exhibit reduced head stability when devices were removed compared to healthy individuals. In particular, persons with SCI would show increased head accelerations and larger, more variable head displacements relative to the pelvis compared to healthy persons. The ability to preserve head stability with and without ADs may offer insights into how a person’s
nervous system prioritizes and solves walking balance dysfunction after injury when given the opportunity to explore two different sensory experiences.

Methods

Participants

A convenience sample of 10 individuals with chronic, incomplete SCI (6 males; mean age=42.6 years, SD=14.2) (Table 4-1) and 10 healthy persons comprising a control group (3 males; mean age=56.1 years, SD=3.3) participated in this cross-sectional study. Participants with SCI were involved in a larger randomized controlled trial with the following inclusion criteria: 1) at least 18 years of age, 2) injury sustained at least 6 months prior to the study, 3) upper motor neuron, motor incomplete spinal cord lesion, 4) ability to ambulate at least 10m with or without an assistive device, and 5) injury of traumatic or non-traumatic origin, excluding those of congenital etiology. The subset of individuals selected for this study from the larger trial included those who could generate at least three steps without an assistive device, BWS, or physical assistance. The healthy adult controls were a convenience sample from a larger ongoing cross-sectional study. All controls were 18 years of age or older and full-time ambulators without assistive devices or physical assistance. All experimental procedures were conducted at the Brain Rehabilitation Research Center, Malcom G. Randall Veteran Affairs Medical Center in Gainesville, Florida. Each participant signed a written informed consent approved by both the VA Subcommittee for Clinical Investigation and the University of Florida Health Science Center Institutional Review Board.
Experimental Procedures

A licensed physical therapist assessed bilateral upper and lower extremity motor and sensory function in persons with SCI based on the American Spinal Injury Association (ASIA) International Standards for Neurological and Functional Classification of Spinal Cord Injury (American Spinal Injury Association 2002). This assessment established SCI severity and categorized injuries according to the ASIA Impairment Scale (AIS). A physical therapist also assessed performance on standardized balance assessments, the Berg Balance Scale (BBS) (Berg et al. 1992) and the Dynamic Gait Index (DGI) (Shumway-Cook and Woollacott 2001), to provide further clinical descriptions of each individual’s balance ability.

Both participants with SCI and healthy controls had reflective marker balls and rigid body clusters positioned on specified body landmarks to acquire 3D motion data. Marker positions were based on the Vicon PlugInGait marker set (modified Helen Hayes set). All individuals wore a safety harness attached to an overhead cable and track system that was suspended from the laboratory ceiling. Walking trials lasted a maximum of 30 seconds and were performed over a split-belt instrumented treadmill (Tecmachine, Inc.). Walking practice was permitted to become accustomed to walking on the treadmill and to obtain the best possible steady-state walking speed. When comfortable, data collections commenced. Those individuals with SCI performed two trials during which they were instructed to walk to the best of their abilities at a self-selected (SS) treadmill speed (Figure 4-1): the first trial with their customary assistive device, the second without any device. Although participants wore a safety harness should they stumble or fall, neither BWS nor manual assistance was provided during any trial. This testing condition was intended to capture true walking capacity post-
injury. Either sitting or standing rest periods were provided as needed between bouts of activity. Additionally, healthy controls performed three separate walking trials at 0.3 and 0.6 m/s as well as a SS treadmill speed. The first two speeds were selected to acquire normal comparisons at the slower speeds that persons with SCI elected to walk. Pepin et al. (2003b) suggest matching for speeds between SCI and control groups to eliminate speed as a potential confound and distinguish outcomes that are consequences of the injury rather than speed. Walking at SS speeds also was conducted as much of the scientific literature assessing head stability in healthy adults utilizes SS speeds (Menz et al. 2003a; Kavanagh et al. 2004; Mazza et al. 2008). Data from SS speeds validated the similarity of our healthy adult results with other studies. This step was necessary to confirm our methodology and calculations prior to initiating comparisons at slower speeds.

**Data Acquisition and Processing**

Twelve camera passive motion analysis (Vicon Motion Systems) and 3D ground reaction forces (GRFs) from four piezoelectric force transducers (Advanced Medical Technology, Inc.) located beneath each half of the treadmill were acquired continuously during walking trials. The split-belt treadmill system allowed collection of GRFs for each stance phase over multiple steps of the gait cycle. Raw kinematic data were collected at 100 Hz, then low-pass filtered using a fourth-order, zero-lag Butterworth filter with a 6 Hz cut-off frequency. GRFs were acquired at a sampling rate of 2000 Hz, and low-pass filtered using a fourth-order, zero-lag Butterworth filter with a 20 Hz cut-off frequency. A 13-segment musculoskeletal model was created using Visual 3D (V3D) processing that fit the model to marker trajectories. Anthropometric and inertial values defined within V3D were applied for segment modeling and segmental center-of-mass (CoM)
calculations. V3D models were used to conduct inverse dynamics analyses for calculation of intersegmental joint kinetics. Custom Matlab programming (Mathworks, Inc.) was developed to calculate the outcome measures.

**Outcome Measures**

Using calculated segmental CoM positions, 3D displacements of head and pelvis CoMs were determined for each step in a trial. Subsequently, position data were double-differentiated to acquire 3D linear accelerations for the same steps. Visual inspection of acceleration profiles located abnormally high and low spikes, which were interpreted as non-physiological noise. Therefore, maximal and minimal thresholds were manually-selected on the profiles via customized Matlab coding, and steps with accelerations exceeding the established thresholds were eliminated.

**Accelerations.** Root mean squares (RMS) of frame-by-frame acceleration values for both the head and pelvis CoMs in each direction were calculated from valid steps. RMS values then were used to calculate attenuation coefficients (AC) for the entire trial as described by Mazza et al. (2008; 2009).

\[ AC = (1 - (\text{RMS}_H/\text{RMS}_P)) \times 100, \text{ where } H = \text{head}, P = \text{pelvis} \]

Attenuation coefficients, expressed as percentages, represent the ability of the individual to reduce accelerations from the pelvis to the head. A positive, larger coefficient indicates greater head stability relative to the pelvis. In contrast, negative coefficients represent larger head accelerations (less stability) relative to the pelvis.

**Displacements.** The means and standard deviations of 3D head and pelvis CoM displacements were calculated. Holt et al. (1999) quantified vertical head displacements as an indicator of head stability, and we expanded our methodology to include the anteroposterior and mediolateral directions in order to parallel our 3D acceleration
outcomes. Using similar methodology to Hong and Earhart (2010) for the relative motion between two segments, the difference between pelvis and head mean displacements (\(\text{mean difference,} \ \text{mean}_{\text{pelvis}} - \text{mean}_{\text{head}}\)) was an outcome indicating which segment produced greater displacements over the walking trial. The difference between pelvis and head displacement standard deviations (\(\text{variability difference,} \ \text{SD}_{\text{pelvis}} - \text{SD}_{\text{head}}\)) was an outcome indicating whether the head or pelvis produced more variable displacements over the course of a walking trial. Similar to the acceleration attenuation coefficients, positive values for difference calculations represent less head motion relative to the pelvis, while negative values represent greater head motion relative to the pelvis.

**Data Analysis**

All statistical analyses were conducted using SAS 9.13 software. A series of normality tests were conducted to ensure that SCI and control data for each outcome met all assumptions for parametric testing. Paired t-tests were conducted to examine differences in the SCI group with and without ADs. For comparisons between the SCI and control group at matched speeds, outcomes for each person with SCI were standardized using standard differences from the control group walking at a similar speed (SCI value minus control group mean divided by control group standard deviation). Thus, the control group mean equaled zero, variance equaled one, and SCI values beyond ±2 standardized scores were considered outliers. Speed matching was established as follows: SCI \(\leq 0.3\) m/s = controls at 0.3 m/s; and SCI \(>0.3\) m/s to \(\leq 0.6\) m/s = controls at 0.6 m/s. Visual analysis of SCI standardized data indicated heterogeneity in the sample with variability in both positive and negative directions relative to controls. Therefore, in order to avoid masking true differences in the SCI
sample and regressing SCI data toward a mean value, which might demonstrate no
difference from controls, absolute values were calculated to determine a participant’s
distance from a central point of zero for each outcome. The mean of all participants’
absolute values was used for group comparisons of SCI and healthy individuals using a
permutation test. Alpha level for significance was set at .05.

Results

With versus Without Assistive Devices in SCI

Accelerations. Figure 4-2 illustrates raw head and pelvis trajectories for one
representative participant with SCI (SCI2) walking without a device. 3D attenuation
coefficients (ACs) exhibited no statistical difference between persons with SCI walking
with and without ADs ($p \geq 0.05$) (Figure 4-3). On average in both walking conditions,
participants demonstrated increased head accelerations relative to the pelvis in both the
mediolateral (ML) and vertical (VT) directions. In the anteroposterior (AP) direction,
head accelerations were also higher when walking with ADs; but without ADs, slightly
lower head accelerations relative to the pelvis were exhibited (i.e. greater head
stability). Figure 4-3 further illustrates the large standard deviations for all ACs. The
individual participants presented with varying combinations of AC values (e.g. a positive
value with AD and negative without AD or vice versa as well as positive or negative
values in both walking conditions). Although averaged AC values showed no differences
across conditions, RMS of the head and pelvis, which were used to calculate
attenuation coefficients, showed significantly greater accelerations at both the head and
pelvis in the ML direction when walking without ADs ($p = 0.009$ and $p = 0.006$,
respectively). RMS of the pelvis in the AP direction also was significantly greater without
ADs ($p=0.002$). Remaining RMS values for the head and pelvis were not significantly different between walking conditions ($p \geq 0.05$).

**Displacements.** Figures 4-4 A&B depict displacement mean differences and variability differences. In the ML direction, both the mean difference and variability difference demonstrated significantly greater head movement relative to the pelvis when walking without ADs (mean, $p=0.038$; variability, $p=0.039$). Regardless of walking condition, ML head displacements were larger than the pelvis (negative values). All mean and variability differences for the AP and VT directions showed no differences between walking conditions ($p \geq 0.05$). On average, the mean differences in these two directions with and without ADs revealed smaller head displacements relative to the pelvis. The opposite pattern was detected in variability as head displacements were overall more variable than the pelvis in the AP and VT directions; although without ADs, AP head and pelvis variability were similar (variability difference $\sim 0$).

**SCI Without Assistive Devices versus Controls at Matched Speeds**

**Accelerations.** Figure 4-5 Illustrates individual subject data for attenuation coefficients. When matched for walking speeds using standard differences from the controls, the SCI group walking without their customary ADs demonstrated significantly different AC values in the AP and VT directions compared to controls ($p \leq 0.0001$). No differences were detected in the ML direction ($p=0.2823$). Regardless of direction, ACs were reduced compared to the control mean, indicating a decreased ability to attenuate head accelerations. RMS values for the head and pelvis in all three directions were significantly different from controls ($p \leq 0.05$).

**Displacements.** Figure 4-6 A&B illustrates the heterogeneity of individual subject data for both mean differences and variability differences, respectively. As a group,
those with SCI showed significant differences in displacement mean differences in the VT direction only compared to controls \((p \leq 0.0001)\). Yet, the variability in head displacements relative to pelvis displacements was significantly different from controls in all three directions \((p \leq 0.05)\).

**Discussion**

The framework and progression of comparisons in this study were developed to parallel the current paradigm shift occurring in SCI rehabilitation. We began with an investigation to understand if our traditional rehabilitation approach using ADs affected walking balance, specifically head stability, differently than when individuals were challenged to walk without ADs. Our model then shifted to evaluate both persons with SCI and healthy individuals in the same walking context (i.e. without any external assistance besides a safety harness). Although a more common experimental design is for studies to *first* show comparisons of patient populations to healthy controls, we specifically wanted to highlight the transition from a conventional evaluation approach to one that is task-specific and emphasizes walking recovery.

**Assistive Devices Impact Head Motion Relative to the Pelvis**

This is the first study to examine head stability, as an indicator of balance control, in this neurological population. In both acceleration and displacement outcomes, and regardless of whether devices were utilized, persons with SCI demonstrated great heterogeneity in their head stabilization as exemplified by large standard deviations around the mean values. Although no statistical differences were detected in ACs when walking with versus without devices, ACs revealed the same patterns of change as shown in the displacement mean and variability differences for their respective ML and AP directions. Specifically, ADs appeared to play a role in reducing the magnitude and
variability of ML head displacements relative to the pelvis. While significant differences were not detected in the AP direction, the opposite pattern of change emerged compared to the ML direction. That is, the displacement mean and variability differences actually became more positive when ADs were removed to the extent that the mean value across participants with SCI reflected less head motion overall compared to the pelvis (i.e. a positive value). This occurrence corroborates research in healthy individuals, which found greater AP head stability relative to the pelvis in comparison to the ML or VT directions (Kavanagh et al. 2004; Mazza et al. 2009). These findings suggest that preservation of stability in the AP direction may be a high priority for the nervous system during forward walking. Perhaps when devices are removed after injury, that particular direction of stability remains a top priority, potentially to achieve accurate visual and vestibular sensory inputs as one progresses forward. This prioritization may occur even at the expense of increasing ML head motion as seen here.

Interestingly, the VT direction demonstrated remarkably higher head accelerations relative to the pelvis with and without ADs. This would appear highly unusual in reference to the healthy literature, which suggests that VT accelerations at the head and pelvis should be the same because the spine mechanically links the two segments (Mazza et al. 2008). However, in clinical observations of walking, persons with SCI often use a quick head thrust in an upward and backward rotation to biomechanically achieve momentum and progress the lower body forward. Even though the head may not be vertically displaced much farther than the pelvis as seen in our study, which is likely due to the spine’s mechanical constraint, it is plausible that the quick motion could manifest
in higher head accelerations. Moreover, our findings in the vertical ACs suggest that this biomechanical strategy persists regardless of AD use.

**Spinal Cord Injury Affects Head Stability Multidirectionally**

Compared to normal individuals walking at comparably slow speeds, ACs in the AP and VT directions were significantly reduced in persons with SCI walking without devices (i.e. they presented with less head stability than is observed in healthy persons). For all participants, ML attenuation levels were similar to healthy controls. Interestingly, both the ML and AP displacement mean differences were not different from healthy individuals; yet, the variability in all directions was significantly different. An increasing amount of literature proposes that the lack of consistency in one’s movement, as opposed to its magnitude, is a more appropriate indicator of balance during walking (Brach et al. 2001; Granata and Lockhart 2008). In accordance with this view, head stability research also has expanded to include measurements of variability (Holt et al. 1999; Laudani et al. 2006). Thus, the increased multidirectional variability of head motion relative to the pelvis post-SCI is suggestive of reduced balance control. If one is unable to create head stability (due to excessive trunk movements perhaps), he/she may be compensating with other sensory inputs. For example, individuals post-incomplete SCI may have spared neural pathways that retain adequate ankle, knee and/or hip proprioception or plantar somatosensation. Thus, their alternative strategies to avoid falling may be to precisely place their feet and counteract the trunk and head motion with compensatory leg movements.

However, another important factor when evaluating someone’s head motion may be to consider whether the variability might be secondary to muscle weakness, in particular paraspinal muscle weakness. Although the passive mechanics of the spine
may have a stabilizing effect, the paraspinals also have been credited with stabilizing the upper body as they activate in a top-down pattern during walking (Prince et al. 1994). Presumably, this recruitment order is a feed-forward mechanism to stabilize the head in anticipation of perturbations. Our participant sample with SCI included seven individuals with cervical injuries and three with upper thoracic injuries. Upper trunk and neck muscle weakness occurs with injuries along those levels (Larson et al. 2010), therefore although not specifically tested, it is plausible to suggest that those with SCI had difficulty producing anticipatory muscle responses during walking. As a result, head stability was compromised. Although Thigpen et al. (2009) reported relatively normal anticipatory muscle responses in the lower extremities during a standing perturbation task in individuals post-SCI, it cannot be assumed that anticipatory responses of the trunk muscles would respond the same way during walking since standing and walking have been shown to possess different stability requirements (Kang and Dingwell 2006).

**Conclusions**

The ability to maintain head stability after SCI varied greatly both with and without assistive devices and relative to healthy individuals. However, patterns of stability emerged which suggest that ADs may influence mediolateral head stability relative to the pelvis and that in the absence of ADs, individuals prioritize the ability to stabilize anteroposterior head motions. Variability of head displacements relative to the pelvis appeared particularly deficient compared to controls in all three dimensions as did the ability to attenuate accelerations in the AP and VT directions. Future work on head stabilization in this population should evaluate other segmental relationships located between the head and pelvis (i.e. pelvis to trunk, trunk to neck, neck to head) to provide a more in-depth view of how head stability is or is not achieved. Continued efforts
should be taken to understand the impact of ADs on dynamic stability, their influence on
the plasticity of the nervous system, and their utility versus other evaluations strategies
(e.g. treadmill-based or overground with a harness only) in assessments of dynamic
stability.
Table 4-1. Participant demographics

<table>
<thead>
<tr>
<th>Participant</th>
<th>Age (yrs)</th>
<th>Sex</th>
<th>Injury site</th>
<th>Time post SCI (mos)</th>
<th>LEMS (max:50)</th>
<th>AIS</th>
<th>BBS (max:56)</th>
<th>DGI (max:24)</th>
<th>Assistive device</th>
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<tr>
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<td>M</td>
<td>C5-6</td>
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<td>D</td>
<td>46</td>
<td>17</td>
<td>RW</td>
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<td>D</td>
<td>51</td>
<td>12</td>
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<td>F</td>
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<td>19</td>
<td>12</td>
<td>RW</td>
</tr>
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<td>10</td>
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<td>40</td>
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<td>RW</td>
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<tr>
<td>SCI10</td>
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<td>D</td>
<td>42</td>
<td>14</td>
<td>SPC</td>
</tr>
</tbody>
</table>

Figure 4-1. Differences in self-selected treadmill speeds with and without customary assistive devices (AD) for participants with SCI.

Figure 4-2. Example of 3D raw acceleration profiles of head and pelvis trajectories (SCI2 without device). Dashed ovals outline regions of reduced head accelerations relative to the pelvis for a particular percent of the gait cycle. ML: mediolateral, AP: anteroposterior, VT: vertical, acc: acceleration.
Figure 4-3. Acceleration attenuation coefficients for persons with SCI walking with and without assistive devices (ADs). ML: mediolateral, AP: anteroposterior, VT: vertical. Error bars indicate standard deviations. No significant differences were detected between walking conditions.
Figure 4-4. Relative motion of head and pelvis displacements. (A) Mean differences and (B) variability differences in persons with SCI walking with and without assistive devices (ADs). ML: mediolateral, AP: anteroposterior, VT: vertical. Error bars indicate standard deviations. (*) denotes significant difference between walking conditions.
Figure 4-5. (A) Mediolateral, (B) anteroposterior, and (C) vertical standardized attenuation coefficients presented for each participant with SCI compared to a control mean of zero (variance=1). (*) denotes SCI group significantly different from controls.
Figure 4-6. Standardized SCI data for relative motion of head and pelvis displacements. (A) Mean differences and (B) variability differences in the mediolateral, anteroposterior, and vertical directions compared to a control mean of zero (variance=1). (*) denotes SCI group significantly different from controls.
Balance during walking requires both upright posture and equilibrium control to avoid falling (Winter et al. 1990). In conjunction with the ability to reciprocally step and to adapt to the environment, balance is critical in achieving functional ambulation (Forssberg et al. 1980a; Barbeau et al. 2006). Persons with spinal cord injury (SCI) report balance dysfunction as a contributor to this population’s high falls incidence and secondary injuries (Brotherton et al. 2007a). Laboratory gait analyses have reported that reliance on assistive devices that restrict excessive movements, the inability to bear weight fully through the lower extremities, and trunk flexion adaptations all indicate poorer balance control post-SCI (Karcnik and Kralj 1999; Field-Fote et al. 2001; Behrman et al. 2005; Leroux et al. 2006; Scivoletto et al. 2008). Spinal cord lesions affect the sensorimotor systems involved in balance control, thus creating precarious walking conditions prone to falls, particularly when compensatory assistive devices are removed (Shumway-Cook and Woollacott 2001; van Hedel et al. 2005; Horak et al. 2009). In order to rehabilitate this loss of balance control, measurement tools that can quantify distinctive balance strategies of the SCI population are necessary to understand the effectiveness of interventions as well as an individual’s progress over time.

Clinicians commonly document the quality of spatial parameters at the feet, such as “wide base-of-support” when observing an increased step width, to describe the stability in a person’s walking pattern. However, these clinical observations do not account for the dynamic control of the body’s center-of-mass (CoM), which is essential to avoid falling. As walking progresses forward, the CoM accelerates past the single
stance limb and causes the CoM to “fall” with each step. Consequently, precise placement of the contralateral swing foot is critical to recapture the CoM within the base of support (Patla et al. 1999). If the foot contacts the ground in a sub-optimal position, subsequent steps must adjust to compensate for this error, thus inducing small perturbations. Studies of soleus H-reflexes during healthy walking confirm this nervous system priority to control placement of the foot and ensure stability when balance is perturbed (Krauss and Misiaszek 2007). Following SCI, each step during level overground walking could be viewed as its own perturbation to the neuromuscular system necessitating corrective foot placements.

Increased or decreased variability in a patient’s spatial walking pattern relative to the amount of variability observed in healthy individuals often is considered indicative of instability (Stolze et al. 2000; Brach et al. 2001). However, the ability to balance may be more related to where one places his/her foot relative to the CoM rather than relative to the opposite foot. Balasubramanian et al. (2010) reported a significant positive relationship between step length asymmetry and asymmetry in the anterior placement of the foot relative to the pelvic CoM in persons post-stroke; however, no relationship existed between step width and the lateral placement of the foot relative to the CoM. This disparity between measures suggests that a clinical examination of foot-to-foot distances as an indicator of balance may not always reflect a person’s true ability to control CoM motion through foot placements.

An additional measure has been developed in the laboratory to characterize balance control beyond spatial parameters and foot placement relative to the CoM. Given the dynamic nature of walking, an expansion of these static measurement
approaches has been suggested to account for the impact of velocity on the CoM position during this task. Hof et al. proposed a measure called the margin of stability (MoS) (Hof et al. 2005; Hof 2008). This measure compares the shortest mediolateral distance between the center-of-pressure (CoP) and the extrapolated center of mass (XcoM) during double limb support as weight acceptance begins (from initial contact of the leading limb to toe-off of the trailing limb). The XcoM is the vertical projection of the CoM in the direction of its velocity. If the CoM reaches the boundary of the CoP and continues to have an outwardly directed velocity vector, a person must take a step to increase the base of support or a fall results.

From the clinic to the laboratory, three methods for quantifying walking balance have been described in a rather hierarchical manner, from basic locations of the feet to inclusion of the more body-centric CoM and its velocity. Yet, their collective potential to examine balance in the SCI population has not been explored. Thus, the primary purpose of this study was to investigate walking balance through spatial foot parameters, foot placement relative to the CoM, and MoS in persons post-SCI. We aimed to understand how the variability of these measures differed between persons with SCI and healthy controls as well as how the variability differed within those groups. Additionally, we sought to bridge the laboratory and clinical interpretations of balance to identify the utility of spatial foot parameters (both magnitude and variability) as potential clinical correlates for more complex balance tools that incorporate CoM and XcoM. Three main hypotheses were established. First, we hypothesized that persons with SCI would demonstrate significantly different variability in all measures compared to healthy individuals. Second, within the healthy group, the variability among spatial foot
parameters would be similar to MoS variability as well as to the variability of foot placements relative to the pelvis CoM in their respective direction. However, those with SCI would exhibit significant within group differences among the same measures. Third, we anticipated that both the magnitude and variability of spatial foot parameters would show a weak positive association with foot placements relative to the CoM in their respective directions as well as to MoS in persons with SCI; yet, a strong positive association would be detected in healthy individuals.

To evaluate these hypotheses, we specifically chose to study those individuals who could walk at least a few steps without devices or assistance, even if they usually walk with a walker or cane, for the following reasons. For those individuals with SCI who are able to walk in any form and prevent themselves from falling, outcomes could provide a window into biomechanically how they are able to accomplish the task. Variability differences between and within SCI and healthy groups were anticipated due to disruptions of central nervous system circuitry post-SCI. Although the spinal circuitry for stepping propagation would be intact in the individuals examined for this study, the cortical and subcortical modulatory control responsible for posture and equilibrium would be interrupted (Jacobs and Horak 2007a). With the balance deficits and lack of distal limb control observed in the SCI population, step length and width may exhibit great variability whereas the foot location relative to a moving CoM may be more tightly regulated from one step to the next. This regulation could be the underlying reason why falls are avoided for some individuals.
Methods

Participants

A convenience sample of 10 individuals with chronic, incomplete SCI (6 males; mean age=42.6 years, SD=14.2) (Table 5-1) and 10 healthy persons comprising a control group (3 males; mean age=56.1 years, SD=3.3) participated in this cross-sectional study. Participants with SCI were involved in a larger randomized controlled trial with the following inclusion criteria: 1) at least 18 years of age, 2) injury sustained at least 6 months prior to the study, 3) upper motor neuron, motor incomplete spinal cord lesion, 4) ability to ambulate at least 10m with or without an assistive device, and 5) injury of traumatic or non-traumatic origin, excluding those of congenital etiology. The subset of individuals selected for this study from the larger trial included those who could generate at least three steps without an assistive device, BWS, or physical assistance. The healthy adult controls were a convenience sample from a larger ongoing cross-sectional study. All controls were 18 years of age or older and full-time ambulators without assistive devices or physical assistance. All experimental procedures were conducted at the Brain Rehabilitation Research Center, Malcom G. Randall Veteran Affairs Medical Center in Gainesville, Florida. Each participant signed a written informed consent approved by both the VA Subcommittee for Clinical Investigation and the University of Florida Health Science Center Institutional Review Board.

Experimental Procedures

A licensed physical therapist assessed bilateral upper and lower extremity motor and sensory function in persons with SCI based on the American Spinal Injury Association (ASIA) International Standards for Neurological and Functional
Classification of Spinal Cord Injury (American Spinal Injury Association 2002). This assessment established SCI severity and categorized injuries according to the ASIA Impairment Scale (AIS). A physical therapist also assessed performance on standardized balance assessments, the Berg Balance Scale (BBS) (Berg et al. 1992) and the Dynamic Gait Index (DGI) (Shumway-Cook and Woollacott 2001), to provide further clinical descriptions of each individual.

Both participants with SCI and healthy controls had reflective marker balls and rigid body clusters positioned on specified body landmarks to acquire 3D motion data. Marker positions were based on the Vicon PlugInGait marker set (modified Helen Hayes set). All individuals wore a safety harness attached to an overhead cable and track system that was suspended from the laboratory ceiling. Walking trials lasted a maximum of 30 seconds and were performed over a split-belt instrumented treadmill (Tecmachine, Inc.). Walking practice was permitted to become accustomed to walking on the treadmill and to obtain the best possible steady-state walking speed. When comfortable, data collections commenced. Those individuals with SCI performed one walking trial during which they were instructed to walk to the best of their abilities at their self-selected (SS) treadmill speed. All walked without assistive devices or braces. In addition, although participants with SCI wore a safety harness for each trial should they stumble or fall, neither bodyweight support (BWS) nor manual assistance was provided during any trial. This testing condition was intended to capture true walking capacity post-injury. Either sitting or standing rest periods were provided as needed between bouts of activity. Additionally, healthy controls performed two separate walking trials at
0.3 and 0.6 m/s for normal comparisons to the speeds which persons with SCI elected to walk.

Data Acquisition and Processing

Twelve camera passive motion analysis (Vicon Motion Systems) and 3D ground reaction forces (GRFs) from four piezoelectric force transducers (Advanced Medical Technology, Inc.) located beneath each half of the treadmill were acquired continuously during walking trials. The split-belt treadmill system allowed collection of GRFs for each stance phase over multiple steps of the gait cycle. Raw kinematic data were collected at 100 Hz, then low-pass filtered using a fourth-order, zero-lag Butterworth filter with a 6 Hz cut-off frequency. GRFs were acquired at a sampling rate of 2000 Hz, and low-pass filtered using a fourth-order, zero-lag Butterworth filter with a 20 Hz cut-off frequency. A 13-segment musculoskeletal model was created using Visual 3D (V3D) processing that fit the model to marker trajectories. Anthropometric and inertial values defined within V3D were applied for segment modeling and segmental center-of-mass (CoM) calculations. V3D models were used to conduct inverse dynamics analyses for calculation of intersegmental joint kinetics. Custom Matlab programming (Mathworks, Inc.) was developed to calculate the outcome measures described below.

Outcome Measures

Means were calculated using individual step data in a walking trial to indicate the average magnitude of each outcome for each person. Values for both right and left legs were entered into calculations for controls as well as those with SCI. Variability across each person’s walking trial was calculated using standard deviations.

Spatial and foot placement parameters. Two spatial foot parameters and two foot placement parameters were calculated. The two spatial foot parameters were used
to indicate the mediolateral (ML) and anteroposterior (AP) distances of one foot relative to the contralateral foot. The two foot placement parameters were used to indicate the ML and AP distances of the foot relative to the “falling” body (Balasubramanian et al. 2010). Specifically, these four measures were defined as the following distances at each initial contact: 1) leading foot CoM to trailing foot CoM in the ML direction (“Step width”) and in the AP direction (“Step length”) and 2) leading foot CoM to pelvis CoM in the ML direction (“ML foot placement”) and in the AP direction (“AP foot placement”) (Figure 5-1).

Margin of stability (MoS). Based on Hof and colleagues’ definition (2005), MoS was calculated as the shortest perpendicular ML distance between the CoP and the extrapolated center-of-mass (XcoM) during double limb support (Figure 5-3).

1. \[ X_{\text{coM}} = |(x + (v/\omega))|, \] where \( x = \text{CoM position}, \ v = \text{CoM velocity}, \ \omega = \sqrt{(\text{gravity}/\text{leg length})} \)

2. \[ \text{MoS} = |u_{\text{max}} - X_{\text{coM}}|, \] where \( u_{\text{max}} = \text{boundary of CoP} \)

A larger MoS is consistent with increased dynamic stability as a greater boundary exists between the maximum XcoM and the CoP. Furthermore, the size of the margin is proportional to the impulse that would be required to destabilize or unbalance a person. That is, by “adding” a critical amount of CoM velocity (i.e. impulse) in the direction of the nearest CoP, an unstable situation can be created.

Data Analysis

All statistical analyses were conducted using SAS 9.13 software. Means and standard deviations were calculated for all biomechanical measures. Standard deviations (i.e. variability) of data for each person with SCI were standardized using standard differences from the control group walking at a matched speed. Thus, the
control group mean equaled zero, variance equaled one, and SCI values beyond ± 2 standardized scores were considered outliers. Speed matching was established as follows: SCI ≤ 0.3 m/s matched to controls at 0.3 m/s; and SCI > 0.3 m/s to ≤ 0.6 m/s matched to controls at 0.6 m/s. Visual analysis of SCI standardized data indicated heterogeneity in the sample with variability in both positive and negative directions relative to controls. Therefore, in order to avoid masking true differences in the SCI sample and regressing SCI data toward a mean value, which might demonstrate no difference from controls, absolute values were calculated to determine a participant’s distance from a central point of zero for each outcome. The mean of all participants’ absolute values was used for group comparisons of variability outcomes between the SCI group and healthy individuals using a permutation test. Using non-standardized data, a paired t-test examined differences between measures of variability within the SCI group and within the control group at 0.3 and again at 0.6 m/s. Also using non-standardized data, Pearson’s correlations investigated the relationships between mean magnitudes of each measure as well as between their variability in the SCI group and in controls at both treadmill speeds. Alpha level for significance was set at .05.

Results

Variability of Outcomes: SCI versus Controls (Hypothesis 1)

Of the five outcome measures, participants with SCI displayed significantly different variability from controls as hypothesized (p ≤ 0.007). Figures 5-4a-c show standardized data for individual participants, which illustrate direction and magnitude of differences from a control mean of zero, in addition to the various combinations and degrees of variability across participants. The majority of participants exceeded +2 standardized scores for step width, step length, and ML and AP foot placement.
variability; in comparison, fewer participants with SCI exhibited MoS variability outside of this value. Values beyond two standardized scores demonstrate greater movement variability compared to the variability exhibited by most healthy individuals. However, it should be noted that there were individuals with SCI in our sample (e.g. SCI3 in all five measures) who demonstrated negative deviations from the control mean. Negative values indicate decreased variability, but no participants presented with such limited variability that values occurred below -2 standardized scores.

**Variability of Outcomes within Participant Group (Hypothesis 2)**

Figures 5-5a-c illustrate within group differences in outcome variability. Both the SCI group and controls walking at 0.3 and 0.6 m/s revealed significantly greater variability in step length compared to in AP foot placement ($p \leq 0.001$). Each group also exhibited significantly greater variability in step width compared to MoS ($p \leq 0.05$).

Variability of step width was significantly greater than the ML foot placement for controls walking at 0.6 m/s ($p = 0.0006$) and was only marginally significant at 0.3 m/s ($p = 0.073$). However, no significant difference was detected in this comparison for the SCI group ($p = 0.126$). Step length was significantly more variable than MoS for both those with SCI ($p = 0.004$) and controls at 0.3 m/s ($p = 0.027$), but not for controls walking at the faster speed ($p = 0.276$). These findings are partially contrary to our hypothesis as controls demonstrated significant differences between some measures of variability and the SCI group did not reach significance between some measures.

**Associations Among Outcome Magnitudes by Participant Group (Hypothesis 3)**

Tables 5-2 to 5-4 highlight the associations between mean outcome magnitudes within participant groups. The SCI group only exhibited two significant positive relationships: step width with ML foot placement ($r = 0.901, p = 0.004$) and step length with
AP foot placement \((r=0.641, p=0.042)\). Controls walking at 0.3 m/s showed the same two relationships as the SCI group, but additionally revealed a significant inverse association between step length and MoS \((r=-0.792, p=0.011)\), which are distances orthogonal to one another. Interestingly, no relationship was observed between MoS and step length when controls walked at 0.6 m/s, but step width and MoS were significantly positively related \((r=0.751, p=0.020)\). However, as seen at 0.3 m/s and also in those with SCI, the step length and AP foot placements were significantly related in the positive direction \((r=0.663, p=0.051)\); step width and ML foot placement were marginally related \((r=0.614, p=0.059)\). Controls at 0.6 m/s also showed a significant positive association between the step length and the ML foot placement \((r=0.695, p=0.038)\). As with hypothesis two, hypothesis three was only partially supported. Significant, positive associations were detected within both groups rather than solely within the control group. Moreover, although we anticipated all positive relationships across measures, regardless of group, one inverse relationship resulted.

**Associations Among Outcome Variability by Participant Group (Hypothesis 3)**

Tables 5-5 to 5-7 present associations among mean outcome variability for the SCI group and for the control group at the two different treadmill speeds. Unlike the outcome magnitudes in which only two associations were noted, the SCI group displayed significant positive relationships between spatial parameter variabilities and all other measures. This was true regardless of direction and included the association between step width and step length themselves \((r \geq 0.802, p \leq 0.005)\). The relationship between MoS and step length variability demonstrated a trend toward significance \((r=0.597, p=0.068)\). Fewer associations were detected within the control group. At 0.3 m/s, only the step length and AP foot placement variability were significantly correlated.
 Controls walking at 0.6 m/s also showed the same significant relationship as they did at the slower speed ($r=0.804, p=0.009$), but additionally exhibited a significant correlation for step width and ML foot placement ($r=0.632, p=0.050$). The analysis partially supported our hypothesis for associations among outcome variability as was noted similarly for outcome magnitudes. Both persons with SCI and controls at both speeds exhibited significant positive relationships across measures as well as weaker, non-significant associations.

**Discussion**

**Spinal Cord Injury Alters Movement Variability**

To our knowledge, this is the first study to explore the intrinsic capacity for walking balance (i.e. without assistive devices) in persons with SCI using variability as an indicator of balance. Understanding the way individuals avoid falling by challenging them to engage paretic trunk and limb muscles is compatible with the current shift in neurorehabilitation toward recovery of function. Our comparison of persons post-SCI and healthy individuals revealed that an injury significantly changes the variability of step width, step length, ML and AP foot placements relative to the CoM, and MoS from normal values. Regardless of whether measurements were recorded simply between feet or whether they included the CoM and its velocity, variability across steps was abnormal after injury. Although variability was primarily higher than normal for most persons with SCI, a select few demonstrated variability lower than the control mean.

A debate exists in the scientific literature attempting to discern between “good” or “bad” movement variability (van Emmerik and van Wegen 2002). This discussion extends to the literature on balance (also referred to as stability) with efforts to determine if low variability equates with stability and high variability with instability or
vice versa (van Emmerik and van Wegen 2002). Research of physiological responses such as heart rhythms indicates that a certain degree of variability is normal and a lack of variability, or conversely too much variability, is pathologic (Glass 2001). The same interpretation of variability can be applied to walking balance. Balance is based similarly on underlying physiological mechanisms such as afferent feedback loops; only the result is motor output of the head/trunk/pelvis and extremities (Horak 2006). While a certain level of variability is inherent in healthy individuals, research has described relationships of high and low stepping variability with balance deficits in elderly populations (Granata and Lockhart 2008) and those with certain neurological disorders, such as normal pressure hydrocephalus (Stolze et al. 2000), respectively. The analysis selected to evaluate our first hypothesis allowed for consideration of a normal range of variability (i.e. that even healthy individuals would not have perfectly consistent steps across a walking trial). By standardizing each person with SCI to the healthy group and controlling for treadmill speed, deviations from a normal range (i.e. beyond ±2 standardized scores) were detected.

In order to control CoM motion for walking balance, Dietz (2002) states that the afferent inputs weighted and selected by the central nervous system must meet the equilibrium requirements of that task. Furthermore, since SCIs interrupt the flow of sensory feedback to the supraspinal centers required for integration of information responsible for balance (e.g. brain stem, cerebellum, motor cortex) (Macpherson et al. 1997), the ability of the nervous system to appropriately select and utilize sensory information may be impaired. As a result, the motor output may contain errors, which continually will require corrections. Additionally, Barbeau et al. (1999) reported several
sensorimotor factors post-SCI that impact walking recovery, including balance.

Muscular weakness and dyscoordination as well as hyperactive spinal reflexes were proposed factors. The presence of neuromuscular impairments such as these could feasibly alter the ability of an individual to produce a consistent series of steps, thus creating differences in variability from normal.

**Variability Differences within Groups**

In addition to differences observed between SCI and healthy groups, differences also were shown within groups across measurements of variability. Regardless of group (and regardless of speed for controls), spatial foot parameters revealed significantly greater variability than measures involving the CoM and MoS, although the specific measures showing these differences varied depending on the group. Because these findings were based upon steps without devices during which persons with SCI were capable of staying upright without falling suggests that these individuals maintained balance control via more consistent foot placements relative to the CoM, particularly the XcoM, than via foot-to-foot distances. Healthy individuals responded similarly to those with SCI, even though as indicated by the between-groups comparison, persons with SCI possessed more variable motion than normal overall. Consequently, it may not always be possible to simply examine one measure of variability and presume that variability is occurring to a comparable degree in another measure. For example, the amount of variability in step width across a person’s walking trial may not possess the same amount of variability in MoS. Based on our results, this specific example would be true for persons with SCI at their SS speeds as well as for healthy individuals walking at slower than normal speeds of 0.3 and 0.6 m/s. Individuals exhibit more variability in their
step widths than in their MoS, thus implying that the CoM motion provides us with different information about balance.

**Variability May Be a Better Clinical Correlate for Balance than Magnitude**

Our investigation also sought to understand the potential of spatial parameters to function as clinical correlates for control of CoM motion. Based on our findings, we suggest that outcome variability is a more robust quantification of balance control than magnitude. Outcome magnitudes showed strong positive associations, which were common among the SCI group and controls alike. In particular, outcomes measured along the same plane (step width and ML foot placement as well as step length and AP foot placement) demonstrated these commonalities. For all groups, these relationships may be one strategy by which individuals regulate control of the CoM. The size of spatial parameters further reflects the magnitude of the respective AP or ML foot placement. These results are partially consistent with literature correlating measures of foot-to-foot distances with foot placement relative to the CoM in other populations (Hof et al. 2007). Persons post-stroke who produced asymmetric step lengths also placed their feet asymmetrically relative to their pelvic CoM (Balasubramanian et al. 2010); however, in contrast to findings in those with stroke, our sample of persons with SCI also showed that step width and ML foot placement relative to the pelvic CoM were significantly positively related. Since the magnitude relationships seen in our study were similar across SCI and control groups, the challenge remained to determine why and quantify how persons with SCI were unstable as reflected on standardized balance tests and assistive devices used for walking support (Table 5-1). Our analysis of variability began to address this challenge.
While literature exists that correlates the magnitudes of measures similar to those in our study (Hof et al. 2007; Balasubramanian et al. 2010), correlations of variabilities have not been examined until now. We showed that measures of variability exhibited a more extensive set of significant correlations for persons with SCI than for healthy controls at 0.3 or 0.6 m/s. While controls at either speed showed only one or two strong correlations, which incidentally were among the same measures as those seen in magnitudes, persons with SCI revealed strong positive associations in all measures, with the exception of a marginal significance for step length and MoS. These associations were even present between step widths and step lengths themselves. The wide array of associations, irrespective of direction, suggests that individuals with SCI avoid falling via a continuous pattern of step adaptations within a walking trial. A sequence of recovery steps occurs in healthy persons also when the lower extremities are perturbed. Purportedly, this recovery strategy is due to the nervous system’s automatic control to recapture CoM within the base of support (Oddsson et al. 2004). If each step in a person with SCI produces its own perturbation caused by step error, individuals could be attempting to correct for errors one step after the next. Yet, the series of steps may be quite different in the combination of step lengths and step widths. On the contrary, the lack of multiple significant associations across variability measures in controls may indicate that they have the ability to control parameters in different directions independently (i.e. AP foot placement and step width).

**Suggestions for Clinical Translation**

Using our analysis of relationships among outcomes of variability, a clinician could plausibly measure the variability of step widths and step lengths in a patient’s walking pattern and cautiously draw some general conclusions about the patient’s balance.
abilities (assuming the individual is walking at a slow speed comparable to those that were studied here, i.e. 0.6 m/s or slower). If the patient’s level of variability is abnormal in a particular spatial foot parameter, then based on our correlation analysis, the amount of variability for measures including control of the CoM and XcoM is most likely abnormal and similarly increased or decreased. These abnormalities would suggest walking balance deficits and could be examined over time. Because of time constraints in the clinical environment, performing calculations and comparisons with the literature is frequently not possible. Therefore, adapting equipment that collects spatial parameters, such as a GaitRite walkway or the more recently evolving inertial sensors, and modifying software to include literature-based normal values of variability could assist clinicians in deriving quick interpretations about a patient’s balance. At this time, additional studies are necessary that 1) evaluate controls at a greater range of speeds, 2) evaluate larger samples of both controls and persons with SCI, and 3) determine specific cut-off values of normal variability for step widths and lengths to create a more efficient and clinically friendly interpretation of this measurement tool.

**Further Clinical Considerations within a Recovery-Based Framework**

Quantifying walking balance following SCI is essential to understand progression of an individual’s function over time. However, the interpretation of the measurement tools implemented needs to be considered in light of the conditions under which an individual was evaluated. A unique feature of this study was the testing environment utilized to investigate walking balance. By having individuals with SCI walk in a safety harness while on a treadmill, they were afforded the opportunity to explore any immediate balance responses without the constraint of an assistive device. Thus, this testing condition creates a recovery- or activity-based evaluation environment, which
parallels the rehabilitation paradigm shift toward activity-based therapies (Dromerick et al. 2006). This is in contrast to conventional evaluation and treatment approaches, which allow an individual to compensate for impairments with assistive devices and produce different movement patterns than they would if devices were removed and upper extremities were unloaded. Leroux and colleagues (2006) also examined individuals with SCI without their assistive devices during walking on a treadmill in a harness. While they assessed the ability of an individual to adapt to walking on a treadmill at different inclines, our intentions were to examine balance responses during level ground walking and only to the perturbations induced by a person’s individual walking pattern (i.e. without external perturbations or environmental changes during the task). Given this testing environment, the findings presented in this study should be carefully considered when attempting to translate to a clinical arena without such equipment capabilities. The potential exists that outcomes might have been different if individuals were walking with their customary assistive devices or overground rather than on a treadmill. However, in remaining consistent with the paradigm shift in neurorehabilitation, the outcomes presented here complement the transition toward activity-based therapies and can assist in understanding the effects of these therapeutic strategies.

**Conclusions**

Whether examining only spatial foot parameters or foot placement relative to CoM and XcoM, persons after SCI exhibit a significantly different amount of variability (usually greater) compared to normal levels of variability. However, it appears that even in the presence of this variability, individuals retain the ability to stay upright and avoid falling via a continuous pattern of adaptive foot placement strategies. Future
investigations should continue to examine the variability of movements that individuals possess post-SCI as an indicator of balance control during walking. Understanding the repertoire of balance strategies in this population will assist clinicians in targeting therapies that address specific balance deficits.
Table 5-1. Participant demographics

<table>
<thead>
<tr>
<th>Participant</th>
<th>Age (yrs)</th>
<th>Sex</th>
<th>Injury site</th>
<th>Time post SCI (mos)</th>
<th>Assistive device</th>
<th>LEMS (max:50)</th>
<th>AIS</th>
<th>BBS (max:56)</th>
<th>DGI (max:24)</th>
<th>Self-selected treadmill speed (m/s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>SCI1</td>
<td>45</td>
<td>M</td>
<td>C5-6</td>
<td>10</td>
<td>RW</td>
<td>43</td>
<td>D</td>
<td>46</td>
<td>17</td>
<td>0.3</td>
</tr>
<tr>
<td>SCI2</td>
<td>55</td>
<td>M</td>
<td>C4</td>
<td>45</td>
<td>SPC</td>
<td>45</td>
<td>D</td>
<td>31</td>
<td>14</td>
<td>0.25</td>
</tr>
<tr>
<td>SCI3</td>
<td>48</td>
<td>F</td>
<td>C5</td>
<td>25.5</td>
<td>SPC</td>
<td>46</td>
<td>D</td>
<td>51</td>
<td>12</td>
<td>0.3</td>
</tr>
<tr>
<td>SCI4</td>
<td>26</td>
<td>M</td>
<td>T3-4</td>
<td>11</td>
<td>RW</td>
<td>40</td>
<td>D</td>
<td>21</td>
<td>15</td>
<td>0.15</td>
</tr>
<tr>
<td>SCI5</td>
<td>66</td>
<td>M</td>
<td>C7</td>
<td>78</td>
<td>SPC</td>
<td>49</td>
<td>D</td>
<td>48</td>
<td>17</td>
<td>0.5</td>
</tr>
<tr>
<td>SCI6</td>
<td>47</td>
<td>F</td>
<td>C4</td>
<td>6.5</td>
<td>RW</td>
<td>43</td>
<td>D</td>
<td>19</td>
<td>12</td>
<td>0.12</td>
</tr>
<tr>
<td>SCI7</td>
<td>40</td>
<td>F</td>
<td>T2-3</td>
<td>11</td>
<td>RW</td>
<td>38</td>
<td>D</td>
<td>10</td>
<td>9</td>
<td>0.2</td>
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<td>SCI8</td>
<td>21</td>
<td>M</td>
<td>C6</td>
<td>7</td>
<td>RW</td>
<td>45</td>
<td>D</td>
<td>17</td>
<td>8</td>
<td>0.2</td>
</tr>
<tr>
<td>SCI9</td>
<td>27</td>
<td>M</td>
<td>T6</td>
<td>12</td>
<td>RW</td>
<td>40</td>
<td>D</td>
<td>12</td>
<td>11</td>
<td>0.03</td>
</tr>
<tr>
<td>SCI10</td>
<td>51</td>
<td>F</td>
<td>C4-5</td>
<td>7.5</td>
<td>SPC</td>
<td>45</td>
<td>D</td>
<td>42</td>
<td>14</td>
<td>0.25</td>
</tr>
</tbody>
</table>

LEMS: Lower Extremity Motor Score, BBS: Berg Balance Scale, DGI: Dynamic Gait Index, RW: Rolling Walker, SPC: Single Point Cane
Figure 5-1. Spatial foot parameters using feet center-of-mass (CoM) as body reference points. White arrows represent (A) step width and (B) step length. This representative motion analysis graphic depicts SCI10 walking without a device with a left forefoot initial contact.

Figure 5-2. Foot placement relative to center-of-mass (CoM) distances. Body reference points are the leading foot CoM and the pelvis CoM. White arrows represent (A) mediolateral foot placement, (B) anteroposterior foot placement. This image depicts SCI10 as in Figure 5-1.
Figure 5-3. Raw center-of-pressure (CoP), center-of-mass (CoM), and extrapolated center-of-mass (XcoM) trajectories. The vertical black line connecting the peak of the XcoM with the CoP represents the margin of stability (MoS) for a single step. The gray vertical line extending beyond the MoS distance illustrates the time in the gait cycle (end of double limb stance) when the shortest MoS occurred for that particular step.
Figure 5-4. Variability of standardized outcomes for individual participants with SCI. Values represent direction of deviation from the control mean of zero (variance=1). (A) Step width and mediolateral (ML) foot placement; (B) Step length and anteroposterior (AP) foot placement; (C) Margin of stability (MoS). (*) indicates significant difference from controls at $p \leq 0.05$. 

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Figure 5-5. Variability of outcome measures by participant group. (A) SCI at self-selected speeds; (B) Controls at 0.3 m/s and (C) at 0.6 m/s. ML: mediolateral, AP: anteroposterior, MoS: Margin of stability. (*) indicates trend toward significance. (**) indicates significant difference at $p \leq 0.05$. 
Table 5-2. Associations between mean outcome magnitudes for participants with SCI at SS speeds.

<table>
<thead>
<tr>
<th>Outcome</th>
<th>Step width</th>
<th>Step length</th>
</tr>
</thead>
<tbody>
<tr>
<td>r</td>
<td>--</td>
<td>-0.241</td>
</tr>
<tr>
<td>Step width p</td>
<td>--</td>
<td>0.502</td>
</tr>
<tr>
<td>r</td>
<td>-0.241</td>
<td>--</td>
</tr>
<tr>
<td>Step length p</td>
<td>0.502</td>
<td>--</td>
</tr>
<tr>
<td>ML Foot placement r</td>
<td>0.901</td>
<td>-0.088</td>
</tr>
<tr>
<td>p</td>
<td>**0.0004</td>
<td>0.808</td>
</tr>
<tr>
<td>AP Foot placement r</td>
<td>-0.155</td>
<td>0.649</td>
</tr>
<tr>
<td>p</td>
<td>0.669</td>
<td>**0.042</td>
</tr>
<tr>
<td>r</td>
<td>0.392</td>
<td>-0.072</td>
</tr>
<tr>
<td>MoS p</td>
<td>0.262</td>
<td>0.844</td>
</tr>
</tbody>
</table>

MoS: margin of stability, ML: mediolateral, AP: anteroposterior. (*) indicates trend toward significance. (**) indicates significant difference within group at p ≤ 0.05.

Table 5-3. Associations between mean outcome magnitudes for healthy controls at 0.3 m/s.

<table>
<thead>
<tr>
<th>Outcome</th>
<th>Step width</th>
<th>Step length</th>
</tr>
</thead>
<tbody>
<tr>
<td>r</td>
<td>--</td>
<td>-0.167</td>
</tr>
<tr>
<td>Step width p</td>
<td>--</td>
<td>0.668</td>
</tr>
<tr>
<td>r</td>
<td>-0.167</td>
<td>--</td>
</tr>
<tr>
<td>Step length p</td>
<td>0.668</td>
<td>--</td>
</tr>
<tr>
<td>ML Foot placement r</td>
<td>0.772</td>
<td>0.242</td>
</tr>
<tr>
<td>p</td>
<td>**0.015</td>
<td>0.531</td>
</tr>
<tr>
<td>AP Foot placement r</td>
<td>-0.172</td>
<td>0.942</td>
</tr>
<tr>
<td>p</td>
<td>0.657</td>
<td>**0.0001</td>
</tr>
<tr>
<td>r</td>
<td>0.513</td>
<td>-0.792</td>
</tr>
<tr>
<td>MoS p</td>
<td>0.158</td>
<td>**0.011</td>
</tr>
</tbody>
</table>

Table 5-4. Associations between mean outcome magnitudes for healthy controls at 0.6 m/s.

<table>
<thead>
<tr>
<th>Outcome</th>
<th>Step width</th>
<th>Step length</th>
</tr>
</thead>
<tbody>
<tr>
<td>r</td>
<td>--</td>
<td>0.157</td>
</tr>
<tr>
<td>Step width p</td>
<td>--</td>
<td>0.687</td>
</tr>
<tr>
<td>r</td>
<td>0.157</td>
<td>--</td>
</tr>
<tr>
<td>Step length p</td>
<td>0.687</td>
<td>--</td>
</tr>
<tr>
<td>ML Foot placement r</td>
<td>0.614</td>
<td>0.695</td>
</tr>
<tr>
<td>p</td>
<td>*0.059</td>
<td>**0.038</td>
</tr>
<tr>
<td>AP Foot placement r</td>
<td>0.423</td>
<td>0.663</td>
</tr>
<tr>
<td>p</td>
<td>0.256</td>
<td>**0.051</td>
</tr>
<tr>
<td>r</td>
<td>0.751</td>
<td>0.014</td>
</tr>
<tr>
<td>MoS p</td>
<td>**0.020</td>
<td>0.972</td>
</tr>
</tbody>
</table>
Table 5-5. Associations between outcome variabilities for participants with SCI at SS speeds.

<table>
<thead>
<tr>
<th>Outcome</th>
<th>Step width</th>
<th>Step length</th>
</tr>
</thead>
<tbody>
<tr>
<td>Step width</td>
<td>r 0.802</td>
<td>p **0.005</td>
</tr>
<tr>
<td>Step length</td>
<td>r 0.802</td>
<td>p **0.005</td>
</tr>
<tr>
<td>ML Foot placement</td>
<td>r 0.888</td>
<td>p **0.001</td>
</tr>
<tr>
<td></td>
<td>r 0.857</td>
<td>p **0.002</td>
</tr>
<tr>
<td></td>
<td>r 0.915</td>
<td>p **0.0002</td>
</tr>
<tr>
<td>MoS</td>
<td>p **0.0002</td>
<td>p *0.068</td>
</tr>
</tbody>
</table>

MoS: margin of stability, ML: mediolateral, AP: anteroposterior. (*) indicates trend toward significance. (**) indicates significant difference within group at p≤0.05.

Table 5-6. Associations between outcome variabilities for healthy controls at 0.3 m/s.

<table>
<thead>
<tr>
<th>Outcome</th>
<th>Step width</th>
<th>Step length</th>
</tr>
</thead>
<tbody>
<tr>
<td>Step width</td>
<td>r 0.234</td>
<td>p 0.545</td>
</tr>
<tr>
<td>Step length</td>
<td>r 0.234</td>
<td>p 0.545</td>
</tr>
<tr>
<td>ML Foot placement</td>
<td>r 0.459</td>
<td>p 0.214</td>
</tr>
<tr>
<td></td>
<td>r 0.277</td>
<td>p 0.471</td>
</tr>
<tr>
<td></td>
<td>r 0.044</td>
<td>p 0.910</td>
</tr>
<tr>
<td>MoS</td>
<td>p 0.910</td>
<td>p 0.215</td>
</tr>
</tbody>
</table>

Table 5-7. Associations between outcome variabilities for healthy controls at 0.6 m/s.

<table>
<thead>
<tr>
<th>Outcome</th>
<th>Step width</th>
<th>Step length</th>
</tr>
</thead>
<tbody>
<tr>
<td>Step width</td>
<td>r -0.169</td>
<td>p 0.665</td>
</tr>
<tr>
<td>Step length</td>
<td>r -0.169</td>
<td>p 0.665</td>
</tr>
<tr>
<td>ML Foot placement</td>
<td>r 0.632</td>
<td>p **0.050</td>
</tr>
<tr>
<td></td>
<td>r -0.172</td>
<td>p 0.658</td>
</tr>
<tr>
<td></td>
<td>r 0.274</td>
<td>p 0.475</td>
</tr>
<tr>
<td>MoS</td>
<td>p 0.475</td>
<td>p 0.552</td>
</tr>
</tbody>
</table>
CHAPTER 6

EXPERIMENT 3: DIFFERENTIAL EFFECTS OF MANUAL-ASSISTED VERSUS ROBOTIC-ASSISTED LOCOMOTOR TRAINING ON DYNAMIC STABILITY POST-SCI

Individuals with spinal cord injuries (SCI) often possess disruptions in dynamic stability, defined as the ability to control the center-of-mass during walking (Barbeau et al. 2006; Brotherton et al. 2007a). In an effort to prevent falls from dynamic stability dysfunction and yet promote mobility, clinicians treating patients with SCI commonly prescribe assistive devices to provide support and/or to compensate for impairments in the trunk and lower extremities. Unfortunately, use of assistive devices post-SCI reduces lower extremity muscle activity, increases upper extremity weight bearing and consequently induces trunk flexion (Melis et al. 1999). Furthermore, removal of these devices reveals chronic postural limitations, as the trunk appears to adapt to a flexed position over time. (Leroux et al. 2006). Maintenance of such abnormal postures may prevent acquisition of normal balance strategies, particularly given evidence that healthy persons who voluntarily adopt a flexed trunk during walking exhibit altered muscle synergy patterns, ground reaction forces, kinematics and kinetics (Grasso et al. 2000). Therefore, therapies are necessary that can address functional recovery of dynamic stability rather than emphasize compensatory strategies and induce chronic deficits by prolonged training with assistive devices.

Locomotor training (LT) using a bodyweight support (BWS) and treadmill system may be one rehabilitation technique that provides an optimal environment for persons with SCI to maximize functional recovery of dynamic stability (Barbeau 2003; Behman et al. 2005; Dobkin et al. 2006). Removal of upper extremity weightbearing, and in turn, promotion of loading through the lower extremities, upright posture and appropriate stepping kinematics, allows individuals to achieve a sensorimotor experience consistent
with normal bipedal walking (Dietz and Harkema 2004; Behrman et al. 2005; Wirz et al. 2005; Behrman and Harkema 2007; Dietz 2008). At this time, two primary LT environments are available: 1) manual-assisted LT (MLT) with trainers at the lower extremities, pelvis, and trunk physically facilitating the best possible walking pattern through sensory cueing and 2) robotic-assisted LT (RLT) with an exoskeleton and computer system driving the walking pattern. Both training scenarios allow for an intense, repetitive stepping experience as they attempt to engrain the nervous system with the constituents of walking through principles of activity-dependent plasticity (Kleim and Jones 2008).

MLT creates an environment of increased walking variability due to the physical fatigue and frequent rotation of trainers on a bout-to-bout and day-to-day basis. Different trainers applying variable forces and amounts of assistance does not assure reproducibility of where the foot is placed or how the pelvis is rotated or laterally shifted (Galvez et al. 2007). Although variability may develop, motor learning theory purports that such a training scenario should increase retention and transfer of walking to real-world contexts (Riolo-Quinn 1999). Conversely, RLT promotes an environment in which a computer system creates consistent and relatively symmetrical stepping based on hip and knee range parameters established for the exoskeleton motion. Trochanteric pads immobilize the pelvis to prevent rotation or shifting, trunk straps enable an upright posture as needed, and toe lifters are available to dorsiflex the ankle passively when active motion is minimal or absent. The lateral bulk of the robotic exoskeleton may hinder a natural arm swing normally observed in healthy walking, whereas trainers may physically facilitate arm swing with bilateral, horizontal poles in the MLT setting.
(Behrman et al. 2005; Hidler et al. 2009). Although the robotic system was developed with the intention of simulating the same training situation as in the MLT environment, the need exists to understand whether these two environments do indeed have the same training effect. Awareness of each environment’s benefits in altering the walking stability capacity of an individual’s nervous system has potential to direct his/her rehabilitation plan more effectively and efficiently.

Therefore, the purpose of this study was to examine the differential effects of MLT versus RLT on dynamic stability during walking in persons with chronic, motor incomplete SCI. More specifically, we aimed to understand how closely persons with SCI approximated normal balance abilities before they trained and how each mode of LT shifted those abilities relative to normal walking. Since recovery of dynamic stability is the goal, normal walking is the reference by which recovery can be assessed. We hypothesized that compared to a healthy control group, persons with SCI prior to their respective LT intervention would exhibit significant differences in measures of dynamic stability based on the literature describing a high falls incidence and balance deficits in this population (Brotherton et al. 2007b). However, post-training, those who underwent RLT would show outcomes similar to control values, whereas those who underwent MLT would continue to demonstrate significant differences. We anticipated these outcomes because RLT provides a consistent walking pattern for individuals with SCI who are newly learning to walk without their assistive devices. Since functional walking requires one to resist both internally-generated and externally-provided perturbations, we believe an individual might need to learn how to consistently resist their own internal perturbations (e.g. neuromuscular errors resulting from treadmill speed or bodyweight
support changes), prior to being introduced to a variable environment that creates external perturbations (e.g. variability in trainer forces). Also, testing was conducted on a level ground treadmill without assistive devices or bodyweight support. Since external perturbations were not introduced during testing (i.e. only “internal” perturbations were present), those who went through RLT would have been trained in such a context. Finally, because of the anticipated injury heterogeneity among participants, any differences hypothesized before and after LT were expected to present bi-directionally (i.e. more or less of a given motion) relative to controls.

Methods

Participants
Ten individuals with chronic, motor incomplete SCI (6 males; mean age=42.6 years, SD=14.2, AIS D) (Table 6-1) and 10 healthy persons comprising a control group (3 males; mean age=56.1 years, SD=3.3) were recruited to participate in this study. Participants with SCI were involved in a larger randomized controlled trial with the following inclusion criteria: 1) at least 18 years of age, 2) injury sustained at least 6 months prior to the study, 3) upper motor neuron, motor incomplete spinal cord lesion, 4) ability to ambulate at least 10m with or without an assistive device, and 5) injury of traumatic or non-traumatic origin, excluding those of congenital etiology. The subset of individuals selected for this study from the larger trial included those who could generate at least three steps without an assistive device, BWS, or physical assistance. The healthy adult controls were a convenience sample from a larger ongoing cross-sectional study. All controls were 18 years of age or older and full-time ambulators without assistive devices or physical assistance. All experimental procedures were conducted at the Brain Rehabilitation Research Center, Malcom G. Randall Veteran
Affairs (VA) Medical Center in Gainesville, Florida. Each participant signed a written informed consent approved by both the VA Subcommittee for Clinical Investigation and the University of Florida Health Science Center Institutional Review Board.

**Locomotor Training Intervention**

Participants with SCI were randomized to the MLT or RLT environment as part of the parent clinical trial. Five participants trained via the manual intervention and five via the robotic intervention. Training occurred 5 days per week for 45 sessions total. At least one licensed physical therapist was present for each training session. The intervention goal was for participants to achieve 30 minutes of quality stepping per session; however, the length and number of individual bouts necessary to obtain this goal varied daily depending on each participant’s functional ability and trainer fatigue. Rest periods, typically standing with assistance for posture and with BWS lowered to promote loading, were provided between stepping bouts. Vital signs were obtained before, after, and at regular intervals during training sessions to assess participants’ physiological responses to exercise and ensure possible episodes of autonomic dysreflexia or postural hypotension were avoided or addressed as necessary. Over the 45 sessions, participants were challenged with progression of training parameters while maintaining optimal stepping patterns, including an upright trunk. In particular, BWS was decreased, treadmill speed increased and either robotic guidance force or trainer manual assistance decreased. Over the course of nine weeks, the MLT group increased mean treadmill speeds from 0.71 m/s (SD=0.14) to 1.0 m/s (SD=0.15) and number of training bouts changed from 8.43 (SD=2.51) to 9.27 (SD=9.74). The RLT group increased mean treadmill speeds from 0.69 m/s (SD=0.06) to 0.87 m/s (SD=0.11) and number of training bouts changed from 4.71 (SD=1.16) to 4.30 (SD=1.43). BWS
decreased in the MLT group from 35.98% (SD=8.01) to 18.29% (SD=8.77), while the RLT group decreased BWS from 34.84% (SD=4.70) to 13.24% (SD=3.91).

**Experimental Procedures**

The following sequence of experimental procedures occurred prior to initiating the LT protocol. A licensed physical therapist assessed bilateral upper and lower extremity motor and sensory function in persons with SCI based on the American Spinal Injury Association (ASIA) International Standards for Neurological and Functional Classification of Spinal Cord Injury (American Spinal Injury Association 2002). This assessment established SCI severity and categorized injuries according to the ASIA Impairment Scale (AIS). A physical therapist also assessed performance on standardized balance assessments, the Berg Balance Scale (BBS) (Berg et al. 1992) and the Dynamic Gait Index (DGI) (Shumway-Cook and Woollacott 2001), to provide further clinical descriptions of each individual’s ability to balance.

The following motion analysis collections were conducted pre- and post-LT for persons with SCI and once for healthy controls. Both participants with SCI and healthy controls had reflective marker balls and rigid body clusters positioned on specified body landmarks to acquire 3D motion data. Marker positions were based on the Vicon PlugInGait marker set (modified Helen Hayes set). All individuals wore a safety harness attached to an overhead cable and track system that was suspended from the laboratory ceiling. Walking trials lasted a maximum of 30 seconds and were performed over a split-belt instrumented treadmill (Tecmachine, Inc.). Walking practice was permitted to become accustomed to walking on the treadmill and to obtain the best possible steady-state walking speed. When comfortable, data collections commenced. Those individuals with SCI performed one walking trial during which they were
instructed to walk to the best of their abilities at their self-selected (SS) treadmill speed. All walked without assistive devices or braces. In addition, although participants with SCI wore a safety harness for each trial should they stumble or fall, neither BWS nor manual assistance was provided during any trial. This testing condition was intended to capture true walking capacity post-injury. Either sitting or standing rest periods were provided as needed between bouts of activity. Additionally, healthy controls performed three separate walking trials at randomly introduced treadmill speeds: 0.3, 0.6, and 0.9 m/s. These speeds were selected to represent a likely range within which persons with SCI would elect to walk for normal, speed-matched comparisons.

**Data Acquisition and Processing**

Twelve camera passive motion analysis (Vicon Motion Systems) and 3D ground reaction forces (GRFs) from four piezoelectric force transducers (Advanced Medical Technology, Inc.) located beneath each half of the treadmill were acquired continuously during walking trials. The split-belt treadmill system collected GRFs for each stance phase over multiple steps of the gait cycle. Raw kinematic data were collected at 100 Hz, then low-pass filtered using a fourth-order, zero-lag Butterworth filter with a 6 Hz cut-off frequency. GRFs were acquired at a sampling rate of 2000 Hz, and low-pass filtered using a fourth-order, zero-lag Butterworth filter with a 20 Hz cut-off frequency. A 13-segment musculoskeletal model was created using Visual 3D (V3D) processing that fit the model to marker trajectories. Anthropometric and inertial values defined within V3D were applied for segment modeling and segmental center-of-mass (CoM) calculations. V3D models were used to conduct inverse dynamics analyses for calculation of intersegmental joint kinetics. Custom Matlab programming (Mathworks, Inc.) was developed to calculate the outcome measures described below.
Dynamic Stability Biomechanical Outcome Measures

Primary outcome

**Center-of-mass trajectory length.** Motion of the whole body CoM, particularly in the transverse plane, is an indicator of dynamic stability during walking as the CoM must be controlled within a continually moving base of support or a fall will result (Winter 1995). Normal CoM movement over several gait cycles follows a smooth, sinusoidal path that varies minimally from one stride to the next. Thus, the mean length of this path over a series of strides as well as its variability from stride to stride relative to normal values provides a gauge of how well an individual is controlling his/her CoM. For each stride in a walking trial, the length of the whole body CoM trajectory in the transverse plane was calculated and normalized to stride length.

Secondary outcomes

**Trunk excursions.** Rotational movements of the trunk have been used to quantify balance strategies in elderly with and without balance deficits in response to walking perturbations (Grabiner et al. 2008). Similarly, in this study 3D trunk angular displacement ranges were determined for each step in a walking trial. Trunk displacements were defined as the angles of rotation about each axis. The established reference axes were positioned orthogonal to one another with the x-axis directed mediolaterally, the y-axis anteroposteriorly, and the z-axis vertically. Proximal and distal ends of the modeled trunk segment were used to calculate movement about the x- and y-axes, while markers positioned on bilateral acromion processes were used to calculate rotation about the z-axis. Both mean displacement ranges and variability over a walking trial were calculated in each direction.
Spatial foot parameters. Two spatial parameters were calculated separately for
each step: step width and step length. Step width was defined as the mediolateral
distance between the heel marker of the leading foot at initial contact and the heel
marker of the trailing limb at that same point in time. Step length was defined as the
anteroposterior distance between the heel marker of the leading foot at initial contact
and the heel marker of the trailing limb at that same point in time (Perry 1992). Both
means and variability of each parameter over a walking trial were calculated.

Data Analysis

All statistical analyses were conducted using SAS 9.13 software. Outcomes for
each person with SCI were standardized using standard differences from the control
group walking at a speed similar to each person’s self-selected speed. Thus, the control
group mean equaled zero, variance equaled one, and SCI values beyond ± 2
standardized scores were considered outliers. Speed matching was established as
follows: SCI ≤ 0.3 m/s matched to controls at 0.3 m/s; SCI >0.3 m/s to ≤ 0.6 m/s
matched to controls at 0.6 m/s; and SCI >0.6 m/s to ≤ 0.9 m/s matched to controls at 0.9
m/s. Visual analysis of standardized data demonstrated heterogeneity in the direction of
outcome measures relative to controls. Therefore, in order to avoid masking true
differences in the SCI sample and regressing data toward a mean value, which might
demonstrate no difference from controls, absolute values were calculated to determine
a participant’s distance from a central point of zero. The mean of all participants’
absolute values were used for group comparisons using a permutation test. Alpha level
for significance was set at .05.
Results

Pre-Locomotor Training Biomechanical Outcomes

Mean magnitude of outcomes. When self-selecting treadmill speeds for biomechanical testing prior to intervention, all participants elected to walk at extremely slow speeds with comparable speeds observed in both groups (RLT: mean=0.23 m/s, SD=0.1; MLT: mean=0.25 m/s, SD=0.2) (Table 6-2). Although we had hypothesized both groups overall would demonstrate differences from controls in all outcomes, the RLT group was significantly different only in CoM trajectory length and trunk rotation ($p \leq 0.05$). However, all outcomes were significantly different from controls in the MLT group ($p \leq 0.01$) with the exception of trunk rotation ($p=0.111$) (Table 6-3).

Variability of outcomes. All outcome measures except trunk rotation and step length in the RLT group showed significantly different variability from controls ($p \leq 0.05$). The MLT group demonstrated significantly different variability in all outcome measures before training ($p \leq 0.01$) (Table 6-4).

Post-Locomotor Training Biomechanical Outcomes

Mean magnitude of outcomes. Post-LT, all participants in the RLT group and three out of five in the MLT group elected to walk at higher treadmill speeds than during pre-testing; on the whole, both groups improved speed similarly (RLT $\Delta = +0.25$ m/s; MLT $\Delta = +0.21$ m/s) (Table 6-2). In the RLT group, CoM trajectory length was no longer significantly different from controls indicating an overall shift to within a normal distribution as originally hypothesized ($p=0.09$), yet trunk rotation remained significantly different ($p \leq 0.01$). Moreover, mediolateral trunk excursions, step length, and step width altered to become significantly different from controls ($p \leq 0.05$), even though these measures indicated normal movements prior to training. As anticipated, the MLT group
exhibited significant differences from controls in all outcomes post-intervention ($p \leq 0.01$), which includes a shift in trunk rotation outside of the normal distribution, although it had been normal pre-training (Table 6-3). Figure 6-1 illustrates individual pre-post data comparisons for all participants. Although group analyses show overall deviations from a central point of zero, the individual data depict the heterogeneity of responses within the sample including directional and magnitude differences.

Variability of outcomes. Contrary to our hypotheses, the RLT group post-training remained significantly different in variability from controls in the same outcomes as pre-RLT ($p \leq 0.0001$) except trunk rotation still showed no difference ($p = 0.988$). Conversely, step length variability changed such that it became significantly different ($p \leq 0.0001$) even though it was not prior to RLT. As expected, the MLT group continued to display significantly different levels of variability from controls in all outcomes ($p \leq 0.0001$) (Table 6-4). Figure 6-2 highlights the individual participant changes in variability following either RLT or MLT.

Discussion

This study is the first to examine the impact of LT environments on dynamic stability recovery post-SCI. Restitution of walking function for those with incomplete SCI as opposed to sole utilization of spared musculature is consistent with our current rehabilitation paradigm shift. To date, the activity-based LT literature using manual-assisted and/or robotic-assisted environments for humans with SCI has emphasized primarily the generation of stepping patterns or overground walking with customary assistive devices as outcomes (Behrman and Harkema 2000; Field-Fote and Tepavac 2002; Wirz et al. 2005; Nooijen et al. 2009). Our study expanded the focus of walking recovery in this population to address dynamic stability as another essential walking
subtask (Barbeau 2003). This subtask has been studied only in animal models of SCI (Howland et al. 1995; Bolton and Misiaszek 2009; Karayannidou et al. 2009) and in healthy and other human patient populations, such as persons with lower extremity amputations and peripheral neuropathy (Dingwell et al. 2000; Hof et al. 2007).

Additionally, we chose to evaluate dynamic stability during walking in a novel manner on a treadmill. Evaluation in this environment paralleled the task-specific, recovery-based interventions implemented. This aspect of the evaluation was intended to detect more purely what each intervention specifically trained (i.e. if particular aspects of stability that were trained in each environment translated to walking without assistance). This is in contrast to the testing in the overground environment with assistive devices, which was not component of the training. Removal of all assistance, including BWS and devices, allowed individuals to reveal the ways in which their nervous systems solved the problem of dynamic stability.

**MLT and RLT Both Trained Dynamic Stability Post-SCI**

The effectiveness of each person’s solution to dynamic stability manifests in the control of his/her CoM, i.e. “the balance point of the body” (Winter 1995). Therefore, the improved or maintained control of the CoM trajectory lengths relative to normal in eight of ten SCI participants suggests that both the manual-assisted and robotic-assisted environments have benefits for training dynamic stability. The significant differences from normal values that remained in the MLT group as a whole may be a result of the more impaired CoM control that this group presented with pre-LT. In addition to pre-training deficits, the number of training sessions in which participants engaged may have impacted the degree of change in CoM control. Currently, the optimal parameters for LT, including duration and frequency, are unknown and vary widely among studies.
Field-Fote and Tepavac 2002; Hidler et al. 2009). The training protocol for this study was established at 45 training sessions. If persons in the MLT group possessed greater impairments initially, but their trends toward improvement were too small to be considered “normal,” the number of sessions may have been the limiting factor.

**Adaptive Movement Strategies Developed to Maintain Dynamic Stability Post-LT**

Secondary outcomes of trunk excursions and spatial parameters demonstrated heterogeneity among and within groups as well; in many cases, individuals who initially showed normal values shifted beyond normal post-LT, even if their CoM trajectory exhibited improvements. Post-hoc video analysis confirmed the apparent generation of new adaptive balance strategies and exaggerated movements (e.g. increased arm swing resulting in increased trunk rotation). Our laboratory previously observed this development of increased arm swing in persons with SCI at slow treadmill speeds immediately post-MLT and several months beyond, even though healthy controls possessed much lower amplitude arm swing at the same speeds (NJ Tester, unpublished data). Thus, our current findings suggest that amplified trunk rotation may be propagated by increased arm swing (or vice versa). Literature in healthy individuals supports this linkage between arm swing and upper trunk rotation, which move concomitantly to counteract pelvic rotation when walking speeds become increasingly normal (van Emmerik and Wagenaar 1996; Wagenaar and van Emmerik 2000). Moreover, Pijnappels et al. (2010) reported enhanced asymmetric arm swinging with associated increases in trunk rotation as a recovery strategy when healthy subjects were tripped during walking. Interestingly, the most dramatic representation of excessive trunk rotation was evident in four of five persons trained in the MLT environment (SCI 6, 7, 9, 10), with the fifth individual in the group (SCI 8) demonstrating
a trend toward the same pattern. Unlike the RLT environment that hinders arm swing with the bulk of its exoskeleton, MLT promotes arm swing. At times, trainers provide arm swing deliberately using parallel poles to assist the alternating arm motion. Thus, the arm swing emphasized in MLT may be responsible for the prominent trunk rotation observed.

One additional, potential explanation for those individuals who displayed excessive trunk motion as well as increased variability among step lengths and step widths is that post-testing often was the first opportunity to attempt walking without BWS or a device. Up until this point, with the exception of pre-testing conditions, BWS was provided during training or customary assistive devices were used outside of the study. Both afford external, supportive influences (Melis et al. 1999) and ultimately were removed post-testing. This testing condition destabilized the body, which then had to achieve stabilization using strategies such as increased trunk movements and/or altered foot placements. An intriguing factor is that the majority of participants from both groups also demonstrated a preference for higher treadmill speeds at post-testing, likely influenced by walking at closer to normal speeds during training in a similar treadmill environment. However, most individuals never had the opportunity to practice walking at those higher speeds without BWS or devices. Thus, the exaggerated trunk motion and variability in stepping parameters perhaps reflects the person’s immediate balance responses to manage their increase in functional capacity and preference for faster walking speeds.

The variability of step length and width among individuals in both training groups requires additional attention based on their pattern of outcomes. Some individuals in the manual-assisted group, who demonstrated values outside of normal pre-LT, shifted in
the direction of normal; however, several individuals in the RLT group who showed more normal pre-training values became more variable after training. Improvements in degree of variability in the MLT group suggest retention and motor learning, which the variability of this specific intervention is believed to promote. Furthermore, MLT allowed for occasional bouts of “independence” in which the speed was decreased, BWS increased if needed, and body segments one at a time permitted to move without trainer assistance (although other body segments continued to have assistance). In contrast, although BWS and speed were altered in RLT, the computerized exoskeleton provided a relatively error-free experience, lacking opportunities to adapt to errors in foot placement. These findings are consistent with recent literature in the stroke population that also compared the manual-assisted and robotic-assisted interventions (Lewek et al. 2009). Lewek et al. reported improved coordination and kinematic consistency at the hip and knee following four weeks of training in the manual-assisted, but not in the robotic-assisted environment. In addition, similar studies have been conducted in the animal literature. Cai et al. (2006) found that spinal mice produced more consistent stepping patterns following robotic-assisted training with an “assist-as-needed” capability than with robotic-assisted training using fixed stepping trajectories. Furthermore, Edgerton and Roy (2009) indicate that after minutes of repetitive stepping along the same path with no variation, the nervous system becomes less responsive to sensory stimuli for stepping and muscle activation diminishes.

**AIS Categorization Does Not Reflect Dynamic Stability**

Of final note are the variations in dynamic stability outcomes in consideration of participants’ AIS categorizations. Following a physical therapist’s assessment of upper and lower extremity sensory and motor function, each person included in this study had
SCIs classified as AIS D and all with lower extremity motor scores $\geq 38$ out of a possible 50. Previous literature regarding the AIS suggested that this classification system, and specifically the motor score, correlated well with walking function post-SCI (Waters et al. 1994). Thus, individuals with injuries categorized as AIS D purportedly have the highest degree of walking ability. In this study, both primary and secondary measures of dynamic stability during walking revealed great heterogeneity, not only following interventions, but also prior to training. Pre-LT, participants in the MLT group were significantly different from controls in five measures of mean magnitude and all measures of variability, while the RLT group was only different in two measures of mean magnitude and four measures of variability. Therefore, clinicians and researchers should reflect upon the value of grouping all persons together solely with this classification system when evaluating dynamic stability. Tests of isolated, muscle strength in a supine may not fully characterize the requirements necessary for dynamic stability in a weight bearing position. Alternative means of stratifying individuals on this walking subtask, perhaps via standardized balance assessments, may be necessary ultimately to direct clinicians toward the most effective intervention strategy.

**Conclusions**

Both MLT and RLT demonstrated benefits for training dynamic stability post-SCI with the majority of participants exhibiting increased CoM control. However, the evolution of adaptive strategies, resulting at the trunk and feet, differed between interventions and within participants. Although values for trunk motion and spatial parameters at the feet often shifted outside of normal ranges post-LT, even if individuals demonstrated normal values pre-training, they continued to increase in walking speed overall. Thus, the seemingly “abnormal” movements may have been solutions to
maintain stability as speed capacity increased. Future work should investigate methods of categorizing persons post with SCI in alternative ways beyond AIS prior to examining dynamic stability. Additionally, investigations into optimal training parameters, such as duration or frequency, for achieving dynamic stability in either the MLT or RLT environment would be beneficial in directing both researchers and clinicians.
<table>
<thead>
<tr>
<th>Participant</th>
<th>Training group</th>
<th>Age (yrs)</th>
<th>Sex</th>
<th>Injury site</th>
<th>Time post SCI (mos)</th>
<th>Assistive device</th>
<th>LEMS (max:50)</th>
<th>AIS</th>
<th>BBS (max:56)</th>
<th>DGI (max:24)</th>
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<tr>
<td>SCI1</td>
<td>RLT</td>
<td>45</td>
<td>M</td>
<td>C5-6</td>
<td>10</td>
<td>RW</td>
<td>43</td>
<td>D</td>
<td>46</td>
<td>17</td>
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<td>55</td>
<td>M</td>
<td>C4</td>
<td>45</td>
<td>SPC</td>
<td>45</td>
<td>D</td>
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<td>14</td>
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<td>M</td>
<td>T3-4</td>
<td>11</td>
<td>RW</td>
<td>40</td>
<td>D</td>
<td>21</td>
<td>15</td>
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<td>RLT</td>
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<td>F</td>
<td>C4</td>
<td>6.5</td>
<td>RW</td>
<td>43</td>
<td>D</td>
<td>19</td>
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<td>SPC</td>
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<td>D</td>
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<td>48</td>
<td>F</td>
<td>C5</td>
<td>25.5</td>
<td>SPC</td>
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<td>D</td>
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<td>12</td>
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<td>M</td>
<td>C7</td>
<td>78</td>
<td>SPC</td>
<td>49</td>
<td>D</td>
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<td>17</td>
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<td>MLT</td>
<td>40</td>
<td>F</td>
<td>T2-3</td>
<td>11</td>
<td>RW</td>
<td>38</td>
<td>D</td>
<td>10</td>
<td>9</td>
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<td>M</td>
<td>C6</td>
<td>7</td>
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<td>M</td>
<td>T6</td>
<td>12</td>
<td>RW</td>
<td>40</td>
<td>D</td>
<td>12</td>
<td>11</td>
</tr>
</tbody>
</table>

Mean (SD) | 44.8 (11.2) | 16.0 (16.3) | 43.2 (2.0) | 31.8 (12.1) | 14.4 (1.8) |

SCI6 | SCI7 | SCI8 | SCI9 | SCI10 | Mean (SD) | 40.4 (17.8) | 26.7 (29.5) | 43.6 (4.5) | 27.6 (20.2) | 11.4 (3.5) |

Table 6-2. SCI self-selected treadmill speeds before and after intervention

<table>
<thead>
<tr>
<th>Participant</th>
<th>Training group</th>
<th>Treadmill speed (m/s)</th>
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<tbody>
<tr>
<td>SCI1</td>
<td>RLT</td>
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<td>SCI2</td>
<td>RLT</td>
<td>0.25</td>
<td>0.4</td>
<td></td>
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<tr>
<td>SCI3</td>
<td>RLT</td>
<td>0.15</td>
<td>0.25</td>
<td></td>
</tr>
<tr>
<td>SCI4</td>
<td>RLT</td>
<td>0.12</td>
<td>0.3</td>
<td></td>
</tr>
<tr>
<td>SCI5</td>
<td>RLT</td>
<td>0.25</td>
<td>0.7</td>
<td></td>
</tr>
<tr>
<td>SCI6</td>
<td>MLT</td>
<td>0.3</td>
<td>0.9</td>
<td></td>
</tr>
<tr>
<td>SCI7</td>
<td>MLT</td>
<td>0.5</td>
<td>0.9</td>
<td></td>
</tr>
<tr>
<td>SCI8</td>
<td>MLT</td>
<td>0.2</td>
<td>0.2</td>
<td></td>
</tr>
<tr>
<td>SCI9</td>
<td>MLT</td>
<td>0.2</td>
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<td></td>
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<tr>
<td>SCI10</td>
<td>MLT</td>
<td>0.03</td>
<td>0.15</td>
<td></td>
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</tbody>
</table>

Mean (SD)    0.21 (0.1) 0.46 (0.2)

RLT: Robotic-assisted Locomotor Training, MLT: Manual-assisted Locomotor Training
<table>
<thead>
<tr>
<th>Training Group</th>
<th>Testing Session</th>
<th>Variable</th>
<th>Mean</th>
<th>SD</th>
<th>Mean(ABS)</th>
<th>p-value</th>
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<tbody>
<tr>
<td>RLT</td>
<td>Pre</td>
<td>CoM trajectory length</td>
<td>0.633</td>
<td>1.518</td>
<td>1.383</td>
<td>*0.027</td>
</tr>
<tr>
<td></td>
<td>Post</td>
<td></td>
<td>0.144</td>
<td>1.606</td>
<td>1.606</td>
<td>0.091</td>
</tr>
<tr>
<td></td>
<td>Pre</td>
<td>Trunk ML</td>
<td>0.712</td>
<td>1.366</td>
<td>1.059</td>
<td>0.166</td>
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<tr>
<td></td>
<td>Post</td>
<td></td>
<td>2.258</td>
<td>1.638</td>
<td>2.258</td>
<td>*&lt;0.0001</td>
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<td>0.771</td>
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<td></td>
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<td>0.705</td>
<td>0.733</td>
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<tr>
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<tr>
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<td>2.294</td>
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<tr>
<td></td>
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<tr>
<td></td>
<td>Post</td>
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<td>0.619</td>
<td>2.760</td>
<td>2.310</td>
<td>*&lt;0.0001</td>
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<tr>
<td></td>
<td>Pre</td>
<td>CoM trajectory length</td>
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<td>1.777</td>
<td>*0.002</td>
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<tr>
<td></td>
<td>Post</td>
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<td>1.116</td>
<td>1.514</td>
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<tr>
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<td>Trunk ML</td>
<td>1.566</td>
<td>1.516</td>
<td>1.566</td>
<td>*0.007</td>
</tr>
<tr>
<td></td>
<td>Post</td>
<td></td>
<td>4.426</td>
<td>2.189</td>
<td>4.426</td>
<td>*&lt;0.0001</td>
</tr>
<tr>
<td></td>
<td>Pre</td>
<td>Trunk AP</td>
<td>1.385</td>
<td>1.243</td>
<td>1.667</td>
<td>*0.004</td>
</tr>
<tr>
<td></td>
<td>Post</td>
<td></td>
<td>1.902</td>
<td>2.12</td>
<td>2.065</td>
<td>*0.0001</td>
</tr>
<tr>
<td></td>
<td>Pre</td>
<td>Trunk Rotation</td>
<td>1.125</td>
<td>0.988</td>
<td>1.136</td>
<td>0.111</td>
</tr>
<tr>
<td></td>
<td>Post</td>
<td></td>
<td>2.855</td>
<td>0.917</td>
<td>2.855</td>
<td>*&lt;0.0001</td>
</tr>
<tr>
<td></td>
<td>Pre</td>
<td>Step length</td>
<td>-1.409</td>
<td>2.593</td>
<td>1.869</td>
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</tr>
<tr>
<td></td>
<td>Post</td>
<td></td>
<td>-0.501</td>
<td>2.503</td>
<td>1.650</td>
<td>*0.004</td>
</tr>
<tr>
<td></td>
<td>Pre</td>
<td>Step width</td>
<td>1.398</td>
<td>4.318</td>
<td>3.966</td>
<td>*&lt;0.0001</td>
</tr>
<tr>
<td></td>
<td>Post</td>
<td></td>
<td>-1.379</td>
<td>3.897</td>
<td>2.876</td>
<td>*&lt;0.0001</td>
</tr>
</tbody>
</table>

RLT: Robotic-assisted Locomotor Training, MLT: Manual-assisted Locomotor Training, CoM: center-of-mass, ML: mediolateral, AP: anteroposterior, SD: standard deviation, Mean(ABS): mean of absolute values representing overall distance from control mean of zero. (*) indicates significantly different from controls at $p \leq 0.05$. 
Table 6-4. Variability of standardized outcomes separated by intervention

<table>
<thead>
<tr>
<th>Training Group</th>
<th>Testing Session</th>
<th>Variable</th>
<th>Mean</th>
<th>SD</th>
<th>Mean(ABS)</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>RLT</td>
<td>Pre</td>
<td>CoM trajectory length</td>
<td>1.662</td>
<td>1.990</td>
<td>1.772</td>
<td>*0.002</td>
</tr>
<tr>
<td></td>
<td>Post</td>
<td></td>
<td>4.256</td>
<td>3.554</td>
<td>4.256</td>
<td>*&lt;0.0001</td>
</tr>
<tr>
<td></td>
<td>Pre</td>
<td>Trunk ML</td>
<td>0.611</td>
<td>2.084</td>
<td>1.426</td>
<td>*0.021</td>
</tr>
<tr>
<td></td>
<td>Post</td>
<td></td>
<td>3.524</td>
<td>3.497</td>
<td>3.653</td>
<td>*&lt;0.0001</td>
</tr>
<tr>
<td></td>
<td>Pre</td>
<td>Trunk AP</td>
<td>1.136</td>
<td>2.870</td>
<td>2.166</td>
<td>*&lt;0.0001</td>
</tr>
<tr>
<td></td>
<td>Post</td>
<td></td>
<td>1.990</td>
<td>1.803</td>
<td>1.990</td>
<td>*&lt;0.0001</td>
</tr>
<tr>
<td></td>
<td>Pre</td>
<td>Trunk Rotation</td>
<td>0.416</td>
<td>0.926</td>
<td>0.723</td>
<td>0.582</td>
</tr>
<tr>
<td></td>
<td>Post</td>
<td></td>
<td>0.107</td>
<td>0.404</td>
<td>0.286</td>
<td>0.988</td>
</tr>
<tr>
<td></td>
<td>Pre</td>
<td>Step length</td>
<td>0.618</td>
<td>0.790</td>
<td>0.772</td>
<td>0.514</td>
</tr>
<tr>
<td></td>
<td>Post</td>
<td></td>
<td>2.278</td>
<td>2.830</td>
<td>2.374</td>
<td>*&lt;0.0001</td>
</tr>
<tr>
<td></td>
<td>Pre</td>
<td>Step width</td>
<td>2.927</td>
<td>2.011</td>
<td>2.927</td>
<td>*&lt;0.0001</td>
</tr>
<tr>
<td></td>
<td>Post</td>
<td></td>
<td>3.315</td>
<td>2.248</td>
<td>3.490</td>
<td>*&lt;0.0001</td>
</tr>
<tr>
<td>MLT</td>
<td>Pre</td>
<td>CoM trajectory length</td>
<td>6.498</td>
<td>9.169</td>
<td>7.756</td>
<td>*&lt;0.0001</td>
</tr>
<tr>
<td></td>
<td>Post</td>
<td></td>
<td>3.688</td>
<td>3.853</td>
<td>3.779</td>
<td>*&lt;0.0001</td>
</tr>
<tr>
<td></td>
<td>Pre</td>
<td>Trunk ML</td>
<td>3.658</td>
<td>3.179</td>
<td>3.718</td>
<td>*&lt;0.0001</td>
</tr>
<tr>
<td></td>
<td>Post</td>
<td></td>
<td>4.379</td>
<td>4.917</td>
<td>4.410</td>
<td>*&lt;0.0001</td>
</tr>
<tr>
<td></td>
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<td>Trunk AP</td>
<td>5.749</td>
<td>6.412</td>
<td>5.749</td>
<td>*&lt;0.0001</td>
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<tr>
<td></td>
<td>Post</td>
<td></td>
<td>6.908</td>
<td>8.433</td>
<td>7.552</td>
<td>*&lt;0.0001</td>
</tr>
<tr>
<td></td>
<td>Pre</td>
<td>Trunk Rotation</td>
<td>1.752</td>
<td>1.655</td>
<td>1.768</td>
<td>*0.002</td>
</tr>
<tr>
<td></td>
<td>Post</td>
<td></td>
<td>2.242</td>
<td>1.955</td>
<td>2.293</td>
<td>*&lt;0.0001</td>
</tr>
<tr>
<td></td>
<td>Pre</td>
<td>Step length</td>
<td>4.728</td>
<td>5.334</td>
<td>4.728</td>
<td>*&lt;0.0001</td>
</tr>
<tr>
<td></td>
<td>Post</td>
<td></td>
<td>2.503</td>
<td>3.759</td>
<td>2.528</td>
<td>*&lt;0.0001</td>
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<tr>
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<td>Pre</td>
<td>Step width</td>
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<td>8.503</td>
<td>7.632</td>
<td>*&lt;0.0001</td>
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<tr>
<td></td>
<td>Post</td>
<td></td>
<td>4.507</td>
<td>7.008</td>
<td>4.891</td>
<td>*&lt;0.0001</td>
</tr>
</tbody>
</table>

RLT: Robotic-assisted Locomotor Training, MLT: Manual-assisted Locomotor Training, CoM: center-of-mass, ML: mediolateral, AP: anteroposterior, SD: standard deviation, Mean(ABS): mean of absolute values representing overall distance from control mean of zero. (*) indicates significantly different from controls at $p\leq0.05$. 

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Figure 6-1. Mean outcome magnitudes standardized to control data pre vs. post locomotor training for 10 SCI participants (separated by training group). Vertical blue shading at ±2 standardized scores indicates normal distribution about a control mean of zero.
Figure 6-2. Variability of outcome measures standardized to control data pre vs. post locomotor training for 10 SCI participants (separated by training group). Vertical blue shading at ±2 standardized scores indicates normal distribution about a control mean of zero.
CHAPTER 7
CONCLUSIONS

Experimental Limitations

When examining the results of these three experiments, limitations in methodology should be considered. First, participants with SCI exhibited a continuum of stepping ability; particularly when walking without assistive devices, which was a condition that occurred in each experiment. Consequently, participants produced varying numbers of steps for analysis. Furthermore, during pre-analysis data processing certain steps were removed secondary to noise in the acceleration profiles. Ultimately, the number of steps included in analysis therefore was limited for some participants. This may have had an impact on the standard deviations used to determine variability during a walking trial. Another avenue for increasing the number of available steps might be to eliminate only certain phases of the gait cycle with unusable data while retaining other phases.

A second limitation was the manual selection of head and pelvis acceleration thresholds as a method to eliminate “spikes” or noise in the data and to select only valid steps for analysis. The potential for human error existed in this process such that the thresholds may have been too low or high, thus inadvertently eliminating valid steps or keeping steps which should have been removed. However, only one individual conducted the manual selection, thereby creating relative consistency in the process.

A third potential confound across all three experiments was the use of a harness during walking trials. In these studies, which evaluated dynamic stability, the harness may be viewed as an “assistive device” even though no bodyweight support was given at any point. The harness surrounding the trunk and pelvis provided sensory input to the
nervous system, and the possibility exists that this input may have influenced stability. Given the safety issues and fall risks of individuals with SCI walking without their customary assistive devices, implementing a harness was necessary. However, healthy individuals across these experiments also walked in harnesses, which created a better-controlled environment for comparison among groups.

Overall Conclusions

Spinal cord injury (SCI) impairs dynamic stability during walking. As a result, individuals who sustain SCIs frequently fall and are prone to secondary injuries (Brotherton et al. 2007). Clinicians treating this population conventionally instruct patients on the use of assistive devices and residual motor function to compensate for dynamic stability deficits. More recently, however, SCI rehabilitation has been shifting away from this traditional approach toward one that encourages the use of weak, and previously believed unusable, musculature for the restitution of walking (Behrman and Harkema 2007). However, this transition in interventions also necessitates a concomitant shift in measurement strategies. Thus, the main objective of this dissertation was to systematically examine dynamic stability in persons post-SCI via a series of experiments, which reflect the SCI rehabilitation paradigm shift. Embedded within these experiments were measurement tools that quantify biomechanical movement strategies, deemed by existing literature as nervous system priorities, essential for maintaining stability.

The first experiment investigated both the influence of assistive devices (ADs) on head stability during walking post-SCI as well as the effect of injury on head stability when ADs were eliminated. These two aspects of the experiment reflect functionally compensated and uncompensated evaluations. Although results demonstrated a large
degree of heterogeneity among participants with SCI, they suggest that ADs play a role in reducing mediolateral head motion. Remarkably, when ADs were removed, individuals displayed the ability to decrease anteroposterior head motion to the extent that it was moving less than the pelvis overall. Stability in the AP direction appeared to be a priority. Furthermore, when compared to healthy individuals, those with SCI walking without ADs exhibited highly variable head displacements in all directions. This lack of consistent movement relative to normal levels of variability is purported as an indicator of impaired dynamic stability.

In the second experiment, dynamic stability during walking was examined in persons post-SCI walking without ADs using three measurement strategies: spatial foot parameters (i.e. step width and step length), anteroposterior and mediolateral foot placements relative to the pelvic center-of-mass (CoM), and margin of stability, which accounts for CoM velocity. Findings indicated that over a walking trial, persons with SCI produce greater variability in all measurements than healthy individuals. Moreover, within groups (SCI and controls), measures demonstrated different amounts of variability suggesting that these outcomes provide different kinds of information about dynamic stability, regardless of whether one has had an injury or not. Lastly, correlations between variability of spatial foot parameters and measures involving the CoM were strong and positive. These relationships may have revealed not only the continuous step-by-step adaptations that persons with SCI generate to avoid falling, but also the potential for spatial foot parameters to function as clinical correlates.

The final experiment remained within the paradigm shift toward recovery by examining the effects of activity-based walking interventions on persons post-SCI.
Specifically, we studied the differential effects of manual-assisted and robotic-assisted locomotor training (MLT and RLT) on dynamic stability after SCI to determine how interventions altered balance abilities relative to normal. Results suggested that both MLT and RLT possess training features that positively impact dynamic stability as quantified by the length of the CoM trajectory (i.e. indicative of CoM control). However, responses among individuals were quite variable, even though all were categorized by sensorimotor impairment in the same way (AIS D) prior to training. The most compelling findings were the evolution of movement strategies, such as trunk rotation, that increased beyond normal values, particularly after MLT. These movements appeared to be the result of motions including arm swing and pelvic rotation, which are emphasized in the MLT environment. Regardless of intervention approach, individuals post-SCI demonstrated newly generated adaptive balance strategies which appear consistent with each environment’s training properties and that further allowed for simultaneous increases in treadmill speed.

**Summary and Future Work**

Following SCI, a subset of the population demonstrates the ability to maintain dynamic stability during walking when assistive devices are removed. However, the manner in which they move may differ from normal and also may differ from other persons with SCI. The three experiments presented here illustrate the variety of solutions individuals generate at the head, trunk, pelvis and feet to avoid falling. Since the same individuals were included as participants across all three studies, a collective picture of each person’s balance ability can be gained from the different measurement tools implemented. Another avenue for analysis with this current data set would be to examine the relationships between biomechanical outcome measures from these
experiments and clinical measures of balance/falls (i.e. Berg Balance Scale and Dynamic Gait Index), which were collected as well. Future studies should continue to evaluate measures of dynamic stability, including those utilized in these studies, in larger samples of individuals with SCI. In doing so, alternative classification strategies for participants may need to be considered beyond AIS classification only. Additional measures of dynamic stability might include a biomechanical evaluation of the upper extremities, muscle synergies in the upper and lower limbs, or neurophysiological testing, such as H-reflexes, to understand their roles in dynamic stability with this population. Additionally, small accelerometers that attach to body segments might be useful tools to explore dynamic stability in the laboratory and be feasible for use in the clinic as well, thus easing translation of information from the literature to the clinic. Investigations should continue to evaluate dynamic stability post-SCI within the framework of activity-based therapies and restitution of pre-injury walking. The continuation of treadmill-based evaluations would be consistent with this framework and would allow clinicians to both examine and train a patient in a single environment while gaining essential information about an individual’s progress toward dynamic stability recovery during walking.
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BIOGRAPHICAL SKETCH

Kristin Alayne Vamvas Day, a native of Cincinnati, Ohio, graduated from Ohio University in 2000 with a bachelor's degree in biological sciences. A year prior to earning her undergraduate degree, she was admitted in absentia to Ohio University School of Physical Therapy and ultimately graduated with her master's degree in physical therapy in 2002. Upon passing her national licensure examination, she began practicing as a staff physical therapist in Greenville, South Carolina, at Greenville Memorial Hospital, a level one trauma center. At this facility, she gained an immense passion for the care of patients with neurologically-traumatic injuries, mentored by a team of highly experienced interdisciplinary therapists. Kristin returned to Cincinnati in 2004 and continued to practice in acute care and inpatient rehabilitation at Good Samaritan Hospital.

A self-proclaimed "aggressive and progressive" physical therapist, Kristin recognized what had evolved as the routine nature of physical therapy practice and further realized that the lack of evidence to guide decision-making may be largely responsible. Therefore, in January 2006, she returned to graduate school with the goal of advancing the science behind neurological rehabilitation. She enrolled in the Rehabilitation Science Doctoral program at the University of Florida under the mentorship of Dr. Andrea Behrman, an expert in the field of spinal cord injury rehabilitation and recovery. During her four-and-a-half years at Florida, Kristin investigated locomotor interventions targeted at walking recovery in individuals with spinal cord injury. Furthermore, her dissertation investigated an unexplored area in this patient population, measurement of walking balance recovery. Throughout her doctoral education, Kristin received full financial support through the Neuromuscular Plasticity
Training program, funded through a National Institutes of Health T32 training grant, as well as through Department of Physical Therapy teaching assistantships. She additionally received a one year Early Career Rehabilitation Research Award from the Florida Department of Health Brain and Spinal Cord Injury Program. Kristin graduated with a Doctor of Philosophy in May 2010, aspiring to translate her experiences in spinal cord injury mobility research as well as her clinical background in traumatic brain injury to explore physical treatments for persons with disorders of consciousness.