NEUROMECHANICAL AND NEUROPHYSIOLOGICAL EXAMINATION OF WALKING
WITH AN ANKLE FOOT ORTHOSIS IN NON-INJURED INDIVIDUALS AND PERSONS
WITH INCOMPLETE SPINAL CORD INJURY

By

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To Lord Ganesha and Shri Gajanan Maharaj for dwelling upon me at all times!!
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Clinicians often use orthotic devices to compensate for walking related impairments after incomplete spinal cord injury (ISCI). Orthotic devices such as an ankle foot orthosis (AFO) are commonly used to stabilize the ankle joint and aid toe clearance during walking. Compensatory stepping achieved with an AFO has led therapists to assume that such devices could be integrated in newer, neurobiologically driven, recovery-based interventions such as locomotor training (LT) for individuals with ISCI. In spite of the appeal of such compensatory strategies, their use during LT is still controversial. This is due to the lack of information about the possible effect of the device in optimizing or hindering afferent input from lower limb motion; joint, muscle and cutaneous receptors fundamental to the training. After ISCI, pattern generating neural network within the spinal cord increases its reliance on motion-related afferent input from these receptors for maintaining locomotor control. Limiting ankle excursion with an AFO may alter the interconnected limb joint assembly specific to walking and in turn influence the afferent information critical for stepping. Our study explored the therapeutic use of such devices from a walking recovery based paradigm.
The aim of this project was to investigate the mechanical and neurophysiological implications of the use of an AFO during stepping in non-injured individuals and persons with ISCI. Specifically, we examined the effect of wearing a posterior leaf spring ankle foot orthosis (PAFO) on transition phase joint kinematics and kinetics and soleus H-reflex modulation during walking. In the first experiment, we examined the transition phase mechanics with and without a PAFO in healthy, non-injured individuals. Our study identified and measured the changes that occurred in normal joint kinematics and kinetics as a result of wearing a PAFO. The results suggested that proximal hip extension; crucial for the transition from stance-to-swing and the rate of loading during the swing-to-stance phase were significantly decreased. In the second experiment, we compared transition phase mechanics observed while walking with and without the PAFO in individuals with ISCI to normal mechanics. The comparison assessed the effect of the PAFO on pre-existing stepping related deficits in individuals with ISCI and also measured deviance or likeness of the change observed in these individuals from normal. The results suggested that the use of a PAFO decreased hip extension thereby impacting the provision of at least one critical afferent input key to the restoration of walking.

In the third experiment, soleus H-reflexes were compared in non-injured individuals while walking with and without the PAFO in ten different phases of the gait cycle. The result showed that walking with the PAFO did not affect soleus H-reflex excitability in these individuals. In the fourth and final experiment, soleus H-reflexes were compared in the mid-stance and mid-swing phase in individuals with ISCI, while walking with and without the PAFO. A significant increase in the soleus H-reflex amplitude was observed in the mid-swing phase of walking. Our findings suggest that the PAFO increased afferent inflow and modulated reflex activity. However, increase in afferent input in the mid-swing phase of the gait cycle may not be favorable to
retraining the task of walking. In summary, our results suggest that walking with a minimally restrictive PAFO alters transition phase mechanics and soleus H-reflex modulation during mid-swing phase of walking. Therefore, during LT, use of a compensatory PAFO to achieve stepping may not coincide with the principles of training.
Walking is a fundamental motor function of human beings. It consists of three neurally controlled and coordinated tasks that allow 1) generation of a basic reciprocal, stepping pattern required for propulsion of the body 2) maintenance of equilibrium during propulsion and 3) adaptability of the walking pattern to the environment and to the behavioral goals of the individual.¹⁻³ Humans can move around from one place to another, maintain an upright posture, interact with the environment and perform a flurry of activities characteristic to human nature due to their ability to walk. Although walking is an essential element in daily living, its importance is usually only recognized when it is impaired or lost.

Spinal cord injury (SCI) is a debilitating condition resulting in walking impairment or inability secondary to deficits in voluntary strength (of the limbs and trunk) and sensation.⁴,⁵ Although all persons with SCI express a desire to walk only twenty five to thirty three percent of these individuals regain the ability to do so.⁶⁻⁹ Persons with incomplete spinal cord injury (ISCI) have a greater potential for walking recovery compared to individuals with complete injury due to some sparing of motor and sensory function below the level of the lesion.¹⁰ However, walking in persons with ISCI may be slow with asymmetrical steps, flexed posture and impaired balance and adaptability.

Current rehabilitation strategies after SCI are based on the assumption that deficits due to SCI are irremediable from surgical, medical, or therapeutic means.⁴,¹¹ In particular, the spinal cord, being a hard-wired conduit of information from supraspinal structures to the muscles is viewed post-injury as irreparable and the deficits permanent and irreversible.¹¹⁻¹³ To compensate for irremediable deficits post-SCI, therapists employ braces, assistive devices, and wheelchairs to
teach new behavioral strategies to achieve mobility.\textsuperscript{4,14-16} Assistive devices, such as canes, crutches, and walkers provide upright support through upper extremity weight bearing to compensate for lower limb and trunk muscle weakness. Likewise, single joint (ankle foot orthosis-AFO) or multi joint (knee ankle foot orthosis-KAFO) are used to stabilize joints and aid stepping.\textsuperscript{16} For example, posterior leaf spring ankle foot orthosis (PAFO) is often used to compensate inadequate toe clearance and loss of heel strike during stepping.\textsuperscript{16,17}

In contrast to the assumption of ceased neurological function below the level of injury, neuroscientists examining the neurobiological control of walking in both animals and humans have provided convincing evidence for recovery.\textsuperscript{18-26} Evidence suggests that the adult mammalian spinal cord is plastic and is known to reorganize after injury when provided with the appropriate stimulus that is intense, task-specific and repetitive in nature.\textsuperscript{3,22,27-29} Reciprocal stepping, for example, defined as the repetitious, mechanical sequence of limb motion and weight shift such that, one limb maintains contact with the ground and supports the body while the other limb swings forward is facilitated by a host of afferent inputs that modulate the transition from support-to-swing and vice-versa.\textsuperscript{30-32,33}

Afferent inputs for the neurobiological control of stepping involve motion related changes in joint position, muscular force and limb load sensed by joint, muscle and cutaneous receptors in the lower extremity.\textsuperscript{34-37} Evidence from animal and human studies indicate that terminal hip extension and unloading are critical sensory inputs required for the afferent initiation of the transition from support-to-swing.\textsuperscript{31,34,38} Limiting hip extension and/or transfer of weight on either limb delays the transition from support-to-swing and swing-to-support and limits forward progression of the body.
Evolving rehabilitation approaches such as locomotor training have effectively integrated the provision of afferent input required for the neural control of stepping with repeated task specific practice to retrain the ability to step in individuals with SCI.\textsuperscript{25,37,39,40} Using a treadmill, body weight support and manual assistance the training facilitates repeated cueing of critical motion related inputs including hip extension and limb unloading required for stepping.\textsuperscript{24-26,41-43} The repeated volley of motion related input via the training can drive the pattern generating body within the spinal cord, induce reorganization of the nervous system and facilitate the recovery of walking in persons with SCI having limited access to supraspinal input.\textsuperscript{25,26,44} Therefore, this training strategy personifies the intrinsic repair potential of the spinal cord which can be tapped into using appropriate physical rehabilitation strategies to facilitate recovery.

Ironically, addition of orthotic devices as training variables to the physiological-based training may hinder rather than facilitate recovery.\textsuperscript{24} Single joint, rigid orthotic devices such as AFOs might restrict the range of excursion of the distal and proximal linked joints and alter the afferent information related to joint position and load pivotal for stepping.\textsuperscript{24} Furthermore, the use of an assistive/orthotic device might change the basic stepping pattern thereby modifying the afferent information utilized for stepping. In addition, the acquisition of a new motor skill such as walking with an orthotic device could induce a new pattern of skill-dependent plasticity that could impact their ability to reacquire normal walking potential.\textsuperscript{45} Thus the role of such compensatory strategies in promoting use of the intrinsic mechanisms for stepping lack mechanical and neurophysiological evidence justifying their use and need to be investigated.\textsuperscript{24,46}

The overall purpose of this study was therefore to examine the mechanical and neurophysiological effect of walking with one such single joint, range limiting orthotic device the AFO. In this four part project, I examined the effect of wearing an AFO on 1) immediate
phase specific H-reflex modulation and 2) walking mechanics in healthy, non-injured individuals and 3) immediate phase specific H-reflex modulation and 4) walking mechanics in persons with ISCI.

In these experiments, healthy, non-injured individuals and persons with ISCI walked on an instrumented treadmill at speeds close to normal walking speeds. Kinematic, kinetic and H-reflex data was collected while these individuals walked with and without wearing an AFO. Specifically, the mechanical changes in joint position, joint powers, vertical and horizontal ground reaction forces in the transition phases were examined in these experiments accompanied by the neurophysiological changes in the soleus H-reflex amplitude. Interpretation of the mechanical and neurophysiological data served as a critical first step in interpreting how restricting lower extremity single joint excursion while walking with a AFO modified the consequent sensory information i.e. hip joint position and load required for reciprocal stepping.

In summary, these studies enhanced our understanding of the impact of altered walking mechanics associated with these compensatory devices on the neurally controlled task of reciprocal stepping in healthy, non-injured individuals and persons with SCI. The findings of this study will be useful in 1) clinical decision-making for the use of such devices in physiological-based training interventions and 2) designing neuromechanically compatible assistive and orthotic devices.

The following literature review is composed of sections that will serve to orient the reader to the foundation principles underlying the purpose of this project. An overview of SCI related walking impairment and the use of orthotic devices to compensate for impairment is described first. Juxtaposed are the neurobiological control of walking and the subtask of stepping highlighting the mechanical and neural characteristics of stepping. The newer evolving recovery
based intervention called locomotor training that has stemmed from our knowledge of the neural control of walking and plasticity of the nervous system is described next. This section is followed by the gap in knowledge pertaining to the use of an AFO from a neural control standpoint. Methodological considerations for investigation of an AFO and interpretation of results in relation to the neural control of stepping is described that will provide the basis for the experimental paradigm used in this project. The clinical and scientific relevance of the studies are discussed in the final section.
CHAPTER 2
LITERATURE REVIEW

Overview of Human Spinal Cord Injury: Consequence and Rehabilitation

Introduction to the Problem

Spinal cord injury (SCI), an injury to the neural elements within the spinal cord, results in a multitude of dysfunction, loss of sensation and motor function being the most profound. Injury could occur as a result of motor vehicular accidents, diving accidents that fractures, dislocates or compresses the vertebrae protecting it, or may also result from a gunshot or knife wound that penetrates and cuts the cord. Additionally, secondary damage usually occurs with a traumatic injury as a result of bleeding, swelling and inflammatory processes that compress the cord. Non-traumatic injuries to the cord occur as a result of tumors, vascular problems, spina bifida and several other conditions.

Although there is a significant rate of mortality associated with injury, survival after SCI has improved considerably because of efficient critical care and improved urinary rehabilitation and respiratory management. About 253,000 people currently in the US live with SCI and there are an additional 11,000 new cases every year. The current ten-year survival rate of spinal cord injured patients is approximately 86% of normal.

Consequence of Injury

The effects of SCI vary according to the level and type (complete or incomplete) of injury. In a complete injury, there is bilateral, total sensory and motor loss below the level of injury. A person with an incomplete SCI (ISCI) retains some sensation below the level of injury. Incomplete injuries are variable, and a person with such an injury may have patchy motor involvement such as he or she might be able to move one limb more than another, may be able to feel parts of the body that cannot be moved, or may have functioning on one side of the body
more than the other. Depending on the level of injury, paralysis can involve all four extremities a condition called quadriplegia or tetraplegia occurring as a result of cervical injuries or only the trunk and lower body a condition called paraplegia occurring as a result of injury at the thoracic level or below.

The American Spinal Injury Association (ASIA) impairment scale (AIS) is used to classify the level and severity of injury in relation to the loss of sensation and motor function.\textsuperscript{58,60} The scale consists of five categories that classify sensory and motor function. ASIA A is defined as a complete injury with no motor or sensory function preserved in the sacral segments S4-S5. ASIA B, C and D are all incomplete injuries but classification varies based on the level of motor involvement below the level of the lesion. ASIA B is defined as an incomplete injury with some sensory but no motor function preserved below the level of injury including the sacral segments S4-S5. To be classified as an ASIA C more than half of the muscles are graded less than 3/5 voluntary strength. ASIA D is defined as an incomplete injury with at least half of the muscles graded more than 3/5. A person with SCI is classified as an ASIA E if he/she has no neurological deficits that are detectable on a neurological examination of this type.

**Walking Potential after SCI**

Walking ability after SCI has been defined in the literature in several different ways. For example, studies have characterized it based on the ability to ambulate upright fifty feet without assistance\textsuperscript{61} or the ability to walk in the community or in the household\textsuperscript{8,9} or the ability to walk reciprocally for at least two hundred feet with/without orthotics or assistive devices\textsuperscript{62} or walking functional independence measure of $> 3/7$.\textsuperscript{63} The ability to walk ranks as one of the top five priorities of individuals with SCI based on the level and severity of injury but interestingly only 25-33% of these individuals are able to do so.\textsuperscript{52,6,64} Key predictors of ambulation potential after SCI include: ASIA score D or E at admission,\textsuperscript{9,9} age,\textsuperscript{61,64} ASIA lower extremity motor scores
greater than or equal to 10 by one month,\textsuperscript{65} manual muscle testing score in the quadriceps greater
than 2/5\textsuperscript{66} and sparing of pin prick sensation below the level of injury.\textsuperscript{62} Based on these factors, a
significant number of persons who regain the ability to walk are persons with ISCI compared to
persons with complete SCI.\textsuperscript{10,61} Maynard et al. (1979), for example, reported that of 123 patients
with incomplete sensory deficits 72 hours after injury, 47\% were ambulatory and 87\% of the
patients with incomplete motor lesions were walking at one year.\textsuperscript{67}

Although upright mobility may be spared or achieved with assistance, walking is typically
impaired in persons with ISCI as a result of varying levels of muscular paralysis, sensory
deficits, spasticity and poor trunk control.\textsuperscript{68-70} Gait in an individual with ISCI is often
characterized by one or combination of the following deviations (i) inadequate active hip
extension during stance; (ii) limited hip flexion; (iii) limited knee flexion; (iv) excess ankle
plantar flexion during swing; and (v) impaired initial foot contact.\textsuperscript{5} Consequently, these
individuals are often seen taking slow, asymmetrical and uncoordinated steps over a wide base of
support and having limited adaptability to the environment.\textsuperscript{5,71}

Rehabilitation of Individuals with ISCI

The International Classification of Functioning, Disability and Health (ICF) is the
framework developed by the World Health Organization (WHO) to describe functioning and
disability at both the individual and population levels.\textsuperscript{72} Conventional physical therapy
interventions for improving walking function in persons with ISCI targets two main domains of
the model namely, body function/structures and activities. At the level of body function and
structures, the level and severity of injury using the ASIA scores is assessed and interventions
maximizing residual muscle strength and endurance in muscles that can be voluntarily activated
above and below the lesion are implemented.\textsuperscript{14,15} Similarly, interventions at the activity level
emphasize the use of assistive and orthotic devices to improve ambulation potential\textsuperscript{14,16} and teach
new strategies for upright mobility.\textsuperscript{14,15,73} Patient performance at this level however is a critical indicator of type and extent of orthotic supports and the patient's tolerance to ambulation.

**Orthotic Devices: Rationale for Use and Prescription**

Orthotic devices range from simple single joint braces such as the ankle foot orthosis (AFO) to multi joint orthotics such as reciprocating gait orthosis (RGOs).\textsuperscript{15,74,75} The prescription of these devices varies based on several factors including the ASIA impairment score of motor complete or incomplete made by the clinician to the needs and desires of the client.\textsuperscript{74} The goals of prescribing orthotic devices for walking are to support the paralyzed or weakened musculoskeletal structure, add stability to joints, improve mobility, correct alignment and improve overall functional independence.\textsuperscript{16} Posterior leaf spring ankle foot orthosis (PAFOs) are usually prescribed for higher functioning individuals with ISCI to provide support for weakened musculature around the ankle joint.\textsuperscript{16,76} Since it supports one single joint it is considered least cumbersome and called a device of minimal assistance. The guiding principles for recommendation are to control the ankle joint by limiting excursion range, provide safe joint mechanics, prevent toe drag during the stance-to-swing transition, minimize the risk of falls and enhance the ability to walk faster and efficiently.\textsuperscript{16,17}

The current rationale for use of orthotic devices for persons with ISCI has stemmed from the hierarchical model of motor control that has been well accepted by rehabilitation practitioners and continues to serve as the basis for rehabilitation to date.\textsuperscript{11,12} The model portrays a hard-wired, immutable central nervous system (CNS) that controls all voluntary movements by sending commands from the cerebral cortex to the periphery. The spinal cord serves simply as a cable between the brain and the peripheral musculature receiving stimuli from the periphery and relaying cortical commands to the periphery. Therefore if the spinal cord is injured, the damage is considered irreparable, non-malleable and permanent.\textsuperscript{12,77} Conventional rehabilitation
therefore typically consists of a compensatory approach to deal with walking impairments in persons after ISCI. The approach utilizes the use of compensatory orthotic devices that utilize other spared abilities to accomplish the task or modify the task and/or the environment to make it easier for a person to accomplish the goal.

**Neurobiological Control of Walking**

The prevailing assumption that neural recovery is not possible following SCI has led to the aggrandizement of orthotic devices that compensate for walking impairments. Compelling evidence from neuroscience examining the neural control of walking however contradicts this assumption. Neuroscientists have investigated the role of the nervous system, particularly the spinal cord to adapt and reorganize after complete transections at the level of the cord. The ability of the spinal cord to respond to peripheral sensory input, generate and modulate rhythmic activity in the lower limbs and reorganize after injury make it a viable substrate for intervention. The subsequent sections expand on the role of the spinal cord in the control of walking and its plasticity after injury that have led to a proposed paradigm shift in SCI rehabilitation from compensation-based to a recovery-based model.

Conceptual models portray walking as three neurally controlled and coordinated tasks: 1) Generation of a basic reciprocal, stepping pattern required for propulsion of the body, 2) Maintenance of equilibrium during propulsion and 3) Adaptability of the walking pattern to the environment and to the behavioral goals of the individual. However, convergence and processing of multiple afferent inputs occurs at every level of the nervous system to bring about the smooth, patterned orchestration of several joints and muscular synergies that characterize walking. While walking, a single alpha motor neuron might receive as many as ten thousand inputs. Determining which input has a greater relative influence on sculpting the locomotor pattern has been a challenging task in the complex human nervous system. Neuroscientists have
therefore utilized reduced, non-human, vertebrate preparations such as cat models to study the influence of exclusive afferent inputs on locomotor behavior and extrapolated their research findings to human control of locomotion. The premise for extrapolation being the locomotor framework which is remarkably similar throughout the vertebrate phylum, in spite of the form of locomotion which is species specific.

For example, the basic principle of organizing rhythmic stepping behavior is similar in humans compared to other non-human vertebrates. In a spinalized cat model the reflex or automatic behavior associated with walking i.e. reciprocal stepping does not require control by cerebral cortex rather it is controlled by subcortical and spinal centers which are subject to cortical intervention. Reciprocal stepping is generated at the level of the spinal cord as long as the weight is supported and the ground is moving under their feet. Assemblies of premotor interneurons in the spinal cord are synaptically interconnected with each other and with motoneuronal pools that are capable of sustaining alternating movements required for walking. These networks of neurons and interneurons are called central pattern generators (CPGs). The central pattern generators simplify the control of locomotion by harnessing the large degrees of freedom and provide the basic framework for walking. Sensory input driven from the periphery is known to control the rhythm of walking by making the required phase transitions, shaping the pattern of activity and reinforcing ongoing activity. Sensory information from muscle spindles which are sensitive to changes in the muscle length, golgi tendon organs that are responsive to muscle tension and flexor reflex afferents involving the mechanoreceptors, cutaneous afferents and nociceptors is processed at the level of the CPG. When the task is consistent and unaltered, like during stable state walking, the spinal cord
demonstrates remarkable autonomy in producing the reflexive stepping pattern without much cortical control.

Neural descending pathways from the brain to the cord are predominantly involved in fine-tuning this pattern. For example, the corticospinal tracts influence the locomotor performance by acting monosynaptically or oligosynaptically on the alpha-motorneurons or indirectly via connections on the CPG. Thus, selective muscle control required for fine tuning and/or modulation of locomotor synergy i.e. speed of locomotion could be achieved. Also, the excitability of the CPGs is governed strongly by locomotor centers in the midbrain and brainstem i.e. mesencephalic, pontine or subthalamic locomotor regions that are under limbic and cortical control and that dictate purposeful locomotion such as starting or stopping.

Similarly, higher brain centers such as the cortex, cerebellum and basal ganglia are responsible for integrating afferent information from different sensory sources with cortical motor commands. Integration of the sensory input with the motor output facilitates control of the other two subtasks i.e. maintaining balance and adapting to the environment by engaging corrective and reflexive postural mechanisms required for walking. When the cerebellum, the brainstem and the spinal cord are spared in cats (decerebrate preparation), it is seen that the animal is able to generate rhythmic activity, support body weight and propel itself. However, the other two subtasks for successful locomotion i.e. dynamic balance and adaptability to the environment are noticeably deficient (Figure 2-1).

Dynamic balance in the decerebrate cat requires integration of sensory systems: visual, kinesthetic and vestibular system. Walking in a cluttered environment for example requires integration of information from visual, somatosensory and vestibular inputs to maintain balance and perform the task of walking successfully. The reticulospinal, rubrospinal, vestibulospinal
and corticospinal tracts\textsuperscript{111,112} are descending spinal tracts that project motor commands for fine postural adjustment to the spinal cord after processing afferent information. Corrective postural responses are then integrated and adjusted to the current state or phase of stepping by the CPG.\textsuperscript{113}

Lack of cortical control above the level of transection also leads to a loss of adaptability to the environment and to the person’s behavioral goals (Figure 2-1). The motor cortex and the basal ganglia are important for "skilled" locomotion in which the feet must be guided to establish firm contact with narrowly specified points in the environment.\textsuperscript{97} Studies on decorticate cats showed that although the loss of the cortex has minimal impact on the locomotor process, the context in which the locomotor movements are performed is affected.\textsuperscript{84,114} For example, decorticate cats exhibit limited range of options in locomotor movements and are hyperactive to stimuli that would tend to elicit a minimal response. Likewise, ablation of the caudate nucleus of the basal ganglia in cats results in the animal following anything that moves termed "compulsory approach syndrome"; while diencephalic cats (whose thalamus and hypothalamus have been removed) demonstrate "obstinate progression" i.e. walking into obstacles and not attending to environmental stimuli.\textsuperscript{114} Therefore cortical centers and the basal ganglia are important for adaptive control of movement to the environment.

While the different levels of control exerted by the nervous system are important for the coordination of the overall task of walking, the proposed studies focus on one of the subtasks of walking i.e. stepping. The next two sections elaborate on characterizing the stepping pattern mechanically and describing the neural assembly required for stepping.
Stepping Mechanics

Basic Instrumentation and Terminology

Stepping is a motor task that obeys laws governing static and dynamic bodies and can be quantified using our knowledge of basic mechanics. Kinematics is the branch of mechanics dealing with the motion of body segments without being concerned with the forces that cause the motion. Kinematic analysis, using automated motion analysis systems measures positions, angles, velocities, and accelerations of body segments and joints during motion. Joint angle (also called inter-segmental angle) is defined as the angle between the two segments on either side of the joint, usually measured in degrees. For the interconnected chain of segments involving the ankle, knee and hip joints of the lower limbs, joint angles are particularly useful in determining the relative motion of one joint with respect to the other. However, these measurements only describe the motion performed and are limited in what they can tell us about the cause of the motion.

Kinetic analysis on the other hand measures forces acting at a particular joint, segment or body as a whole that cause the specific walking pattern. Forces in walking can be internal such as muscle activity, ligamentous constraint, or external such as ground-reaction forces created from external loads. During walking concentric and eccentric contraction of the limb musculature around a joint results in the generation and absorption of mechanical energy necessary to accomplish the movement that we observe and is referred to as “joint power”. Joint power therefore is the rate at which energy is either generated or absorbed and is the product of a joint moment and the joint angular velocity. The joint moment (also known as torque or moment of force) being the rotational potential of the forces acting on a joint. The joint moment usually is calculated around a joint center. The units used to express moments or torques are Newton-meters (N-m) and for research purposes usually are normalized to the subject's body.
Joint power is generated when the moment and the angular velocity are in the same direction and said to be absorbed when they are in opposite directions. The units used to measure joint power are Watts (W).

During stepping, ground reaction forces (GRFs) produced as a result of body weight transferring onto and moving across the supporting foot constitutes the external forces acting on the body. GRF as the name denotes, is basically the reaction to the force the body exerts on the ground. GRF is comprised of three components: 1) vertical force, 2) fore-aft shear and 3) medio-lateral shear.\textsuperscript{115-117} Information on these forces is obtained from a force platform or force plate, which is a transducer set into the floor to measure the forces and torques applied by the foot to the ground. These devices provide a quantified measure of the three components of the resultant GRF vector.

Measurement of vertical ground reaction forces produced during walking provides information on the load imposed on the joints during weight bearing.\textsuperscript{33} Normally this force is represented as two peaks with a valley in between. The first peak occurs in response to weight acceptance while loading the limb. The second peak is caused by acceleration of the body forward. Fore-aft or horizontal ground reaction forces measure propulsion as the body weight shifts from one lower limb to the other. Horizontal forces have a negative and a positive component. The negative component is referred to as the braking force and is indicative of a backward horizontal friction force between the floor and the foot to prevent the foot from sliding forward. The positive component is referred to as the propulsive force and is indicative of the foot pushing back on the floor to propel the body forward. The medio-lateral forces measure stability of the body during walking.\textsuperscript{117} The exchange of body weight from one limb to the other
generate the medio-lateral shear forces. Peak medial shear occurs while loading the limb and peak lateral shear occurs while unloading the limb.

**Characterization of Stepping Pattern**

Reciprocal stepping can therefore be functionally characterized using biomechanical measures. Stepping is defined as the repetitious sequence of limb motion such that, one limb maintains contact with the ground and supports the body while the other limb advances forward. The supporting or weight bearing phase is termed as the stance phase and the limb is referred to as the "stance limb" while the forward stepping phase is termed as the swing phase and the limb is referred to as the "swing limb". A single sequence of support (stance) and advancement (swing) executed by one limb is called a “gait cycle”.

The transition from stance-to-swing is characterized by two phases; terminal stance phase and pre-swing phase. The terminal stance phase begins with the heel rise of the supporting limb and continues till the other foot strikes the ground. The pre-swing phase begins with foot strike of the other limb and continues till toe-off of the supporting limb. The objective of these phases together is the progression of the body forward. Power is generated in this phase to propel the limb and body forward. The transition from stance-to-swing is kinematically characterized by extension of the hip joint, flexion at the knee joint and plantarflexion at the ankle joint (Figure 2-2). Kinetically this phase is characterized by a horizontal propulsive force to aid forward progression of the body.

Similarly, the transition from swing-to-stance is characterized by two phases: initial contact and loading response. Initial contact begins when the swinging limb strikes the floor. Loading response begins with initial floor contact and continues until the other foot is lifted for swing. The demand for immediate transfer of body weight onto the limb as soon as it contacts the ground requires initial limb stability and shock absorption while simultaneously preserving
the momentum of progression. Therefore the objective of these phases together is to provide weight bearing stability and preserve progression. Swing-to-stance is kinematically characterized by flexion of the hip joint, extension of the knee joint and plantarflexion of the ankle joint (Figure 2-2). Kinetically this phase is characterized by a horizontal braking force to aid weight acceptance.

The transition from stance-to-swing occurs between 30 to 60% of the gait cycle and the transition from swing-to-stance occurs between 0 to 10% of the gait cycle. The temporal sequence of these transitions is the result of interactions between a tripartite neural system consisting of supraspinal, spinal and sensory input.

**Neural Assembly for Stepping**

**Central Pattern Generator for Stepping**

Rhythmic and reciprocal stepping can be triggered by descending supraspinal command, which delegate the motor coordination to specialized spinal circuitry for pattern generation called the CPG. Primarily, the concept of specialized spinal circuitry existed as far back as 1911, when Brown et al, showed that cats with a transected spinal cord and with cut dorsal roots still produced rhythmic alternating contractions in ankle flexors and extensors. This provided the basis of the concept of a spinal locomotor center that Brown referred to as the ‘half-center’ model. One half of this center induced activity in flexors, the other in extensors.

Much later, Grillner and Wallen coined and demonstrated the existence of CPGs as assemblies of premotor interneurons that were synaptically connected with each other and with the motorneuron pools and capable of creating an elaborate flexor and extensor synergy between different muscles of a limb required for locomotion. Although the anatomical details of CPGs are known for a few cases only, the motor commands originate from the spinal cords of a variety of vertebrates. For example in cats, the nature of pattern generation is still uncertain because the
exact connections of only a few interneurons such as Ia interneurons, Renshaw cells, ventral spinocerebellar cells and spinobulbar cells connecting to lateral reticular nucleus are known with certainty. These neurons could be driven by the CPG or could well be a part of the CPG itself. However, what is clear is the fact that no rhythmic input is required to activate these circuits. These circuits can function in vitro when isolated from the brain, as evidenced by locomotion in decerebrate cats and when isolated from the motor and sensory apparatus of the limbs. The rhythms can often be initiated by simple tonic (i.e. non-oscillating) electric or pharmacological stimulation.

In humans, rhythmic, alternating electromyographic activity of the lower limbs in the absence of supraspinal and movement related afferent input has been interpreted as evidence for central pattern generation. Such evidence was also provided in a study of six subjects with complete SCI, where researchers were able to induce rhythmic, alternating, locomotor-like EMG pattern on continuous epidural spinal cord stimulation.

Therefore the CPG is considered the elementary building block on which rhythmic movement is based. As soon as rhythmic movement is initiated, feedback from the moving limbs, termed as "motion-related feedback" in this review, arrives at the spinal cord to inform the nervous system of the local conditions. This feedback assists in shaping the pattern of walking, reinforcing ongoing activity and controlling the phase transitions. An ensemble of motion-related input arising from skin, joint receptors, muscle spindle, golgi tendon organs, mechanoreceptors, nociceptors is believed to influence the pattern of stepping. Specifically, hip joint position sensed by muscle spindle and load sensed by the golgi tendon organs are two of the several motion-related inputs contributing to the control of stepping.
Several animal and human studies examining the neurobiological control of stepping have validated these findings.

**Sensory Drive Required for Stepping**

Sherrington was the first to propose that proprioceptors responding to hip extension are important for initiating swing. Grillner and Rossignol found that preventing the hip from attaining an extended position in chronic spinal cats inhibited the generation of the flexor burst and hence the onset of the swing phase. The most direct evidence for this conclusion however came from vibrating the hip flexor muscle (iliopsoas) during stance which led to an earlier onset of swing in walking decerebrate cats. The receptors signaling hip extension were probably the primary and secondary endings of muscle spindles in hip flexor muscles (Group Ia afferents).

Similarly, in humans, involuntary and alternating stepping-like movements were observed in an individual after incomplete SCI upon extending the hip in the supine position. Furthermore, hip walking movements (i.e. facilitating hip joint excursion with the knees fixed in an extended position) induced by a driven gait orthosis (DGO) in individuals with complete SCI produced pattern of leg muscle EMG activity that corresponded to that normal stepping in healthy, non-injured individuals. Researchers examining infant stepping also support the role of hip extension position for the initiation of swing. From the recorded hip motion and electromyographic data these scientists concluded that the preferred hip position was always one directly opposite the direction of walking during infant stepping. It was thus suggested that the hip position is important in initiating the stance-to-swing transition.

Another important sensory input regulating the stance-to-swing transition is the extensor load relayed by the Golgi tendon organs (Group Ib afferents) in the ankle extensor muscles. During locomotor activity, electrical stimulation of the group Ib afferents from the ankle extensor inhibits the generation of flexor bursts and hence prolongs the duration of extensor
activity. Duysens and Pearson (1980) observed that gradually increasing load applied to the achilles tendon resulted in an increase in both the amplitude and duration of the rhythmic EMG bursts of ankle extensors.\textsuperscript{89} Similarly, cutaneous afferents innervating the skin of the foot (group II afferents) are also load monitors. Electrical pulses applied to the foot pad innervated by the sural nerve were able to prolong the extensor burst in the stance phase in pre-mamillary cats preparations thereby providing evidence that load-related cutaneous input from the foot can inhibit the CPG for the generation of flexion during swing.\textsuperscript{36,133}

In humans, researchers found that unloading the ankle extensors by a portable device in the stance phase of walking reduced the soleus EMG activity and the reduction was maintained even when transmission in la afferents was blocked by local anesthesia. This finding thus pointed to group Ib and/or group II afferents contributing to the extensor EMG activity in the stance phase.\textsuperscript{134} Harkema et al. (1997) observed that the amplitude of extensor muscle activation in the legs was directly related to the level of body weight loading on the legs during stepping of healthy, non-injured individuals and SCI subjects during manually-assisted stepping on a treadmill.\textsuperscript{135} Dietz et al. (2002) also found that physiological locomotor-like leg movements alone (100\% body unloading) generated by the application of the DGO on a treadmill are not sufficient to generate leg muscle activation in either healthy, non-injured individual subjects or in subjects with complete para-/tetraplegia.\textsuperscript{127} In this study, leg movements in combination with loading of the legs led to appropriate leg muscle activation.

In summary, during stance phase, load of the lower limb is detected by group I extensor muscle afferents and group II cutaneous afferents which activate the extensor half center (EHC) of the CPG. Extensor activity is reinforced during the loading period of the stance phase. At the
end of the stance phase, group Ia afferents of flexor muscles excite the flexor half center (FHC) which inhibits the EHC and thereby initiates the onset of the swing phase (Figure 2-3).

The importance of motion-related sensory input for locomotor control is evident when descending supraspinal input is compromised or altered such as after SCI. In contrast to an intact nervous system processing multiple sources of afferent input, after complete or incomplete SCI, the spinal circuitry does not become silent (Figure 2-4 A & B). The circuitry instead adapts to its altered combination of inputs and predominantly utilizes motion-related afferent input to facilitate locomotor response. The weighted response of the spinal circuitry to ascending afferent input illustrates the high level of spinal automaticity for locomotor control. Therapeutic strategies that optimize motion-related afferent input to the spinal cord can therefore be utilized to regain locomotor control after SCI. For example, stepping can be initiated by shifting the body weight to one leg and moving the head and trunk so that the hip position of the contralateral leg is extended.

Apart from influencing the stepping pattern, sensory input also modulates spinal reflex behavior during stepping. Spinal reflexes are those in which the sensory stimuli arise from receptors in muscles, joints and skin, and in which the neural circuitry responsible for the motor response is entirely contained within the spinal cord. Spinal stretch reflexes are the simplest “stimulus-response” behaviors exhibited by the mammalian nervous system. The stretch reflex as defined by Wolpaw is “the initial, purely spinal, largely monosynaptic response to sudden muscle stretch that is accessible anatomically and physiologically.” H-reflex is an electrical analogue of the spinal stretch reflex and is a commonly studied spinal reflex in human beings. It was originally described by Paul Hoffman in 1910 and is electrically elicited leading the stimulus to bypass the effect of the gamma motorneuron. Therefore, H-reflex is a valuable tool
in assessing modulation of monosynaptic reflex activity in the spinal cord and has been used since to assess the response of the nervous system to various neurologic conditions, application of therapeutic modalities, exercise training and motor task performance.

**H-reflex Elicitation and Modulation During Stepping**

**Elicitation**

The technique used to evoke the H-reflex involves electrical stimulation of a mixed peripheral nerve. When a percutaneous stimulation of increasing intensity is applied to a mixed nerve, action potentials travel along afferent portion of the reflex arc along the Ia sensory afferents, until they synapse on the alpha motoneuron. The efferent portions of the reflex pathway results from action potentials generated by the alpha motor neuron until they reach the myoneural junction and produce a twitch response which is recorded by surface electrodes on the muscle of interest (Figure 2-5A).

In addition to the H-reflex, electrical stimulation of the peripheral nerve also causes direct activation of the efferent fibres that conduct orthodromically to produce a response in the EMG known as a muscle response or M-wave. When the stimulus intensity is really low only the Ia afferent fibres of the mixed nerve undergo depolarization leading to the appearance of the H-reflex tracing on the EMG. As the stimulus intensity is increased, more Ia afferent fibres are recruited, resulting in activation of more alpha motor neurons and increasing the amplitude of the H-reflex. Continuing to increase the intensity beyond the point of elicitation of the H-reflex also results in direct stimulation of the motor axons and thereby production of the M-wave (Figure 2-5A). The M-wave on a recording is usually seen before the H-reflex due to the relatively short path that the action potentials need to travel. Compared to this the H-reflex is characterized as a latency response since it does not occur right after the application of the stimulus (Figure 2-5B). The latency of the reflex response is a result of the length of the H-reflex pathway, which
involves the afferent and efferent length of the path and also the overall length of the limb. For example, in the soleus muscle, the H-reflex tracing usually appears approximately 30 milliseconds after the delivery of the stimulus whereas the M-wave is usually seen after 6 to 9 milliseconds of stimulus application.\textsuperscript{138}

The amplitudes of the H-reflex and M-wave both increase fairly linearly with increase in stimulation intensity until the maximum H-reflex (H-max) representing the fullest extent of reflex activation is reached.\textsuperscript{138} Increasing stimulus intensity further beyond H-max, maximum M response (M-max) representing the maximum muscle activation is reached.\textsuperscript{138,149} Therefore, by increasing the stimulation intensity a recruitment curve depicting, stimulation intensity sufficient to evoke a sequence of H-reflex, H-max, disappearing H-reflex tracing, M-max and M-max plateau can be obtained (Figure 2-5C).\textsuperscript{138,149}

\textbf{Task-specific/ Phase-dependent modulation of the H-reflex}

Soleus H-reflexes are the most widely assessed reflexes in locomotor studies.\textsuperscript{138} In healthy, non-injured individuals, soleus H-reflexes are strongly modulated during the gait cycle with the highest amplitude registered during the stance phase and the lowest amplitude recorded during the swing phase.\textsuperscript{146} H-reflexes are minimal at the time of heel contact, rise to a maximum shortly after midstance, decrease rapidly at the time of toe-off and are minimal during swing in both young and older age groups.\textsuperscript{150} Therefore, soleus H-reflexes in healthy, non-injured individuals demonstrate phase-dependent modulation during stepping. Along with the phase-dependent modulation of the reflex during the step cycle, H-reflexes are also known to display considerable differences in modulation between different motor tasks.\textsuperscript{147,151} Both phase-dependent and task-dependent modulation of the reflex is critical to the optimal performance of motor behaviors. Modulation of the H-reflex during stepping has been attributed to both, sensory input\textsuperscript{152,153} and to the higher central mechanisms controlling motion.\textsuperscript{154} Although the bulk of literature supports the
primary role of central reflex modulation during walking, the secondary role of sensory inputs is critical as well.

The role of sensory inputs is especially critical when the central control of reflex modulation is compromised for e.g. post SCI. As a result of injury to the spinal cord, phase-dependent modulation is impaired and reflex amplitudes are higher than normal throughout the gait cycle. In such a scenario, the reflex muscle response is likely to be dependent on peripheral sensory inputs deficient of central modulation. Several studies have reported that the soleus H-reflexes can be modulated by sensory inputs of peripheral origin such as hip joint position, leg load, and cutaneous receptors in sole of the foot. Sensory input related to hip joint position or from hip joint proprioceptors for example is shown to markedly influence soleus H-reflexes during passive movement of the hip from flexion to extension phase in both healthy, non-injured individuals and persons with spinal cord injury. 

Similarly, mechanical loading of the foot sole, ranging from 15 to 70 N is known to significantly inhibit soleus H-reflex amplitude in both seated healthy, non-injured individuals and persons with complete SCI. Although performed in non-locomotor tasks, the above studies are suggestive of a possible alteration in H-reflex amplitude via peripheral sensory inputs even after SCI. Thus post-SCI, provision of critical motion related sensory inputs may play a role in phase-dependent H-reflex modulation and could be useful in optimizing task performance.

Recovery of Walking after SCI

Plasticity at the Level of the Spinal Cord

The lifelong ability of the nervous system to reorganize neural pathways structurally or functionally in response to experience or activity termed as “neuroplasticity” was once solely considered to be a supraspinal phenomenon. Acquisition and maintenance of normal motor performance however involved skill-dependent plasticity at multiple sites including the spinal
Early evidence of this phenomenon was seen as early as in 1951 when Shurrager and Dykman reported that treadmill walking in spinalized cats improved with training. Locomotion was much better in cats exposed to treadmill training than in cats that received standard care after injury. The improvements in training seen after training persisted even after training was stopped. Similarly, De Leon et al performed a series of experiments in spinalized cats either trained to stand or step on the treadmill. Their results revealed that cats trained to stand improved standing function and those trained to step improved in stepping. Locomotor ability was exactly reversed in these groups when they were retrained to perform the other task.

These researchers also tested the effect of glycinergic inhibitor, strychnine in spinalized cats that were trained to either walk or stand on the treadmill. They observed that locomotion improved in cats trained to stand when strychnine reduced glycinergic inhibition and had no effect on the cats trained to step. These findings suggested that glycinergic inhibition in the spinal cord interfered with stepping ability in spinal animals and locomotor training improved stepping ability by reducing the levels of inhibition on the spinal networks. Skill-dependent training can therefore markedly change or modify physiological and biochemical state of multiple neurotransmitter-modulator systems in the spinal cord and enhance locomotor recovery. Studies of motor unit properties after spinal cord transection, with and without training, have also provided supplemental evidence by indicating that training induced improvements in walking and standing are not attributable to peripheral changes in muscle strength.

Several studies in humans have also provided compelling evidence for activity dependent plasticity of the spinal cord caudal to injury. Similar to cats, treadmill training using a body weight support showed significant improvements in walking behavior in people after severe
spinal cord injuries. Dietz et al. (2003) also acknowledges that alteration in glycnergic and GABAergic systems seen in animal models as a result of the training could be true for humans as well. The spinal cord as a result of the training learns to respond to specific sensory inputs associated with locomotion which would potentially reorganize the circuitry involved in locomotion. Histologic data supporting the possibility of sprouting or synaptogenesis in the human spinal cord although not direct are also evident.

Plasticity of the Spinal Stretch Reflex (SSR)

Contradictory to their conceptualization as “hard-wired”, the spinal stretch-reflexes are modulated during movement and adapt to training. Reflex modulation refers to the change in strength (or amplitude) of the reflex over the course of a behavior and is essential for the optimal performance of the motor behavior. Researchers have shown that in monkeys, rats and humans these reflexes can be operantly conditioned i.e. the amplitude of the response can be either up-trained or down-trained. In these protocols the amplitude is measured as electromyographic activity and reward occurs when the amplitude is either above or below a criterion level. While change in the tonic descending activity motivated by the probability to seek reward initiates reflex change, for this reflex to respond consistently in this fashion over several sessions certain alterations occur somewhere in the spinal arc of the reflex.

Similarly, short term and long term changes have been noted in the reflex amplitude in animals and humans as a result of task-specific training. A single training session of short bouts of balancing on an unstable platform in normal subjects demonstrated a progressive decrease in the soleus H-reflex amplitude of about 26%. Hess et al also observed a modulation of the reflex size during adaptation to a new motor task. H-reflex modulation over time was evaluated for normal subjects over five runs of treadmill walking (three with normal treadmill walking and two with randomly stepping over the obstacle 100 times). The largest adaptations
with a significant increase of reflex amplitude occurred during the first obstacle run. This increase lasted only briefly and the reflex amplitudes decreased to their previous values. During the later obstacle run, no H-reflex modulation occurred. Similarly, normal subjects training to walk backward showed progressive adaptation of their soleus H-reflex in mid-swing phase as early as twenty minutes after training.¹⁷⁵ Long-term maintenance of this modulation was also noted as late as five months after cessation of training indicating acquisition and maintenance of novel motor skills.

Operant or training induced conditioning of the reflex can have potential implication in rehabilitation. Studies based on application of this phenomenon in subjects with neurological injury have shown significant changes in the reflex behaviors of these individuals. For example, studies have shown that operant conditioning of the spinal reflex reduced hyperactive biceps stretch reflex in people with spinal cord injuries with the reduction persisting for four months following cessation of training.¹⁷⁸ Similar results in locomotor studies have only been documented in spinalized animal models thus far but might have potential ramifications in improving human locomotion as well.⁴⁵,¹⁷⁹,¹⁸⁰ In four subjects with incomplete spinal cord injury, for example, a single bout of step training over the treadmill increased overground walking speed by 25% and reduced soleus H-reflex amplitude during overground walking providing evidence for activity-dependent plasticity of the reflex.¹⁸¹

To this end, since reflex behaviors function as parts of complex behaviors, conditioning them in accordance to the requirements of the task i.e. task specific training may help improve functional outcomes in the targeted population.¹⁸² Also from a rehabilitation standpoint, provision of appropriate task-specific stimuli and/or adopting different feedback strategies that could facilitate or depress these reflex responses in desired ways could help in retraining or
reeducation of function. For example, providing optimal hip extension during stepping would up-train the stretch response of the hip flexors to facilitate hip flexion for swing.

In summary, short-term and long-term adaptive changes occur within the reflex components as a result of conditioning. These changes are also known to persist despite removal of the descending input thereby exhibiting learned behavior at the level of the spinal cord. This adaptability and amenability of the reflex to persistent change or conditioning constitutes spinal cord plasticity.

**Locomotor Training and Skill-dependent Plasticity of the Nervous System**

Animal and human research on the neurobiological control of stepping and the skill-dependent plasticity of the nervous system have challenged the compensation based approach of rehabilitation after SCI. Research has revealed that task-specific, repetitive training following SCI in animals and humans promotes skill-dependent plasticity in the spinal cord and plays a critical role in the recovery of locomotor abilities including stepping. The emerging training strategy is to provide the CNS with peripheral sensory input related to locomotion in order to stimulate a stepping response. Processing of task-specific kinematic and kinetic information facilitates performance of the task and learning. This strategy is based upon evidence that the lumbosacral spinal cord is capable of recognizing and processing functional sensory cues to produce a functional motor response. Simply put, generation of the stepping pattern would involve the provision of motion-related afferent input associated with stepping. Two of these inputs related to stepping are loading and hip position, which have been discussed earlier. A therapeutic intervention termed as "locomotor training" has been developed based on such research.

In the locomotor training environment this specific sensory input is made available by having a person with SCI walk over a treadmill in a harness connected to a BWS system.
assembly provides an environment where normal walking speeds, bilateral limb loading and proper limb and trunk kinematics can be safely and effectively trained to enhance the neural output generating walking.\textsuperscript{43} Evidence supporting the benefits of locomotor training in facilitating the recovery of walking in humans has been recorded in the literature as early as the 1990s.\textsuperscript{25,39,41,42,186-188} Persons trained in this training environment demonstrated significant improvements not only in pattern of stepping, walking velocity, endurance but also in neurophysiological predictors of improved coordination like the electromyographic activity in the muscles of the lower extremity.\textsuperscript{189-191} Soleus H-reflexes have also been shown to respond to training post-SCI. A single bout of training has shown to increase walking speed with a corresponding decrease in reflex amplitude, which is usually high after injury.\textsuperscript{181}

**Is Orthotic use Appropriate during Locomotor Training?**

Despite the compelling evidence supporting the concept of locomotor training controversy and variability still exists in training parameters in the training protocol.\textsuperscript{24-26,42,46,192} Not all the parameters for optimizing recovery are known thus far and those that are, for example, BWS, manual assistance and speed are still being reviewed for mode of delivery and dosage.\textsuperscript{193} Therefore although the principles of training have been established, there is no universal agreement on ideal training parameters for locomotor training.\textsuperscript{24,46,82} For example, controversy still exists about the use of parallel bars during training. In a study done by Conrad et al improvements in step symmetry were observed using conventionally used parallel bars during training.\textsuperscript{68,194} On the contrary, Visintin et al reported improvements in symmetry using vertical body weight support instead of parallel bars providing upright support.\textsuperscript{195}

Similarly, although widely used in conventional gait training, guiding evidence for the use of orthotic devices such as the AFO during LT is lacking.\textsuperscript{24} Proponents of LT hesitate to use an AFO during training based on the assumption that it would interfere with optimizing the
kinematics and therefore afferent input required to drive the neural circuitry. Logical concerns expressed are that the AFO might restrict the range of excursion of the ankle and linked knee and hip joints altering the afferent information related to hip position and load. Additionally, although an AFO might facilitate an alternate stepping pattern, it would serve as a fixed variable unlike BWS and manual assistance that can be adjusted to facilitate independent stepping over the course of the training. Therefore the use of the device during LT qualifies further investigation.

**Altered Stepping Mechanics Resulting from the Use of Orthotic/Assistive Devices**

Although the goal is to improve ambulation potential, prescription of assistive and/or orthotic devices often does not take into account the influence of the device on the user's resultant stepping pattern. For example, rigid support provided at the ankle by the PAFO might limit the excursion of the contingent joints affecting the control of the transition from stance-to-swing. Similarly, loading through the fixed upper extremities over the parallel bars alters the prerequisite loading pattern through the lower extremities required for stepping. However, the hierarchical framework assuming irreversible walking deficits post-injury have promoted compensatory function with such devices rather than examining their implications on the neural sub-tasks of walking.

Under the supposition that neurological function ceases to exist below the level of injury, clinical assessments for the prescription of ambulatory devices utilize a frame work for substitution of impaired function rather than the restitution of function. Therefore, the ambulatory ability of a person in moving from one place to another rather than the walking pattern utilized in achieving this mobility is emphasized. Individuals requiring assistive devices have a decreased ability to provide the supporting, stabilizing, propulsive or restraining forces at the lower extremities necessary for forward progression. With assistive devices the upper
extremities are the structures bearing the load and are being employed in providing the complementary forces for walking. Melis et al. (1999) reported a decrease in hip excursion and a decrease in the ability to unload the limb while walking with a walker. Visintin and Barbeau (1994) investigated the consequences of weight bearing on the upper extremities compared to weight bearing through the legs, both with 40% BWS provided. Her results indicated a decrease in electromyographic activity in the lower limbs and more asymmetry in the limb kinematics.

Similarly while wearing orthotic devices like the PAFO, Ounpuu et al. (1996) reported an inability to generate sufficient power in the transition from stance-to-swing phase of stepping. However, conclusions drawn from these studies contend that ambulatory devices are still capable of fulfilling various assistive functions during walking, although they affect posture and walking pattern. The inability to generate normal walking mechanics was concluded to be the result of the irreversible nature of the injury rather than the inability of the device to provide or assist normal stepping. Therefore the assessment of an assistive and/or orthotic device based on its ability to restore normal walking mechanics is a novel perspective.

**Altered H-reflexes Resulting from the Use of Orthotic Devices**

The H-reflex has been commonly employed as a neural probe in describing and interpreting neural interplay for the control of movement. Elicitation of the reflex and measurement of its amplitude has provided insights into the changes in transmission in spinal pathways during the performance of a motor task. Since afferent information is processed at the multiple levels of the nervous system including the spinal cord, examination of the soleus H-reflex modulation at the ankle joint would be a favorable tool in linking device-dependent sensory information to a motor response. Schneider et al, for example, demonstrated that in normal subjects bracing the ankle joint angle to 90°, such as in preventing foot drop, the burst of
TA activity was eliminated and so too the inhibition of the H-reflex.\textsuperscript{154} A similar inhibition in H-reflex was observed by Brooke et al on bracing the ankle joint suggestive of premotorneuronal mechanism for reflex inhibition.\textsuperscript{204} Nishikawa et al. (1999) examined the effect of applying an ankle brace in sitting to an uninjured ankle.\textsuperscript{205} They reported an increase in H-reflex amplitude in the braced compared to the unbraced condition concluding that ankle bracing increased stimulation of the cutaneous mechanoreceptors around the joint. However, in these experiments the subjects were tested during passive movement of the limb or in a static position instead of active walking.

Garrett et al. studied the effect an knee orthosis on soleus H-reflex modulation during walking.\textsuperscript{206} They reported that even though the walking pattern changed in normal individuals, phase-dependent H-reflex modulation was not influenced by the use of a knee orthosis. In this study however, the ankle was not restricted and hence the specific effect of an AFO is unknown. Therefore differential modulation of the H-reflex while walking with and without wearing an AFO will help determine if walking with an AFO is an inherently different motor task compared to normal walking.

**Rationale of the Studies**

Demonstrating the presence of numerous parallel systems within the CNS that reorganize after injury using intense, task-dependent interventions such as LT has catapulted the goal for rehabilitation of walking after SCI from compensation to recovery.\textsuperscript{21,207,25,208} Recovery of the sub-component of stepping, in this intervention depends on careful selection of training variables that deliver critical motion-related sensory input to facilitate stepping. Conversely, aptness of a training variable such as an AFO to facilitate stepping can be determined by examining the mechanical and neurophysiological deviations observed during stepping.
Therefore, I investigated the effect of an AFO on mechanics and soleus H-reflex modulation in healthy, non-injured individuals and persons with ISCI during stepping. Measuring these two factors in healthy, non-injured individuals provided a frame of reference for subsequently comparing normal and impaired stepping. Whereas, examining changes in the two factors in persons with ISCI and comparing it to normal helped identify the degree of correction or deterioration in stepping provided by the AFO. Therefore, the cross-sectional studies have gleaned specific information about the use of AFOs in physiological paradigms that optimize recovery of walking after SCI.

Methodological Considerations for the Measurement of Mechanics and Soleus H-reflex During Stepping

Measurement of Stepping Mechanics

Study considerations

In the study protocol, measurement of joint excursion at the hip, knee and ankle with and without wearing an AFO were used to assess critical kinematics required for the transition from stance-to-swing. The AFO was fitted on the dominant side for healthy, non-injured control subject and the more involved side for persons with ISCI. Kinematic and kinetic data are susceptible to change with walking speed.\textsuperscript{115} Therefore, the speed of walking was matched in both the walking with and without the AFO conditions. Also kinematic data is affected by limb length and height of the person.\textsuperscript{115} Therefore, for comparisons between people with ISCI and their non-injured counterparts, height, age and weight matched individuals were selected for the study.

Interpretation of kinematic and kinetic data

Walking is a motor task in which each of the interconnected segments of the lower limbs undergoes a characteristic excursion to move the body forward.\textsuperscript{33} If the terminal excursion of any
one of the joints is restricted it has the potential to alter the excursion of the other joints linked to it thereby disrupting stepping pattern. The stance-to-swing transition phase is usually characterized by ankle plantarflexion, knee flexion and hip extension to aid push-off.\textsuperscript{33} If ankle range is restricted using an AFO, plantarflexion will be limited and the knee and hip excursion will also be altered. Similarly, the swing-to-stance transition phase is characterized by ankle plantarflexion, knee extension and hip flexion.\textsuperscript{33} If ankle plantarflexion is limited the knee and hip range will also be altered. Assessment of kinematic data while walking with an AFO will be helpful in elucidating this phenomenon.

Furthermore assessment of kinetic data will indicate the discrepancies in the functional task requirements of stepping. For example, during stance-to-swing phase, plantarflexor power is generated to propel the limb and body forward.\textsuperscript{33,203} If plantarflexor power is limited while walking with an AFO,\textsuperscript{203} the knee and hip joint powers might demonstrate a compensatory increase or decrease to propel the limb forward. Also, with limited propulsion, peak braking and propulsive forces generated in the fore-aft direction will decrease compared to normal push-off. Similarly, the rate of loading will be used to quantify the ability to shift weight from one lower limb to the other and measured by calculating the slope of the vertical ground reaction force.\textsuperscript{209,210} A prolonged loading rate onto the limb donning the AFO will be indicative of an inability to plantarflex the ankle and extend the knee to support the transferred weight.\textsuperscript{210} Collective assessment and interpretation of the above parameters will therefore provide valuable information with regards to the mechanics of stepping with an AFO.

**Reliability of Vicon and Force Plate Measures**

Reliability refers to the ability of an instrument to provide consistent, stable and repeatable measurements. In this project, Vicon motion analysis system (Vicon, Oxford Metrics Ltd, Oxford, England) and kistler forceplates (Kistler Instruments, Inc., Amherst, NY) were used
to identify the mechanical changes during stepping with and without a PAFO. The Vicon motion analysis system is an automated, high-speed three-dimensional (3-D) motion system where cameras track the motion of retro-reflective markers that are placed on subject's body landmarks. Kadaba et al (1989) investigated the repeatability of kinematic, kinetic and electromyographic data using the Vicon motion analysis system on forty normal subjects three times a day on three separate days. Excellent intra-rater reliability was seen for kinematic data in the sagittal plane both within and between test days (Table 2-1). Similarly, frontal and transverse planes joint angle motion yielded good repeatability within test days leading them to conclude that gait variables measured by the Vicon system are quiet repeatable for subjects walking at their normal speed. Similarly, Richards et al (1999) compared accuracy of several automated motion analysis systems by placing two markers 50cms apart on an aluminum bar rotating in the horizontal plane in the camera capture volume. The results of his study indicated that the Vicon was the most accurate with a maximum error of 0.183 cm that is the lowest among all other systems. Several other studies have also reported excellent intra-rater, inter-rater and test-retest reliability of the Vicon motion analysis system making it the current “gold standard” for motion analysis.

Ground reaction forces were measured with two kistler force plates that use triaxial piezoelectric force transducers mounted at the corners of each plate to measure the three components of the ground reaction force vectors. These force plates have been shown to be the reliable standard for measuring dynamic transition from bipedal to single limb stance in healthy, non-injured adults. Intraclass correlation coefficient (ICC) for the magnitude of the propulsive and braking force has been reported to be greater or equal to 0.73 for fast movements and greater or equal to 0.88 at the natural speed.
Force plates have also been used for comparing ground reaction force patterns in non-injured individuals and individuals with ISCI.\textsuperscript{217} Repeatability of initial vertical force peak and time to peak variables measured by force platforms while donning foot orthosis has also been reported.\textsuperscript{218} Excellent ICC results were demonstrated for the vertical force variables, with power greater than 0.80. Measurement of peak vertical ground reaction force during a vertical jump at two time points 48 hours apart was demonstrated to be very reliable (ICC $[2,1] = .94$).\textsuperscript{219}

**Measurement of Soleus H-reflex**

**Study considerations**

Elicitation of soleus H-reflex during walking with and without a PAFO requires control of several extraneous factors that could potentially confound the reflex response. These factors have been identified below and methods to control them have been discussed. For the purpose of consistency, soleus H-reflex will be evoked, on the dominant side of healthy, non-injured control subject and on the more involved side for persons with ISCI. The reflex will be evoked by localizing the tibial nerve in the popliteal fossa.\textsuperscript{138,220} Subject positioning is critical during H-reflex testing since several factors affect the soleus H-reflex. H-reflex is sensitive to various inputs including posture,\textsuperscript{221} joint position,\textsuperscript{149} reciprocal and recurrent inhibition,\textsuperscript{222} behavioral state,\textsuperscript{223,224} caffeine intake\textsuperscript{225} and muscle activity.\textsuperscript{226} However, if the above factors are sufficiently controlled, then H-reflex can provide information of the state of the reflex arc.

Stimulation intensity is another factor that affects reflex response. Intensity was maintained between 8-12% of M-max so as to evoke a direct muscle response. This procedure helped to safeguard against movement of the stimulating electrode that might alter the relative activation of the Ia afferent axons and alter the H-reflex amplitude without changes in synaptic efficacy. Soleus H-reflex amplitude is affected by background EMG activity of the muscle.\textsuperscript{149,226} Background EMG activity was measured 100 ms prior to stimulation in each condition to ensure
similar level of motor neuronal excitability. Stimulus frequency was maintained between 3-5 seconds to avoid post-activation depression of the response.\textsuperscript{222} Since profound reductions in M-max amplitude have been reported to occur across the time course of an experiment, M-max was elicited in each condition and in each tested phase of the gait cycle throughout the experiment for subsequent normalization of the data.\textsuperscript{227}

**Interpretation of H-reflex amplitude**

The H-reflex demonstrates a phase-dependent regulation of its amplitude during walking which is an important component of motor control, allowing afferent feedback to have differing effects in different phases of the step cycle.\textsuperscript{228} This regulation is required to accommodate the functional requirements of the task. For example, the soleus H-reflex is minimal at heel contact, increases to maximum during stance and decreases rapidly just prior to toe-off and is minimal during swing. The observed changes in size facilitate weight support and ankle extension during mid-to-late stance while allowing ankle dorsiflexion during swing and while the body moves over the foot during early to midstance.\textsuperscript{146,228,229} Therefore if the task of walking with an AFO was similar to walking without one then the neural control would be preserved between the two conditions and the soleus H-reflex modulation in the step cycle would be similar. An increase in H-reflex amplitude while walking with an AFO (assuming all other conditions including stimulus strength are maintained constant) compared to walking without one in selective phases of the step cycle would be indicative of an altered task.

Similarly, with regards to adaptation of the reflex to a new motor task studies have shown this to occur in a biphasic fashion.\textsuperscript{137} For example, for the novel task of stepping over an obstacle, Hess et al demonstrated a progressive adaptation in soleus H-reflex amplitude during repetitive stepping. He showed that in normal subjects the soleus H/M ratio increased strongly at onset of the motor learning task and reduced over the course of exercise reflecting the nervous
systems capacity to adapt the locomotor pattern to the actual requirements. The initial increase in soleus H-reflex amplitude is attributable to descending influence of the corticospinal tracts on the spinal reflex arc. The eventual reduction in reflex amplitude is speculated to be the effect of the acquired task being automatically performed and controlled at a spinal or brainstem level. Therefore prolonged successive stepping with an AFO in normal subjects if different from walking without one would also exhibit adaptive changes in reflex amplitude as seen with the acquisition of a new motor task.

Reliability of Soleus H-reflex

The reliability of the soleus H-reflex testing in supine and standing position has been confirmed for inter-session and intra-session procedures. ICC (2, 1) for H-max, M-max and H-max/M-max has been reported to be 0.99 +/- 0.007, 0.95 +/-0.08,0.97 +/-0.009 respectively. Similarly, the intra-session and inter-session reliability of soleus H-reflex over five consecutive days in a standing position have also been established. The standing intra-session and inter-session reliability was established to be 0.85 and 0.80 respectively.

Clinical and Scientific Relevance of the Study

Clinical Relevance of the Study

Walking with orthotic and assistive devices has been the quintessential approach for improving walking potential in persons with incomplete spinal cord injuries. Assessment of such devices in physiological-based training paradigms like locomotor training will provide information about the degree of conformity of such clinical strategies with the principles of neurobiological control of walking. Accordingly, in this study, mechanics of stepping and H-reflex modulation with the AFO will be evaluated which characterize the motion-related afferent input being processed at the level of the spinal cord. Altered stepping mechanics and reflex modulation with compensatory devices would reflect the failure to provide task-specific sensory
input driving the intrinsic spinal circuitry necessary for walking recovery. Knowledge of results from this study would influence clinical decision-making for the use of such devices and strategies in physiological-based training interventions.

**Scientific Relevance of the Study**

Apart from the clinical implications, this study stimulates a strong rationale for assessing archetypical strategies that have historically guided clinical practice so far. Generation of appropriate task mechanics associated with stepping is hypothesized as essential for recovery of stepping pattern in physiological-based training interventions. The transition phase mechanics are modulated by the motion-related information processed in every gait cycle. Assessment of the task mechanics of stepping in these transition phases while walking with orthotic devices is critical to examine how motion related information generated by such strategies influence the ability to step successfully.

Assessment of conventional strategies from a neural framework is relevant to the neurological population to which these strategies commonly apply. The tools used for assessment of walking are relevant to address the task and function of specific events characterizing the walking behavior. Therefore the integration of two different perspectives, biomechanical and neurophysiological, will provide an effective framework for understanding the control of movement and addressing the questions posed.
Figure 2-1. Reduced animal preparation showing the impact of transection at different levels of the nervous system. Adapted from Patla, A. E. in Evaluation and management of gait disorders (ed. Spivack, B. S.) (New York, 1995).
Figure 2-2. Stance-to-swing and swing-to-stance phases of the gait cycle.

Figure 2-3. Model of the flexor and extensor half centers (FHC & EHC) and afferent input regulating stance and swing phase during stepping. Adapted from Van de Crommert, H. W., Mulder, T. & Duysens, J. Neural control of locomotion: sensory control of the central pattern generator and its relation to treadmill training. Gait Posture 7, 251-63 (1998).
**Figure 2-5.** Electrical stimulus (shown here by the grey ellipse) applied to the mixed nerve conducts the stimuli orthodromically in the motor and sensory axons to evoke the M-wave and the H-reflex respectively (A). Stimulus triggered recording of the H-reflex also known as latency response occurring after the M-wave (B). Adapted from Zehr, E. P. Considerations for use of the Hoffmann reflex in exercise studies. Eur J Appl Physiol. 86, 455-68 (2002). Elicitation of the recruitment curve showing maximum H-reflex and M-max amplitude in response to stimulus intensity (C). Adapted from Palmieri, R. M., Ingersoll, C. D. & Hoffman, M. A. The Hoffmann reflex: methodologic considerations and applications for use in sports medicine and athletic training research. J Athl Train. 39, 268-77 (2004).
Table 2-1. Coefficient multiple correlations (CMC) reflecting stride to stride variability between and within days.

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<th>CMC within day</th>
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<th>CMC between days</th>
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<td>Right</td>
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<td>Vertical GRF</td>
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<td>Horizontal GRF</td>
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<td>Hip Flexion/Extension</td>
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<td>Knee Flexion/Extension</td>
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<tr>
<td>Ankle dorsiflexion/plantarflexion</td>
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Spasticity, muscular weakness and co-activation are key motor impairments limiting walking potential in individuals with incomplete spinal cord injury (ISCI). As one component of rehabilitation, clinicians often use orthotic devices to compensate for these impairments and aid walking. Individuals lacking muscular control at the ankle are prescribed single joint ankle foot orthosis (AFO) to stabilize the joint, supplement deficient push-off and aid toe clearance during stepping. An AFO limits ankle excursion and simultaneously influences excursion of the knee and hip, thereby allowing the person to gain more proximal control for walking. AFOs also improve overground walking speed and interlimb kinematics in persons with ISCI. These broader benefits may lead to the assumption that a role exists for AFOs as permissive devices in recovery based interventions such as locomotor training (LT).

The AFO induced alterations in stepping mechanics, however, might not coalesce with the training principles of LT. During LT, stepping is retrained by practice of task-specific repetitive, rhythmic, stepping kinematics over a treadmill using task facilitatory training variables such as body weight support and manual assistance. The training is based on the facilitation of intrinsic mechanisms within the spinal cord that respond to specific afferent input associated with the task of stepping. The spinal cord processes afferent information arising from muscle, joint and cutaneous receptors during stepping to adapt the motor output to the phase of stepping. For example, hip extension and limb unloading are critical afferent inputs required to initiate the transition from stance-to-swing during stepping. Limiting hip extension and/or unloading delay this transition and limit the forward progression of the body.
Proponents of LT therefore hesitate to train with an AFO due to the assumption that it impacts optimal kinematics and motion-related sensory input required for stepping.\textsuperscript{24,26,237} Logical concerns expressed are that the AFO might restrict the range of excursion of the ankle and linked knee and hip joints altering the afferent information related to hip position and load. Additionally, although stance-to-swing transition with the AFO might be alternatively possible, the orthosis is a passive element in an otherwise active training protocol where other facilitatory variables can be adjusted to facilitate independence. Therefore the purpose of the study was to investigate the use of the device during stepping in healthy, non-injured individuals and persons with ISCI.

To assess the production of critical ankle, knee and hip joint kinematics required for stepping the range of excursion of these joints were measured during stepping.\textsuperscript{70,238-241} Furthermore, measurement of the vertical and horizontal forces reflected the rate of loading/unloading the lower limbs and peak braking/propulsive force required for forward progression of the body respectively.\textsuperscript{240-242} As secondary variables of interest, interlimb temporal and spatial measures of symmetry correlated to the functional task requirements of walking in these transition phases such as double limb support time and step length were also assessed.\textsuperscript{5,243} In summary, two experiments were conducted interpreting mechanical information in transition phases (stance-to-swing and swing-to-stance) while walking with and without the AFO. The first experiment (Refer to Chapter 4) examined transition phase mechanics with and without an AFO in healthy, non-injured individuals that provided normative data for subsequently comparing with persons with ISCI. The second experiment (Refer to Chapter 5) compared transition phase mechanics observed while walking with and without the AFO in individuals with ISCI to normal...
mechanics. The comparison assessed changes in pre-existing stepping related deficits while walking with the AFO and measured deviance or likeness of the observed change from normal.
The purpose of the experiment was to examine the effect of an AFO on transition phase mechanics during walking in healthy, non-injured individuals. Specifically, I assessed the immediate effect of wearing a PAFO on ipsilateral lower extremity 1) kinematics and 2) kinetics during the stance-to-swing and swing-to-stance phase of walking.

Hypotheses

1A: Compared to walking without an AFO on a treadmill at speed approximating 1.2 m/s, wearing an AFO in healthy, non-injured individuals will affect the stance-to-swing transition mechanics observed on the AFO side: specifically, decrease peak ankle plantar flexion and hip extension and increase peak knee flexion range and increase peak knee and hip flexor powers and decrease peak ankle plantarflexor power.

1B: Compared to walking without an AFO on a treadmill at speed approximating 1.2 m/s, wearing an AFO in healthy, non-injured individuals will affect the swing-to-stance transition mechanics observed on the AFO side: specifically, decrease ankle plantarflexion and increase hip and knee flexion range of motion and decrease rate of loading and peak braking force.

Methods

Subject Selection

A sample of convenience consisted of fourteen healthy, non-injured individuals living independently in the Gainesville community (Table 4-1). Each subject provided informed consent before participating in the study. The University of Florida Institutional Review Board and the Veteran Affairs Subcommittee approved the study for clinical investigation. Mean age and standard deviation was between 26.9± 3.7 years. Subjects were screened for a medical history of any neurological, musculoskeletal or orthopedic problem that would affect their walking performance over the treadmill. Power analysis for determination of sample size was based on pilot data from three healthy, non-injured participants. The change in hip joint angle and peak braking force were selected for calculating sample size since these are the variables of
primary interest. It was determined that a sample size of fourteen subjects was required to reach an alpha level of 0.05 and a power of 0.80

**Experimental Set-up**

Once the subject had read and signed the informed consent form, motion data was collected and analyzed using a 3-D motion analysis system in conjunction with an ADAL3D instrumented split-belt treadmill custom manufactured and calibrated by TECMACHINE (Cédex, France), mounted flush with the floor and anchored to the foundation. Four Kistler piezoelectric sensors on each half treadmill allow calculation of the two-dimensional location of the center of pressure (COP) and the moment about the vertical axis, in addition to the three-dimensional ground reaction force, under each foot. Belt speeds can be controlled as slow as 0.1 m/s.

Ground reaction forces were recorded at 1000Hz for each limb when in contact with the treadmill belt. The force plates were allowed to warm-up for at least 15 minutes as per manufacturer guidelines and calibrated prior to data collection. The walking pattern of the subjects were captured and analyzed by a Vicon three-dimensional motion analysis system. The system consists of the VICON 612 Datastation with twelve active video channels and a 64 Channel A/D Board for analog signals. There were twelve 1000Hz M2-cameras (Digital CMOS M2 series cameras have a resolution of 1280 x 1024). Included software was: Workstation, Polygon, BodyBuilder, Plug-In Gait, Plug-In Modeller, and Real Time II. The twelve cameras had a frame rate of 60-120 fps and used infrared (IR) light-emitting diode strobes, which were gen-locked. Static calibration of the system used the clinical L-frame, which contains 4 retro reflective markers, being placed in a predetermined position on the motion analysis force platforms. Following this a dynamic calibration was done using a 500mm wand that was moved around the capture area for approximately 20 seconds. Analog video data were also collected
using a standard camcorder recording at 100 fps with its optical axis perpendicular to the plane of interest (i.e. the sagittal plane of motion).

**Subject Preparation**

Subjects were asked to wear tennis shoes and change into appropriate clothing (dark colored cycling shorts and shirt) for testing. For trials using the AFO, each subject was fitted with an off the shelf posterior leaf spring ankle foot orthosis (PAFO). Fitting was assessed by measuring fit inside shoe, length of the calf shell and that of the footplate. Standardized fitting included using a PAFO whose length fits an inch to two below the fibular head when donned and whose footplate length extends till the tip of the toes. Lightweight retro-reflective markers were attached to the following bony landmarks: posterior superior iliac spines (PSIS), anterior superior iliac spines (ASIS), knee-joint axes, lateral malleoli, medial malleoli, clavicular notch, sternum, C7, T10, and acromium processes. The second foot ray, base of the 5th metatarsal and the heel markers were approximated on the subjects’ shoes. Clusters of markers were attached to the pelvis, thigh, shank, and foot segments.  This modified Helen Hayes marker set is commonly used to capture bilateral 3D kinematics using a twelve-camera VICON motion analysis system.

Each subject was fitted with a body weight supporting harness equipped with an additional overhead safety catch. The harness and safety catch when used either with or without BWS provided safety to the person walking on the treadmill and holds or catches the person if he or she should lose their balance, stumble or begin to fall (Figure 4-1).

**Procedure**

After equipment set-up and subject preparation, the walking trials over the instrumented treadmill were recorded. First, subjects were asked to stand with one leg on each belt of the instrumented treadmill to record a static trial. The static trial was used to create the subject
specific model by defining joint center locations and segment lengths. The leg chosen for
donning the AFO and the order of testing with and without it was randomized for each subject.
For the AFO trial, each subject was requested to wear a unilateral, size-fitted PAFO. The insole
of the shoe was removed in order to fit the AFO and to even out the limb length on both sides.
The subject walked on the instrumented treadmill for the collection of kinematic and kinetic data
with the overhead safety and harness. Subjects were permitted to practice walking on the
treadmill until they achieved steady state walking at the speed of 1.2m/s and comfort while
walking in this environment. Once the subject felt comfortable at the set speed and the
investigator viewed a steady-state pattern of walking, kinematic and kinetic data was collected
for 30 seconds in each of the two conditions. After data collection, the trial was processed to
verify if all the desired data was collected properly. Rest was provided during testing, as
requested. This experiment took approximately two hours from the start for set-up and data
collection.

**Data Processing**

Kinetic data (Ground reaction forces and moments) and segment kinematic data was low
pass filtered with zero lag digital Butterworth filter (20 and 9 Hz cut-off frequencies
respectively). Software for Interactive Musculoskeletal Modeling (SIMM) was used to create
subject specific models. Segment inertial properties were calculated for each subject based on the
subject’s mass and segment lengths. SIMM and SDFast performed an inverse dynamics analysis
for each trial. All data were averaged across trials for each subject. The kinematic and
kinetic data from each trial was normalized to percent stride using Matlab code and then
compared between the two conditions.
Data Analysis

For each phase of interest, the two conditions (with and without AFO) were compared using a Hotelling’s $T^2$-test, which is a multivariate analogue of the paired t-test. The test is a multivariate extension of the Student's t-test for paired data in comparison of mean difference vectors, i.e. the differences of two or more dependent variables considered twice in the same subjects.\(^{245}\) The primary dependent variables compared included the rate of limb loading, peak braking force, excursion and power at the ankle, knee and hip joints. For the stance to swing phase the excursion and power at the ankle, knee and hip joints were analyzed collectively. For the swing to stance phase the joint excursion, rate of limb loading and peak braking force were analyzed collectively. Interlimb temporal and spatial measures of symmetry i.e. double limb support time and step length, our secondary dependent variables were compared between conditions using a paired t-test. Significance level was set at $p< 0.05$. To correct for multiple comparisons, a Holm's step down method was used that adjusted p-values for each research question.\(^{246}\)

Results

Tables 4-2, 4-3 and 4-4 and Figures 4-2, 4-3, 4-4 & 4-5 show the mean and standard deviation in kinematic and kinetic measures related to each phase with and without the PAFO ipsilaterally. In the stance-to-swing phase a significant decrease in peak ankle plantar flexion, hip extension and peak plantarflexor power were noted while walking with a PAFO. In the stance-to-swing phase, while walking with a PAFO, a significant increase in hip flexion, decrease in the rate of loading and peak braking force were observed. With regards to interlimb coordination with the PAFO, double limb support time increased significantly on the ipsilateral limb.
Discussion

The main finding of the study was that in the stance-to-swing phase, donning a PAFO kinematically decreased peak ankle plantarflexion and hip extension and kinetically decreased plantarflexor power. Additionally, in the swing-to-stance phase, wearing a PAFO kinematically increased hip flexion, decreased ankle dorsiflexion and kinetically decreased peak braking force and increased rate of limb loading.

In stance-to-swing phase, with a PAFO, non-injured subjects demonstrated a significant decrease in hip extension and ankle plantarflexion which coincided with a decrease in propulsive force. Generation of plantarflexor power is vital for forward progression of the body. The ankle plantar flexors provide ~70% of the joint work during walking.\textsuperscript{119,247} However, bracing the ankle decreased plantarflexor power generation by 17%. Reduction in power may have resulted from a decrease in angular velocity or moment and subsequently contributed to slowing limb progression.\textsuperscript{115} Therefore the PAFO reduced the ability of the ankle to contribute to push-off in the stance-to-swing transition phase. Interestingly, the effect of the brace was not only isolated to the ankle but also observed at the hip joint. A decrease in hip extension observed as a result of the brace could lead to the poor stretch of the hip flexor muscles thereby increasing the difficulty in initiating swing.\textsuperscript{38}

In the swing-to-stance phase, the primary function of the PAFO is to prevent footdrop. However, with a PAFO non-injured individuals demonstrated a significant increase in hip flexion and a decrease in ankle dorsiflexion thereby affecting the heel rocker. The heel rocker is the first phase in the gait cycle after initial contact that determines the limb’s loading response.\textsuperscript{33} The momentum generated by the fall of body weight onto the stance limb is preserved by this heel rocker. Normal initial contact is made by the calcaneal tuberosity, which becomes the fulcrum about which the foot and tibia move. With a PAFO, an increase in hip flexion and a
simultaneous decrease in ankle dorsiflexion in the swing-to-stance phase limit the smooth transfer of body weight onto the stance limb. Kinetically, this coincided with the decrease in peak braking force on the PAFO side and a delay in the rate of loading. Temporally, a decreased loading rate corresponded with an increase in the initial phase of double support time on the ipsilateral side indicative of a delay in shifting the weight from one limb to another.

In non-injured individuals, our study demonstrated that an orthosis altered the transition phase kinematics and kinetics crucial to stepping. Our findings could have potential implications in neurologically impaired individuals in whom brace walking is common. Past studies have evaluated the benefits of an ankle foot orthosis in different neurological populations. However these studies have evaluated the compensatory benefits of using an orthosis on temporal and spatial patterns of walking with the device without accounting for changes in joint kinematics and kinetics in the transition phases of walking. Appropriate joint kinematics and kinetics in the transition phases are crucial for providing optimal motion-related sensory input to a compromised nervous system. Utilization of these sensory inputs has been shown to aid in retraining the neuromuscular system for walking recovery. Our study explored this paradigm shift by re-examining the use of such devices for walking by examining joint kinematics and kinetics during the transition phase of walking. In our study, use of an orthosis failed to produce desired proximal joint kinematics such as hip extension in the stance-to-swing phase of walking and meet the functional task requirements such as rate of loading in able-bodied individuals. Since the orthosis affected walking in non-injured individuals its effect on gait in individuals nervous system disorders could be more pronounced. Therefore, the results of our study suggest that the purpose and functional implication of an ankle foot orthosis needs to be evaluated rigourously in neurological populations. Neurobiologically driven recovery based
interventions such as locomotor training targeted at providing normal walking kinematics and thereby appropriate motion related sensory input to a compromised nervous system need to weigh the use of such devices cautiously.

**Limitation**

We did not collect EMG in these individuals which would help correlate our findings with lower limb muscle activity. Also, examination of kinematic and kinetic changes with a PAFO, limits the generalizability of our results to other more rigid devices such as solid or hinged ankle foot orthoses.

**Conclusion**

A minimally restrictive device such as a PAFO in non-injured individuals impacted the provision of critical afferent input during the transition phases of walking. Proximal hip extension crucial for the transition from stance-to-swing and the rate of loading during the swing-to-stance phase were decreased. Intuitively, use of more rigid devices could exaggerate these findings. Non-injured individuals were able to adapt to walking with the PAFO by increasing the double support time ipsilaterally. Given the clinical relevance of our study, the use of a PAFO for neurological populations needs to be systematically assessed.
Figure 4-1. Participant with safety harness walking on an instrumented treadmill. Reflective markers were applied to bony landmarks on the pelvis and bilateral lower extremities for this study.
Figure 4-2. Ipsilateral and contralateral average joint angles with and without the ankle foot orthosis (AFO) ipsilaterally. The ipsilateral decrease in peak hip extension and increase in hip flexion [A], no change in peak knee flexion [B] and decrease in peak ankle plantarflexion and dorsiflexion [C] during the swing-to-stance and stance-to-swing phase of the gait cycle are highlighted by dotted circles. Vertical lines represent point of toe-off in the gait cycle. Significant changes represented as p<0.05.
Figure 4-3. Stick figure representing the changes in individual joint motion in the stance-to-swing and swing-to-stance phase of the gait cycle with and without the ankle foot orthosis (AFO).
Figure 4-4. Ipsilateral and contralateral average joint powers during the stance-to-swing phase of the gait cycle with and without the ankle foot orthosis (AFO) ipsilaterally. The ipsilateral hip flexor power [A], knee flexor power [B] and ankle plantarflexor power [C] are highlighted by dotted circles. Vertical lines represent point of toe-off in the gait cycle. Significant changes represented as p<0.05.
Figure 4-5. Ipsilateral and contralateral vertical and horizontal (AP) ground reaction forces (GRF) during the swing-to-stance phase of the gait cycle with and without the ankle foot orthosis (AFO) ipsilaterally. The prolonged rate of loading during vertical loading [A] and decrease in horizontal braking force [C] are highlighted by dotted circles (p<0.05) ipsilaterally.
Table 4-1. Demographics of the study participants.

<table>
<thead>
<tr>
<th>ID</th>
<th>Age</th>
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<th>Orthotic</th>
<th>Size</th>
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</thead>
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<td>F</td>
<td>R</td>
<td>Small</td>
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<tr>
<td>N2</td>
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<td>Large</td>
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<tr>
<td>N3</td>
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<td>M</td>
<td>R</td>
<td>X-large</td>
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<td>F</td>
<td>R</td>
<td>Medium</td>
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<tr>
<td>N5</td>
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<td>F</td>
<td>R</td>
<td>Medium</td>
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<td>F</td>
<td>R</td>
<td>Medium</td>
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<td>M</td>
<td>R</td>
<td>Large</td>
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<td>Small</td>
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<td>N11</td>
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<td>L</td>
<td>Large</td>
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<td>L</td>
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<td>N13</td>
<td>23</td>
<td>M</td>
<td>L</td>
<td>Large</td>
</tr>
<tr>
<td>N14</td>
<td>36</td>
<td>F</td>
<td>L</td>
<td>Small</td>
</tr>
</tbody>
</table>

Table 4-2. Hip, knee and ankle joint kinematic and kinetic data while walking with and without an ankle foot orthosis (AFO) during the stance-to-swing phase of the gait cycle.

<table>
<thead>
<tr>
<th>Stance-to-swing phase</th>
<th>Without AFO ± Standard deviation</th>
<th>With AFO ± Standard deviation</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Peak hip joint extension (degrees)</td>
<td>-8.67± 5.58</td>
<td>-6.77± 5.51</td>
<td>.001*</td>
</tr>
<tr>
<td>Peak knee joint flexion (degrees)</td>
<td>-64.87± 3.80</td>
<td>-64.54± 4.15</td>
<td>.541</td>
</tr>
<tr>
<td>Peak ankle joint plantarflexion (degrees)</td>
<td>-19.45± 5.71</td>
<td>-12.01± 5.28</td>
<td>.000*</td>
</tr>
<tr>
<td>Hip joint power (Watts/ body weight)</td>
<td>0.11± 0.04</td>
<td>0.12± 0.04</td>
<td>.899</td>
</tr>
<tr>
<td>Knee joint power (Watts/ body weight)</td>
<td>-0.15± 0.02</td>
<td>-0.15± 0.03</td>
<td>.893</td>
</tr>
<tr>
<td>Ankle joint power (Watts/ body weight)</td>
<td>0.18± 0.03</td>
<td>0.15± 0.03</td>
<td>.000*</td>
</tr>
</tbody>
</table>

*Significant changes.
Table 4-3. Hip, knee and ankle joint kinematic and kinetic data while walking with and without an ankle foot orthosis (AFO) during swing-to-stance phase of the gait cycle.

<table>
<thead>
<tr>
<th>Swing-to-stance phase</th>
<th>Without AFO ± Standard deviation</th>
<th>With AFO ± Standard deviation</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Peak hip joint flexion (degrees)</td>
<td>32.75± 4.73</td>
<td>35.60± 5.13</td>
<td>.028*</td>
</tr>
<tr>
<td>Peak knee joint flexion (degrees)</td>
<td>-17.86± 3.47</td>
<td>-17.81± 3.55</td>
<td>.924</td>
</tr>
<tr>
<td>Peak ankle joint plantarflexion (in degrees)</td>
<td>-8.40± 3.55</td>
<td>-10.90± 4.77</td>
<td>.000*</td>
</tr>
<tr>
<td>Rate of loading (N/kg)</td>
<td>0.06± 0.01</td>
<td>0.05± 0.01</td>
<td>.018*</td>
</tr>
<tr>
<td>Peak braking force (N/kg)</td>
<td>-0.16± 0.02</td>
<td>-0.15± 0.02</td>
<td>.013*</td>
</tr>
</tbody>
</table>

*Significant changes.

Table 4-4. Average interlimb temporal and spatial data while walking with and without an ankle foot orthosis (AFO).

<table>
<thead>
<tr>
<th></th>
<th>No AFO Primary side</th>
<th>Contralateral side</th>
<th>p-value</th>
<th>Unilateral AFO Primary side</th>
<th>Contralateral side</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Step length (meters)</td>
<td>0.57± 0.14</td>
<td>0.58± 0.14</td>
<td>.329</td>
<td>0.56± 0.15</td>
<td>0.58± 0.13</td>
<td>.326</td>
</tr>
<tr>
<td>Double support time (seconds)</td>
<td>0.19± 0.02</td>
<td>0.19± 0.02</td>
<td>.111</td>
<td>0.22 ±0.03</td>
<td>0.17± 0.02</td>
<td>.000*</td>
</tr>
</tbody>
</table>

*Significant changes.
CHAPTER 5
COMPARISON OF WALKING WITH AND WITHOUT ANKLE FOOT ORTHOSIS IN PERSONS WITH INCOMPLETE SPINAL CORD INJURY-1B

Given the gait deficits after ISCI, two investigative steps that would serve to inform clinical-decision making for use of an AFO during walking retraining are 1) to examine the transition phase mechanics in persons with ISCI while walking with and without the AFO and 2) to compare the observed mechanics in each of the conditions to normal walking mechanics. Therefore we proposed to assess the changes in walking mechanics in individuals with ISCI while walking with an AFO and to examine the proximity or deviation of the observed change to normal matched control values.

Specific Aims and Hypothesis

• **Aim 1:** To compare the immediate effect of walking with and without wearing an AFO on lower extremity kinematics and kinetics during the stance-to-swing and swing-to-stance phase of walking in persons after ISCI.

• **Hypothesis 1a:** Compared to walking without wearing a AFO, wearing an AFO in persons with ISCI will significantly change the stance-to-swing transition observed on the AFO side: specifically, peak ankle plantar flexion, knee flexion and hip extension and peak knee, hip and plantarflexor powers.

• **Hypothesis 1b:** Compared to walking without wearing an AFO, wearing an AFO in persons with ISCI will significantly change the swing-to-stance transition observed on the AFO side: specifically, ankle plantarflexion, knee and hip flexion and rate of loading and braking force.

• **Aim 2:** To compare lower extremity kinematics and kinetics during the stance-to-swing and swing-to-stance phase while walking with and without wearing an AFO in individuals with ISCI to that of healthy, non-injured age, weight, height and speed matched controls.

• **Hypothesis 2a:** In persons with ISCI, the stance-to-swing transition will be significantly deviated from normal while walking with an AFO compared to walking without one. Specifically, with an AFO there will be a decrease in peak ankle plantar flexion and hip extension and increase in peak knee flexion and an increase in peak knee and hip flexor powers and decrease peak ankle plantarflexor power.

• **Hypothesis 2b:** In persons with ISCI, the swing-to-stance transition will be significantly deviated from normal while walking with an AFO compared to walking without one. Specifically, with an AFO there will be a decrease in ankle plantarflexion and increase in knee and hip flexion and a decrease in rate of loading and peak braking force.
Methods

Subject Selection

Eight persons with ISCI ranging between 18-80 years and their height, weight and age matched controls were recruited for this experiment and signed an informed consent form approved by University of Florida Institutional Review Board and the Veteran Affairs Subcommittee for Clinical Investigation. Participant demographics are tabulated in Table 5-1. American Spinal Injury Association (ASIA) motor score and impairment scale data were collected from all participants with ISCI to assess the degree of impairment in each leg. The criteria for inclusion in the study were as follows:

1) Persons with ISCI classified as ASIA D
2) Medically stable
3) Have quadriceps strength of at least 3/5
4) Have decreased ankle strength (dorsiflexor strength of less than or equal to 4/5)
5) and/or absent or impaired proprioception at the ankle
6) Can stand unaided for one minute
7) Can walk with minimal assistive device such as the cane but does not use an ankle foot orthosis.

Exclusion criteria included persons who were unable to follow 3 step commands, amputation, medical instability, significant musculoskeletal problems other than SCI that limit hip and knee extension or ankle plantarflexion to neutral. Sample size was determined from previous studies examining similar gait characteristics in these individuals. Pepin et al. found significant differences (p<0.01) in hip extension in a sample size of seven persons with ISCI compared to their non-injured counterparts at matched speeds while comparing the adaptability of gait pattern in individuals with ISCI to different walking speeds. The change in hip joint angle (hip extension change from pilot data=5.25 degrees, SD from Pepin's study=±5)
were used to calculate sample size for our study. It was determined that a sample of 8 subjects will be required to reach an alpha level of 0.05 and a power of 0.80.

**Experimental Set-up**

Once the subject had read and signed the informed consent form, motion data was collected and analyzed using a 3-D motion analysis system in conjunction with an ADAL3D instrumented split-belt treadmill custom manufactured and calibrated by TECMACHINE (Cédex, France), mounted flush with the floor and anchored to the foundation. Four Kistler piezoelectric sensors on each half treadmill allow calculation of the two-dimensional location of the center of pressure (COP) and the moment about the vertical axis, in addition to the three-dimensional ground reaction force, under each foot. Belt speeds can be controlled as slow as 0.1 m/s.

Ground reaction forces were recorded at 1000Hz for each limb when in contact with the treadmill belt. The force plates were allowed to warm-up for at least 15 minutes as per manufacturer guidelines and calibrated prior to data collection. The walking pattern of the subjects were captured and analyzed by a Vicon three-dimensional motion analysis system. The system consists of the VICON 612 Datastation with twelve active video channels and a 64 Channel A/D Board for analog signals. There were twelve 1000Hz M2-cameras (Digital CMOS M2 series cameras have a resolution of 1280 x 1024). Included software was: Workstation, Polygon, BodyBuilder, Plug-In Gait, Plug-In Modeller, and Real Time II. The twelve cameras had a frame rate of 60-120 fps and used infrared (IR) light-emitting diode strobes, which were gen-locked. Static calibration of the system used the clinical L-frame, which contains 4 retro reflective markers, being placed in a predetermined position on the motion analysis force platforms. Following this a dynamic calibration was done using a 500mm wand that was moved around the capture area for approximately 20 seconds. Analog video data were also collected.
using a standard camcorder recording at 100 fps with its optical axis perpendicular to the plane of interest (i.e. the sagittal plane of motion).

**Subject Preparation**

Subjects were asked to wear tennis shoes and change into appropriate clothing (dark colored cycling shorts and shirt) for testing. For trials using the AFO, each subject was fitted with an off the shelf posterior leaf spring ankle foot orthosis (PAFO). Fitting was assessed by measuring fit inside shoe, length of the calf shell and that of the footplate. Standardized fitting included using a PAFO whose length fits an inch to two below the fibular head when donned and whose footplate length extends till the tip of the toes. Lightweight retro-reflective markers were attached to the following bony landmarks: posterior superior iliac spines (PSIS), anterior superior iliac spines (ASIS), knee-joint axes, lateral malleoli, medial malleoli, clavicular notch, sternum, C7, T10, and acromium processes. The second foot ray, base of the 5th metatarsal and the heel markers were approximated on the subjects’ shoes. Clusters of markers were attached to the pelvis, thigh, shank, and foot segments. This modified Helen Hayes marker set is commonly used to capture bilateral 3D kinematics using a twelve-camera VICON motion analysis system.

Each subject was fitted with a body weight supporting harness equipped with an additional overhead safety catch. The harness and safety catch when used either with or without BWS provided safety to the person walking on the treadmill and holds or catches the person if he or she should lose their balance, stumble or begin to fall (Figure 4-1).

**Procedure**

After equipment set-up and subject preparation, the walking trials over the instrumented treadmill were recorded. First, subjects were asked to stand with one leg on each belt of the instrumented treadmill to record a static trial. The static trial was used to create the subject
specific model by defining joint center locations and segment lengths. The leg chosen for
donning the AFO and the order of testing with and without it was randomized for each subject.
For the AFO trial, each subject was requested to wear a unilateral, size-fitted PAFO. The insole
of the shoe was removed in order to fit the AFO and to even out the limb length on both sides.
The subject walked on the instrumented treadmill for the collection of kinematic and kinetic data
with the overhead safety and harness. Subjects were permitted to practice walking on the
treadmill until they achieved steady state walking at the speed of 1.2m/s and comfort while
walking in this environment. Once the subject felt comfortable at the set speed and the
investigator viewed a steady-state pattern of walking, kinematic and kinetic data was collected
for 30 seconds in each of the two conditions. After data collection, the trial was processed to
verify if all the desired data was collected properly. Rest was provided during testing, as
requested. This experiment took approximately two hours from the start for set-up and data
collection.

**Data Processing**

Kinetic data (Ground reaction forces and moments) and segment kinematic data was low
pass filtered with zero lag digital Butterworth filter (20 and 9 Hz cut-off frequencies
respectively). Software for Interactive Musculoskeletal Modeling (SIMM) was used to create
subject specific models. Segment inertial properties were calculated for each subject based on the
subject’s mass and segment lengths. SIMM and SDFast performed an inverse dynamics analysis
for each trial.  All data were averaged across trials for each subject. The kinematic and
kinetic data from each trial was normalized to percent stride using Matlab code and then
compared between the two conditions.
Data Analysis

For each phase of interest, the two conditions (with and without AFO) were compared using a Hotelling’s T²-test, which is a multivariate analogue of the paired t-test. The test is a multivariate extension of the Student's t-test for paired data in comparison of mean difference vectors, i.e. the differences of two or more dependent variables considered twice in the same subjects. The dependent variables compared included the rate of limb loading, peak braking force, excursion and power at the ankle, knee and hip joints. For the stance-to-swing phase, the excursion and power at the ankle, knee and hip joints were analyzed collectively. For the swing-to-stance phase, the joint excursion, limb loading and peak braking force were analyzed collectively. Interlimb temporal and spatial measures of symmetry i.e. double limb support time and step length, our secondary dependent variables were compared between conditions using a paired t-test. Significance level was set at p< 0.05. To correct for multiple comparisons, a Holm's step down method was used that adjusted p-values for each research question. The same analyses were repeated comparing the control data to individuals with ISCI walking with and without the AFO.

Results

Figures 5-1 through Figure 5-6 and Tables 5-2 through Table 5-8 show the change in the kinematic and kinetic measures related to each gait phases with and without the PAFO ipsilaterally in individuals with ISCI and their matched non-injured controls.

Within-Subject Comparisons for Individuals with ISCI

In the stance-to-swing phase, with a PAFO, a decrease in hip extension was observed within subjects (Figure 5-1 & Table 5-2). Likewise, in the swing-to-stance phase, with a PAFO, an increase in peak knee flexion was observed within subjects (Figure 5-1 & Table 5-3). After
correcting for multiple comparisons in these phases, weak statistical support existed for these variables.

**Between-Subject Comparisons for Individuals with ISCI Walking Without a PAFO and Their Matched Control**

In the stance-to-swing phase, without a PAFO, a decrease in peak knee joint flexion and knee joint power were observed in individuals with ISCI. After correcting for multiple comparisons in this phase, weak statistical support existed for these variables (Figure 5-1 & Table 5-4). In the swing-to-stance phase, without a PAFO, a significant increase in peak hip flexion was observed in individuals with ISCI, which after correcting for multiple comparisons was statistically significant (Figure 5-1 & Table 5-5).

**Between-Subject Comparisons for Individuals with ISCI Walking with a PAFO and Their Matched Control**

In the stance-to-swing phase, with a PAFO, a decrease in peak knee joint flexion and knee joint power were observed in individuals with ISCI. After correcting for multiple comparisons in this phase, weak statistical support existed for these variables (Figure 5-1 & Table 5-6). In the swing-to-stance phase, with a PAFO, an increase in peak hip flexion was observed in individuals with ISCI, which after correcting for multiple comparisons was statistically significant (Figure 5-1 & Table 5-7).

**Temporal and Spatial Comparisons Within Subjects**

Both the spatial and temporal measures of symmetry namely step length and double limb support time did not demonstrate significant differences between the two walking conditions (Figure 5-5, 5-6 & Table 5-8).

**Discussion**

The main finding of the study was that in individuals with ISCI, donning a PAFO decreased hip extension in the stance-to-swing phase of walking compared to walking without it.
(Figure 5-4 & 5-8). While walking with a PAFO, step length value and interlimb symmetry did not change between conditions. Interestingly, double limb support time on the ipsilateral limb increased in 5/8 subjects while walking with a PAFO (Figure 5-6). Additionally, symmetry indices for the interlimb double limb support time increased concurrent with the increase in ipsilateral double limb support time.

Furthermore, compared to normal walking, gait in individuals with ISCI walking without a PAFO was characterized by a significant decrease in knee flexion in the stance-to-swing phase that also correlated to a decrease in knee flexor power. Likewise, a significant increase in hip flexion in the swing-to-stance phase of walking was observed. Interestingly, we did not observe a trend for improvement or deterioration in these pre-existing gait deviations with the PAFO when compared to matched controls. The increase in hip flexion in the swing-to-stance transition phase and decrease in knee flexion in the stance-to-swing phase are common gait deviations characteristic to individuals with ISCI. Pepin et al (2003) has demonstrated a significant increase in hip flexion at the time of heel contact in individuals with ISCI.250 Likewise the reduction in knee angular velocity during walking is a common gait deviation observed in individuals with ISCI which could account for the decrease in knee flexion.70

With regards to power generation, during the stance-to-swing phase, in both the conditions, individuals with ISCI demonstrated an ipsilateral decrease in ankle plantarflexor, knee flexor and hip flexor power compared to the contralateral limb. Generation of plantarflexor power is vital for forward progression of the body. The ankle plantar flexors provide ~70% of the joint work during walking.119,247 However, wearing a PAFO did not augment power generation at the ankle. Additionally, a decrease in horizontal propulsive and braking force were also noted on the ipsilateral side with and without PAFO compared to the contralateral limb. The decrease in
propulsive force could result from inability to generate sufficient power ipsilaterally. Likewise, the reduced braking force compared to the contralateral limb could result from the significant increase in hip flexion observed in these individuals.

**Clinical Implication**

Unlike past studies reporting the compensatory benefits of using an ankle foot orthosis, this study uniquely examined the ability of the PAFO to meet the normal kinematic and kinetic task requirements of stepping. Failure to improve limb kinematics and kinetics with a PAFO during treadmill walking is suggestive of the inability of the device to provide a normal walking pattern in individuals after ISCI. Importantly, a distally worn PAFO impacted proximal joint excursion by limiting hip extension. Hip extension is one of the essential kinematic features of the stance-to-swing transition during walking. Studies have shown that preventing the hip from attaining an extended position inhibited the generation of the flexor burst and hence the onset of the swing phase. The use of PAFO for step retraining on a treadmill after SCI may thus hinder achievement of the task-specific, locomotion-related afferent input used to retrain stepping via sensorimotor activation of the neuromuscular system.

**Limitations**

The results of our study demonstrated weak statistical support for within subject comparisons. We had performed our initial power and sample estimates based on preliminary data for our primary outcome of interest: hip extension. Consequently, we found this variable was different between conditions. However, the secondary variables remained under powered to find true differences if they existed. With a larger sample size very small differences would be detected as significant. However, statistical significance needs to be assessed with caution since it does not imply if the difference between the variables is large or important. Additionally, apart
from the kinematic and kinetic data, collection of electromyographic data in these individuals would have helped correlate our findings with lower limb muscle activity.

Conclusion

For the rehabilitation specialist, the characterization of gait deficits observed in ISCI subjects is important for treatment purposes. Likewise, characterization of improvement or deterioration in these deficits with the use of orthotic devices is important particularly for developing and bettering new rehabilitation approaches. As newer interventions are being developed, the therapeutic rationales for the use of orthotic devices might change based on the guiding principles of these interventions. Traditionally, wearing an AFO in individuals with ISCI has been considered compensatory solution or a “quick fix” for remediating gait deficits resulting from muscular weakness, incoordination and spasticity. However, from a neurobiological control of walking based perspective, an AFO may alter the sensory experience necessary for retraining the nervous system and might not produce the desired therapeutic effect.24,252

Interestingly, in our study, the use of a minimally restrictive PAFO decreased hip extension in participants with ISCI. The observed decrease could impact the provision of at least one critical afferent input key to the restoration of walking. Furthermore, use of more rigid devices is likely to exaggerate our findings. Consequently, if the goal of recovery based interventions such as locomotor training is to provide optimal limb kinematics, the use of a PAFO for stepping would not coincide with the principles of training.
Figure 5-1. Ipsilateral and contralateral average joint angles during the swing-to-stance and stance-to-swing phase of the gait cycle with and without the ankle foot orthosis (AFO) ipsilaterally. The ipsilateral peak hip flexion and extension [A], peak knee flexion [B] and peak ankle plantarflexion and dorsiflexion [C] are highlighted by dotted circles. The gray shaded area represents matched control data. Vertical lines represent point of toe-off in the gait cycle. Significant changes in the joint angles represented as p<0.05.
Figure 5-2. Ipsilateral and contralateral average joint powers during the swing-to-stance and stance-to-swing phase of the gait cycle with and without the ankle foot orthosis (AFO) ipsilaterally. The ipsilateral hip flexor power [A], knee flexor power [B] and peak ankle plantarflexor power [C] are highlighted by dotted circles. The gray shaded area represents matched control data. Vertical lines represent point of toe-off in the gait in the gait cycle. Significant changes in the joint powers represented as p<0.05.
Figure 5-3. Ipsilateral and contralateral vertical and horizontal (AP) ground reaction forces (GRF) during the swing-to-stance phase of the gait cycle with and without the ankle foot orthosis (AFO) ipsilaterally. The rate of loading during vertical loading [A] and the horizontal braking force [C] are highlighted by dotted lines ipsilaterally. The gray shaded area represents matched control data.
Figure 5-4. Ipsilateral hip extension values with and without the AFO during the stance-to-swing phase of the gait cycle for spinal cord injured individuals and their matched controls.
Figure 5-5. Step length while walking with and without the AFO in individuals with incomplete spinal cord injury.
Figure 5-6. Double limb support time while walking with and without the PAFO in individuals with incomplete spinal cord injury.
Table 5-1. Participant demographics of individuals with ISCI and control subjects.

<table>
<thead>
<tr>
<th>ID</th>
<th>Age</th>
<th>Sex</th>
<th>Height</th>
<th>Orthotic</th>
<th>Injury level</th>
<th>ASIA score</th>
<th>Injury duration months</th>
<th>Speed (m/sec)</th>
<th>Assistive/Orthotic device</th>
</tr>
</thead>
<tbody>
<tr>
<td>I1</td>
<td>46</td>
<td>M</td>
<td>5'6&quot;</td>
<td>L</td>
<td>C5-C6</td>
<td>D</td>
<td>10</td>
<td>0.8</td>
<td>NA</td>
</tr>
<tr>
<td>I2</td>
<td>33</td>
<td>M</td>
<td>5'11&quot;</td>
<td>R</td>
<td>C6-7</td>
<td>D</td>
<td>14</td>
<td>0.6</td>
<td>Cane on left</td>
</tr>
<tr>
<td>I3</td>
<td>66</td>
<td>M</td>
<td>6' 3&quot;</td>
<td>L</td>
<td>C7</td>
<td>D</td>
<td>79</td>
<td>0.5</td>
<td>Cane on right</td>
</tr>
<tr>
<td>I4</td>
<td>49</td>
<td>F</td>
<td>5'5&quot;</td>
<td>L</td>
<td>C4-C5</td>
<td>D</td>
<td>46</td>
<td>0.7</td>
<td>NA</td>
</tr>
<tr>
<td>I6</td>
<td>49</td>
<td>F</td>
<td>5'10&quot;</td>
<td>L</td>
<td>C7</td>
<td>D</td>
<td>23</td>
<td>0.7</td>
<td>NA</td>
</tr>
<tr>
<td>I7</td>
<td>40</td>
<td>F</td>
<td>5'8&quot;</td>
<td>L</td>
<td>C2-T1</td>
<td>D</td>
<td>253</td>
<td>0.5</td>
<td>NA</td>
</tr>
<tr>
<td>I9</td>
<td>25</td>
<td>M</td>
<td>5'11&quot;</td>
<td>R</td>
<td>T4-5</td>
<td>D</td>
<td>90</td>
<td>0.4</td>
<td>Solid right AFO</td>
</tr>
<tr>
<td>I10</td>
<td>57</td>
<td>M</td>
<td>6'2&quot;</td>
<td>L</td>
<td>C5</td>
<td>D</td>
<td>122</td>
<td>0.3</td>
<td>Cane on right</td>
</tr>
</tbody>
</table>

| C1 | M   | 5'7" | L      |          |              |            |                        | 0.8           |                           |
| C2 | 32  | M   | 5'10"  | R        |              |            |                        | 0.6           |                           |
| C3 | 62  | M   | 6'3"   | L        |              |            |                        | 0.5           |                           |
| C4 | 49  | F   | 5'3"   | L        |              |            |                        | 0.7           |                           |
| C6 | 52  | F   | 5'7"   | L        |              |            |                        | 0.7           |                           |
| C7 | 40  | F   | 5'6"   | L        |              |            |                        | 0.5           |                           |
| C9 | 27  | M   | 5'11"  | R        |              |            |                        | 0.4           |                           |
| C10| 52  | M   | 5'10"  | L        |              |            |                        | 0.3           |                           |
Table 5-2. Hip, knee and ankle joint kinematic and kinetic data during the stance-to-swing phase of the gait cycle while walking with and without an ankle foot orthosis (AFO) in individuals with incomplete spinal cord injury.

<table>
<thead>
<tr>
<th>Stance-to-swing</th>
<th>Without AFO</th>
<th>With AFO</th>
<th>Confidence Interval</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td>Lower</td>
<td>Upper</td>
</tr>
<tr>
<td>Peak hip joint extension (degrees)</td>
<td>-2.57± 10.57</td>
<td>-1.18± 9.72</td>
<td>-2.862</td>
<td>-.313</td>
</tr>
<tr>
<td>Peak knee joint flexion (degrees)</td>
<td>-48.14± 8.94</td>
<td>-47.61± 8.85</td>
<td>-3.114</td>
<td>2.057</td>
</tr>
<tr>
<td>Peak ankle joint plantarflexion (degrees)</td>
<td>-6.32± 8.23</td>
<td>-2.97± 5.22</td>
<td>-6.863</td>
<td>.173</td>
</tr>
<tr>
<td>Hip joint power (Watts/ Body weight)</td>
<td>0.04± 0.02</td>
<td>0.04± 0.02</td>
<td>-.006</td>
<td>.006</td>
</tr>
<tr>
<td>Knee joint power (Watts/ Body weight)</td>
<td>-0.03± 0.02</td>
<td>-0.03± 0.01</td>
<td>-.006</td>
<td>.001</td>
</tr>
<tr>
<td>Ankle joint power (Watts/ Body weight)</td>
<td>0.05± 0.03</td>
<td>0.08± 0.09</td>
<td>-.111</td>
<td>.051</td>
</tr>
</tbody>
</table>

* Represents significant changes
Table 5-3. Hip, knee and ankle joint kinematic and kinetic data during the swing-to-stance phase of the gait cycle while walking with and without an ankle foot orthosis (AFO) in individuals with incomplete spinal cord injury.

<table>
<thead>
<tr>
<th>Swing-to-stance</th>
<th>Without AFO</th>
<th>With AFO</th>
<th>Confidence interval</th>
<th>Lower</th>
<th>Upper</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Peak hip joint flexion</td>
<td>31.15± 8.10</td>
<td>32.13± 6.81</td>
<td>-3.096 1.126</td>
<td>.306</td>
<td></td>
<td></td>
</tr>
<tr>
<td>(degrees)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Peak knee joint flexion</td>
<td>-18.31± 7.50</td>
<td>-19.82± 6.16</td>
<td>.016 3.006</td>
<td>.048*</td>
<td></td>
<td></td>
</tr>
<tr>
<td>(degrees)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Peak ankle joint</td>
<td>-6.73± 6.98</td>
<td>-6.87 ± 5.19</td>
<td>-2.637 2.917</td>
<td>.908</td>
<td></td>
<td></td>
</tr>
<tr>
<td>plantarflexion (degrees)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Rate of loading</td>
<td>0.04± 0.01</td>
<td>0.04± 0.01</td>
<td>-.001 .006</td>
<td>.170</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Peak braking force</td>
<td>-0.06 ± 0.02</td>
<td>-0.06 ± 0.03</td>
<td>-.011 .011</td>
<td>1.000</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>
* Represents significant changes
Table 5-4. Kinematic and kinetic data during the stance-to-swing phase of the gait cycle at the hip, knee and ankle joints while walking without an ankle foot orthosis (AFO) in individuals with incomplete spinal cord injury compared to their matched, non-injured controls.

<table>
<thead>
<tr>
<th>Stance-to-swing</th>
<th>Without AFO</th>
<th>Control</th>
<th>Confidence Interval</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td>Lower</td>
<td>Upper</td>
</tr>
<tr>
<td>Peak hip joint extension (deg)</td>
<td>-2.57± 10.57</td>
<td>-5.61± 6.42</td>
<td>-10.437</td>
<td>4.747</td>
</tr>
<tr>
<td>Peak knee joint flexion (deg)</td>
<td>-48.14± 8.94</td>
<td>-57± 5.20</td>
<td>-15.327</td>
<td>-2.396</td>
</tr>
<tr>
<td>Peak ankle joint plantarflexion (deg)</td>
<td>-6.32± 8.23</td>
<td>-9.07± 7.51</td>
<td>-12.602</td>
<td>7.087</td>
</tr>
<tr>
<td>Hip joint power</td>
<td>0.04± 0.02</td>
<td>0.03± 0.02</td>
<td>-.018</td>
<td>.003</td>
</tr>
<tr>
<td>Knee joint power</td>
<td>-0.03± 0.02</td>
<td>-0.05± 0.02</td>
<td>-.037</td>
<td>-.001</td>
</tr>
<tr>
<td>Ankle joint power</td>
<td>0.05± 0.03</td>
<td>0.07± 0.03</td>
<td>-.009</td>
<td>.042</td>
</tr>
</tbody>
</table>

* Represents significant changes
Table 5-5. Kinematic and kinetic data during the swing-to-stance phase of the gait cycle at the hip, knee and ankle joints while walking without an ankle foot orthosis (AFO) in individuals with incomplete spinal cord injury compared to their matched, non-injured controls.

<table>
<thead>
<tr>
<th>Swing-to-stance</th>
<th>Without AFO</th>
<th>Control</th>
<th>Confidence interval</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td>Lower</td>
<td>Upper</td>
</tr>
<tr>
<td>Peak hip joint flexion (degrees)</td>
<td>31.15± 8.10</td>
<td>24.93± 6.18</td>
<td>-10.087</td>
<td>-2.353</td>
</tr>
<tr>
<td>Peak knee joint flexion (degrees)</td>
<td>-18.31± 7.50</td>
<td>-14.98± 4.57</td>
<td>-3.894</td>
<td>10.561</td>
</tr>
<tr>
<td>Peak ankle joint plantarflexion (degrees)</td>
<td>-6.73± 6.98</td>
<td>-6.35± 5.07</td>
<td>-5.455</td>
<td>6.230</td>
</tr>
<tr>
<td>Rate of loading</td>
<td>0.04± 0.01</td>
<td>0.04± 0.01</td>
<td>-.007</td>
<td>.004</td>
</tr>
<tr>
<td>Peak braking force</td>
<td>-0.06 ± 0.02</td>
<td>-0.07± 0.02</td>
<td>-.015</td>
<td>.005</td>
</tr>
</tbody>
</table>

* Represents significant changes
Table 5-6. Changes in the hip, knee and ankle joint kinematics and kinetics during the stance-to-swing phase of the gait cycle while walking with an ankle foot orthosis (AFO) in individuals with incomplete spinal cord injury compared to their matched, non-injured controls.

<table>
<thead>
<tr>
<th>Stance-to-swing</th>
<th>With AFO</th>
<th>Control</th>
<th>Confidence Interval</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>Lower</td>
<td>Upper</td>
<td></td>
</tr>
<tr>
<td>Peak hip joint extension</td>
<td>-1.18± 9.72</td>
<td>-5.61</td>
<td>-11.556</td>
<td>2.694</td>
</tr>
<tr>
<td>(degrees)</td>
<td></td>
<td>± 6.42</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Peak knee joint flexion</td>
<td>-47.61± 8.85</td>
<td>-57± 5</td>
<td>-16.098</td>
<td>-2.675</td>
</tr>
<tr>
<td>(degrees)</td>
<td></td>
<td>± 5.20</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Peak ankle joint plantarflexion</td>
<td>-2.97± 5.22</td>
<td>-9.07</td>
<td>-14.275</td>
<td>2.070</td>
</tr>
<tr>
<td>(degrees)</td>
<td></td>
<td>± 7.51</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hip joint power</td>
<td>0.04± 0.02</td>
<td>0.03± 0.02</td>
<td>-0.018</td>
<td>.003</td>
</tr>
<tr>
<td>(Watts/ Body weight)</td>
<td></td>
<td>± 0.02</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Knee joint power</td>
<td>-0.03± 0.01</td>
<td>-0.05± 0.02</td>
<td>-0.035</td>
<td>-.005</td>
</tr>
<tr>
<td>(Watts/ Body weight)</td>
<td></td>
<td>± 0.02</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Ankle joint power</td>
<td>0.08± 0.09</td>
<td>0.07± 0.03</td>
<td>-0.112</td>
<td>.085</td>
</tr>
<tr>
<td>(Watts/ Body weight)</td>
<td></td>
<td>± 0.03</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

* Represents significant changes
Table 5-7. Kinematic and kinetic changes at the hip, knee and ankle joints during the swing-to-stance phase of the gait cycle while walking without an ankle foot orthosis (AFO) in individuals with incomplete spinal cord injury compared to their matched, non-injured controls.

<table>
<thead>
<tr>
<th>Swing-to-stance</th>
<th>With AFO</th>
<th>Control</th>
<th>Confidence interval</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Lower</td>
<td>Upper</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Peak hip joint</td>
<td>32.13± 6.81</td>
<td>24.93± 6.18</td>
<td>-11.159</td>
<td>.004*</td>
</tr>
<tr>
<td>flexion (degrees)</td>
<td></td>
<td></td>
<td>-3.251</td>
<td></td>
</tr>
<tr>
<td>Peak knee joint</td>
<td>-19.82± 6.16</td>
<td>-14.98± 4.57</td>
<td>-1.533</td>
<td>.116</td>
</tr>
<tr>
<td>flexion (degrees)</td>
<td></td>
<td></td>
<td>11.218</td>
<td></td>
</tr>
<tr>
<td>Peak ankle joint</td>
<td>-6.87 ± 5.19</td>
<td>-6.35± 5.07</td>
<td>-3.939</td>
<td>.788</td>
</tr>
<tr>
<td>plantarflexion (degrees)</td>
<td></td>
<td></td>
<td>4.996</td>
<td></td>
</tr>
<tr>
<td>Rate of loading</td>
<td>0.04± 0.01</td>
<td>0.04± 0.01</td>
<td>-.001</td>
<td>.170</td>
</tr>
<tr>
<td>Peak braking force</td>
<td>-0.06 ± 0.03</td>
<td>-0.07± 0.02</td>
<td>-.020</td>
<td>.197</td>
</tr>
</tbody>
</table>

* Represents significant changes
Table 5-8. Average interlimb temporal and spatial data while walking with and without an ankle foot orthosis (AFO) in individuals with incomplete spinal cord injury.

<table>
<thead>
<tr>
<th></th>
<th>No AFO Ipsilateral side</th>
<th>No AFO Contralateral side</th>
<th>p-value</th>
<th>Unilateral AFO Ipsilateral side</th>
<th>Unilateral AFO Contralateral side</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Step length (meters)</td>
<td>0.31± 0.02</td>
<td>0.31± 0.03</td>
<td>.921</td>
<td>0.31± 0.02</td>
<td>0.31± 0.02</td>
<td>.301</td>
</tr>
<tr>
<td>Double support time (seconds)</td>
<td>0.36± 0.09</td>
<td>0.36± 0.14</td>
<td>.935</td>
<td>0.38± 0.13</td>
<td>0.35± 0.10</td>
<td>.273</td>
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CHAPTER 6
PHASE DEPENDENT MODULATION OF SOLEUS H-REFLEX IN HEALTHY, NON-INJURED INDIVIDUALS WHILE WALKING WITH AN ANKLE FOOT ORTHOSIS

Introduction

Individuals with incomplete spinal cord injury (ISCI) have weakness and/or spasticity of the musculature below the level of injury making it difficult to meet the functional demands of gait.\(^{68-70}\) In conventional rehabilitation practice, spasticity or loss of muscle strength are substituted by compensatory orthotic devices that stabilize, realign and control the range of excursion of the weakened joint or limb segment to assist with walking.\(^{15,75,78}\) For example, the posterior leaf spring ankle foot orthosis (PAFO) is used to compensate for deficient push off in terminal stance and foot drag in swing.\(^{16,17,203}\) With a PAFO, improvement in overground walking outcomes such as walking speed, stride length, stance knee position and walking energetics have been documented.\(^{236}\)

In spite of the appeal of such compensatory strategies, their use in neurobiologically driven, recovery-based interventions such as locomotor training for individuals with ISCI is still controversial.\(^{24}\) This is due to the lack of information about the use of the device in optimizing or hindering afferent input from joint, muscle and cutaneous receptors fundamental to the training.\(^{34,36,37,133}\) After SCI, pattern generating neural networks within the spinal cord increases their reliance on motion-related afferent input from these receptors for maintaining locomotor control.\(^{25,26,44}\) Limiting ankle excursion with a PAFO may alter the interconnected limb joint assembly specific to walking and in turn negatively influence the afferent information critical for stepping.

The soleus H-reflex has been commonly employed as a neural probe in interpreting the interplay of afferent input and movement control.\(^{149,228,253,254}\) Elicitation of the reflex and measurement of its amplitude have provided insights in spinal transmission during the
performance of a motor task. During walking, the modulation of reflex amplitude is a measure of
the regulation of afferent feedback during different phases of the step cycle.\textsuperscript{146,228,229,253} For
example, reflex amplitude increases to maximum during stance and decreases rapidly to
minimum during swing. This regulation is required to accommodate the functional requirements
of the task, i.e. facilitate weight support and plantarflexion during stance while allowing ankle
dorsiflexion during swing.\textsuperscript{146,228,229,253}

In healthy, non-injured individuals, an increase in peroneal H-reflex amplitude has been
observed while wearing a brace in static sitting or standing position suggestive of a heightened
sensorimotor response due to stimulation of the cutaneous mechanoreceptors.\textsuperscript{154,205,255} Increasing
cutaneous input from the sole of the foot leads to a reduction in Ia presynaptic inhibition of the
soleus muscle. The predicted outcome of this is facilitation of soleus H-reflex amplitude.
Therefore, use of an orthotic device touching the plantar surface of the foot and limiting the
range of motion at the ankle could alter the rich sensory information processed from the ankle-
foot complex and potentially modulate reflex activity in non-injured individuals. However, due
to task-specific nature of H-reflex amplitude, it is difficult to extrapolate the results of a static
task to the dynamic task of walking.\textsuperscript{256} Examination of reflex amplitude while walking with a
PAFO will be useful in determining functional implication of the device in the task of walking.
The purpose of this study was to examine the phase dependent modulation of the H-reflex in the
gait cycle with and without a PAFO in non-injured individuals.

\textbf{Specific Aims}

\begin{itemize}
  \item \textbf{Aim:} To compare phase specific modulation of the soleus H-reflex amplitude in non-
injured individuals while walking with and without a PAFO,
  \item \textbf{Hypothesis:} In non-injured individuals, soleus H-reflex amplitude while walking with an
  AFO will be significantly larger compared to the H-reflex amplitude without an AFO
\end{itemize}
Methods

Subject Selection

A sample of convenience consisted of fourteen healthy, non-injured individuals living independently in the Gainesville community. Each participant provided informed consent before participating in the study. The University of Florida Institutional Review Board and the Veteran Affairs Subcommittee had approved the study for clinical investigation. Age range for our participants was between 18-60 years. Participant demographics are tabulated in Table 6-1. This study did not include any subjects with any detectable gait and postural disorders. Subjects were screened for a medical history of any neurological, musculoskeletal or orthopedic problem that may affect their walking performance over the treadmill.

Using the effect size and standard deviation from pilot data for non-injured subjects (H/M ratio post-pre =0.08, SD from previous study=0.11), to achieve statistical power of 80% at an alpha level of 0.05 we needed 14 normal subjects.

Experimental Set-up

Soleus H-reflexes were evoked, for the purpose of consistency, on the dominant side of healthy, non-injured individuals subjects. Skin was shaved and cleaned for application of electrodes. A bipolar (2 cm inter-electrode distance) Ag–AgCl surface electrode (Therapeutics Unlimited, Iowa City, Iowa) was placed longitudinally over the soleus muscle. These electrodes are embedded in an epoxy mount with preamplifier circuitry and a 2-cm interelectrode distance. The preamplifier and second-stage amplifier provide a total amplification of 1000× with a low-frequency cut off of 20 Hz.

To evoke H-reflexes, one millisecond current pulses were delivered via a constant-current stimulator (Grass Instruments, model S8800 with a modified CCU1) using a 2 cm 1/2 sphere silver cathode placed in the popliteal fossa and a 10 cm silver anode positioned just superior to
the patella. The tibial nerve was localized, in the popliteal fossa by the electrode placement, to evoke a soleus H-reflex at the least current intensity required. Data were acquired at a sample rate of 10 kHz per channel and stored digitally with a commercially available data acquisition system (Data-Pac III by Run Technologies) in a personal computer (Dell Systems, Intel Celeron).

**Subject Preparation**

Subjects were asked to wear tennis shoes and change into appropriate clothing (dark colored cycling shorts and shirt). Skin was shaved and cleaned for application of surface electrodes. For trials using the PAFO, each subject was fitted with an off-the-shelf PAFO. The leg chosen for donning the PAFO and the order of testing with and without it was randomized for each subject. For the PAFO trial, each subject was requested to wear a unilateral, size-fitted PAFO. The insole of the shoe was removed in order to fit the AFO and to even out the limb length on both sides. Fitting of the AFO was assessed by measuring fit inside shoe, length of the calf shell and that of the footplate. Standardized fitting included using an AFO whose length fits an inch to two below the fibular head when donned and whose footplate length extends till the tip of the toes. During treadmill walking, footswitches were placed inside the shoes that were helpful in determining the phases of walking. Each subject was fitted with a body weight supporting harness equipped with an additional overhead safety catch. The harness and safety catch when used either with or without BWS provided safety to the person walking on the treadmill and held or caught the person if he or she lost their balance, stumbled or began to fall.

**Procedure**

The order of testing was randomized for walking with and without PAFO (Figure 6-1). Prior to eliciting reflexes during walking, H-reflexes were first elicited in static standing position for use as a control reference across trials. For this purpose, participants were asked to
stand quietly and H-reflexes were collected in this position. Stimulus intensity was maximized and three maximum M-waves were recorded in the static condition. At least fifteen H-reflexes were then elicited at stimulus intensity within a range of 8-12 % of the M-max in the static position. A recruitment curve was constructed in the static stance position to ensure that the H-reflex was on the ascending limb of the curve.138,149

During walking trials in either of the conditions, H-reflexes were elicited across ten time divided phases of the gait cycle determined by footswitches namely heel strike (HS), HS+100ms, HS+200ms, HS+300ms, HS+400ms and toe-off (TO), TO+100ms, TO+200ms, TO+300ms, TO+400ms. The event and time point of stimulation was achieved by connecting the footswitches to a Schmidt trigger that sensed the event and delivered the pulse. For example, for the phase of heel strike the pulse was delivered at 0ms. For mid-stance, the pulse was delivered after a time delay from heel strike.

Once the subject began stepping on the treadmill at 1.2 m/s, three maximum M-waves were recorded in each of the ten phases of the gait cycle (Figure 6-1).227,257 These recordings were used in determining the stimulus intensity for each tested phase in the gait cycle and were also used for subsequent normalization of the data. Subsequently, stimulation was delivered at stimulus intensity within a range of 8-12 % of the M-max calculated in each of the phases of the gait cycle.138,149 At least fifteen stimuli were delivered in a consecutive or an alternating fashion in each of the ten phases of the gait cycle.138,149

During testing, in both the static and walking condition, the activity in the soleus and TA muscle was recorded over a 100 ms duration prior to electrical stimulation.149 This activity was normalized to three maximum voluntary contractions of the TA and soleus collected at the beginning of the experiment. Also, the M-wave was constantly monitored to make readjustments
to the stimulus intensity if required. A fifteen-minute sitting break was provided after completion of static and walking trials in one condition before proceeding with the other.

Data Processing

After filtering and rectification of the data, mean peak-to-peak amplitude of 10 H-reflexes for each phase of the examined gait cycle was calculated and compared between the two conditions (with and without PAFO). Prior to this comparison, the H-reflex values were normalized to M-max values procured in the respective phases (H/M ratio).\textsuperscript{149}

Data Analysis

For the ten phases of the gait cycle, a Hotelling's T\textsuperscript{2}-test, which is a multivariate analogue of the paired t-test, was performed. The test is a multivariate extension of the Student's t-test for paired data in comparison of mean difference vectors, i.e. the differences of two or more dependent variables considered twice in the same subjects.\textsuperscript{245} The dependent variables compared include the H/Mmax amplitude in the 10 phases of the gait cycle (HS, HS+100ms, HS+200ms, HS+300ms, HS+400ms, TO, TO+100ms, TO+200ms, TO+300ms, TO+400ms). Significant changes between the two conditions (with and without PAFO) were identified using the Holm’s correction which corrects for multiple comparisons by adjusting alpha value.\textsuperscript{246} Additionally, a repeated measures ANOVA was also performed to compare M-max amplitude, actual M wave amplitude used for stimulation of the H-reflex and electromyographic activity recorded 100ms prior to stimulation in the TA and soleus muscles between the two conditions across the gait cycle. Significance level was set at p< 0.05. The same analyses were repeated for the contralateral limb.

Results

Both ipsilaterally and contralaterally the mean H/M ratios were not significantly different between the two walking conditions (p>0.05) for any of the phases (Figures 6-2 through Figure
The mean EMG of soleus and tibialis anterior muscles 100 ms prior to the electrical stimulation was not significantly different in both conditions (p<0.05). Additionally, M-max amplitude (Figure 6-6) and the actual M wave amplitude used to evoke the soleus H-reflexes (Figure 6-7) were not significantly different across the gait cycle in both the conditions.

**Discussion**

Peripheral input from the muscle and cutaneous receptors of the ankle-foot complex have been known to modulate soleus H-reflex amplitude during different tasks. The soleus H-reflex is facilitated by excitation of the plantar cutaneous afferents located around the heel.258 Like wise, the change in ankle joint angle has been shown to modulate H-reflex excitability.259,260,154 Therefore, use of an orthotic device touching the plantar surface of the foot and limiting the range of motion at the ankle could alter the rich sensory information processed from the ankle-foot complex and potentially modulate reflex activity. Although we hypothesized that there would be an increase in soleus H-reflex amplitude with an ankle brace the results of our current study show that soleus H-reflex amplitude remains unchanged while walking with a PAFO in healthy, non-injured individuals.

Previous studies have shown that ankle bracing impacts reflex amplitude in static tasks. For example, Nishikawa et al (1999) reported a 10% increase in peroneus longus (PL) H-reflex amplitude after application of a semi-rigid ankle support in the seated, non-weight bearing position.205 The non-weight bearing position and the testing of the peroneus muscle may account for the differing results between the Nishikawa study and this work. Likewise, Schneider et al reported an increase in soleus H-reflex amplitude on passively imposing rapid knee flexion from static stance position.154 Interestingly, not all studies done in a static task have demonstrated an increase in H-reflex amplitude as a result of bracing the ankle joint. For example, Sefton et al found no effect of a semi-rigid ankle brace on the PL H-reflex during an inversion...
perturbation\textsuperscript{261} and no effect on soleus H-reflex during a single limb stance task.\textsuperscript{262} They rationalized their findings to the fact that the external ankle support provided an increase in mechanical stability, negating the need for neuronal adaptation to maintain upright stance.

The essential difference between our studies and previous studies was that our experiment involved the dynamic task of walking with or without the brace rather than a static task. Since the brace-related afferent input did not alter the lower limb reflexes during walking, it appears that this reflex is centrally modulated in healthy, non-injured individuals. Therefore, even if a peripheral influence is shown to have an effect in one task, it does not follow that the same input will be effective in another task. Several studies in the literature have demonstrated more central modulation of reflex activity during locomotion and locomotor like tasks. For example, Garrett et al (1999) and Schneider et al (2000) reported that the soleus H-reflex amplitude did not change when the knee was braced thereby blocking the normal excursion during locomotion.\textsuperscript{154,206} Similarly, Yang and Wheelan (1993) have shown that inactivity of the tibialis anterior or activity of the soleus muscle during the swing phase of gait did not affect phase-specific modulation of the soleus H-reflex during walking.\textsuperscript{254}

In our study, a kinematically significant decrease in ankle plantarflexion and hip extension were observed as a result of walking with the brace compared to walking without one (Refer Figure 4-1). However, in non-injured individuals, kinematic changes during brace walking did not change the modulation pattern through out the gait cycle suggestive of central modulation of reflex activity. In non-injured subjects, such an occurrence where changes in afferent input from the periphery do not alter H-reflex excitability is probably for maintenance of the locomotor task.

In the event of impaired locomotor control as exists after SCI, the reflex modulation might change during brace walking. This is because the ability of the spinal cord to modulate sensory
input and presynaptic inhibition are both altered post-SCI. Previous studies, have systematically reported that greater H/M ratios were recorded in post-SCI subjects than recorded in non-injured controls. For example, electrical excitation of the plantar cutaneous afferents has been shown to facilitate soleus H-reflexes in persons with SCI and depress reflex amplitude in non-injured subjects in sitting. Therefore, although, reflex modulation during walking did not change between conditions in non-injured individuals, our results may not extend to individuals with neurological impairment. However, this remains to be tested experimentally.

**Limitations**

As has been advocated, soleus H-reflex in our study was evaluated during walking rather than at rest because the reflex undergoes task specific modulation. The factors that could be potential confounds in the study such as speed of walking, testing order, background EMG activity and stimulus intensity were controlled. However, extraneous peripheral afferent input from other than sources such as stimulus generated perturbations or pain associated with repeated stimulation could affect the measured H-reflex amplitude. Also, although a maximal M wave was evoked in each of the tested phases for normalization purposes, a recruitment curve was not constructed during the walking trials. A recruitment curve would assess reflex modulation over the range of intensities during walking and also ensured evaluation of the same proportion of the motor neuron pool. Although the testing phases were randomized, as a result of repeated stimulation, post activation depression of the reflex could have occurred affecting the results of the study. However, this is unlikely because studies have shown that synaptic transmission from Ia fibers to motor neurons depends in a complex fashion on the rate of nerve impulses. During movement tasks, stimuli that produce one or two extra impulses in a neuron that is already conducting tens of impulses per second will not produce significant depression.
Conclusion

In summary, non-injured individuals demonstrate central modulation of reflex activity which attenuates extraneous sensory input from the periphery for the maintenance of the locomotor task. A spinal injury disrupting supraspinal pathways can affect this phase specific reflex modulation. In the presence of impaired central modulation, persistent cutaneous input that is usually presynaptically inhibited during the gait cycle could be facilitated reinforcing the walking related impairment. In the light of these findings, an ankle foot orthosis needs to be evaluated systematically in individuals after SCI for the generation of reflex modulation characteristic of normal walking.
Figure 6-1. Experimental design for the examination of changes in soleus H-reflex amplitude in healthy, non-injured individuals while walking with and without an ankle foot orthosis (AFO).
Figure 6-2. Ipsilateral raw soleus H-reflex data while walking with and without AFO at 300ms from heel strike (HS) and toe-off (TO). Vertical red and blue guide bars capture soleus H-reflex event.

Figure 6-3. Contralateral raw soleus H-reflex data while walking with and without AFO at 300ms from heel strike (HS) and toe-off (TO). Vertical red and blue guide bars capture soleus H-reflex event.
Figure 6-4. Ipsilateral mean H-reflex amplitudes with and without AFO normalized to M-max in each phase of the gait cycle. The gait cycle is represented in 100ms increments from heel strike (HS) and toe off (TO).

Figure 6-5. Contralateral mean H-reflex amplitudes with and without AFO normalized to M-max in each phase of the gait cycle. The gait cycle is represented in 100ms increments from heel strike (HS) and toe off (TO).
Figure 6-6. Ipsilateral [A] and contralateral [B] M-max amplitude with and without the AFO across the gait cycle. The gait cycle is represented in 100ms increments from heel strike (HS) and toe off (TO).
Figure 6-7. Ipsilateral [A] and contralateral [B] actual M wave amplitude used to evoke the soleus H-reflex with and without the AFO across the gait cycle. The gait cycle is represented in 100ms increments from heel strike (HS) and toe off (TO).
Table 6-1. Demographics of non-injured participants recruited for the study.

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CHAPTER 7
IMMEDIATE, PHASE DEPENDENT, SOLEUS H-REFLEX MODULATION IN PERSONS WITH INCOMPLETE SPINAL CORD INJURY WHILE WALKING WITH AN ANKLE FOOT ORTHOSIS

Introduction

The soleus H-reflex has often been employed to examine the neural regulation of afferent information during walking. Elicitation of the reflex and measurement of its amplitude have provided a method to evaluate the modulation of the spinal pathways during walking. In healthy, non-injured individuals, the soleus H-reflex undergoes phase-specific modulation to accommodate to the functional requirements of the task. Specifically, the size of the soleus H-reflex is higher during the stance phase and lower during the swing phase of the step cycle. However, in individuals after ISCI, the soleus H-reflex demonstrates decreased depth of modulation. As a consequence, there is a reduced amplitude modulation across the step cycle and it simply remains increased throughout the step cycle. Consequently, lack of reflex modulation contributes to their walking impairments.

During conventional gait training, individuals with ISCI are often prescribed orthotic devices to assist with walking. Orthotic devices such as an ankle foot orthosis (AFO) are used for assisting foot clearance, increasing gait speed, and improving walking endurance. Apart from mechanically aiding foot clearance, studies in healthy, non-injured individuals suggest that an AFO increases afferent feedback from cutaneous receptors in the foot and shank to improve ankle positioning. For example, ankle bracing in a variety of static motor tasks such as sitting or standing have reported an increase in peroneal and soleus H-reflex amplitude in healthy, non-injured individuals. However, the increase in reflex amplitude with the brace has only been documented in healthy, non-injured individuals under static conditions limiting the extrapolation of results to persons with ISCI in dynamic tasks. In individuals with
ISCI with impaired reflex modulation, an orthotic device touching the plantar surface of the foot and limiting the range of motion at the ankle could alter the rich sensory information processed from the ankle-foot complex and potentially modulate reflex activity.

Additionally, in persons with ISCI, cutaneous stimulation of the foot and sole while walking has been suggested as a potential method to restore reflex modulation comparable to that seen in healthy, non-injured individuals.\textsuperscript{155} Furthermore, simulation of walking kinematics using manual assistance, bodyweight support and a treadmill in persons with ISCI improved reflex modulation and overground stepping speed without bracing the ankle.\textsuperscript{181} Consequently, conclusive evidence supporting or not supporting the use of the brace in improving phase-specific modulation during gait training in individuals with ISCI is not apparent. Also, since afferent stimulation of one limb is known exert a considerable influence on the reflex activity of the contralateral limb, the influence of an AFO on contralateral soleus H-reflex modulation also warrants investigation.\textsuperscript{153}

Therefore, the immediate phase-dependent modulation of the soleus H-reflex in persons with ISCI during walking with and without an AFO was examined ipsilaterally and contralaterally. Specifically, we hypothesized an increase in reflex amplitude only ipsilaterally while walking with the AFO compared to walking without one in individuals with ISCI.

**Specific Aims**

- **Aim 1:** In persons with ISCI and ambulatory, to compare immediate phase-dependent modulation of the soleus H-reflex with and without an AFO in the mid-stance phase of walking.

- **Hypothesis 1:** In persons with ISCI, soleus H-reflex amplitude will be significantly larger in mid-stance while walking with an AFO compared to walking without an AFO.

- **Aim 2:** In persons with ISCI and ambulatory, to compare immediate phase-dependent modulation of the soleus H-reflex in with and without an AFO in the mid-swing phase of walking.
• **Hypothesis 2:** In persons with ISCI, soleus H-reflex amplitude PAFO will be significantly larger in mid-swing while walking with an AFO compared to walking without an AFO.

**Methods**

**Subject Selection**

Nine persons with ISCI ranging between 18-80 years were recruited for this experiment and sign an informed consent form approved by University of Florida Institutional Review Board and the Veteran Affairs Subcommittee for Clinical Investigation (Table7-1). American Spinal Injury Association (ASIA) motor score and impairment scale data were collected from all participants with ISCI to assess the degree of impairment in each leg.

The criteria for inclusion in the study were: Medically stable persons with ISCI classified as ASIA D, having quadriceps strength of at least 3/5, having decreased ankle strength and/or impaired or absent proprioception at the ankle, having ankle dorsiflexor strength of < 4/5, able to stand unaided for one minute and walking with minimal assistive devices but not using an AFO. Exclusion criteria include persons who are unable to follow 3 step commands, amputation, medical instability, significant musculoskeletal problems other than SCI that limit hip and knee extension or ankle plantarflexion to neutral.

**Experimental Set-up**

Soleus H-reflexes were evoked, for the purpose of consistency, on the dominant side of healthy, non-injured individuals subjects. Skin was shaved and cleaned for application of electrodes. A bipolar (2 cm inter-electrode distance) Ag–AgCl surface electrode (Therapeutics Unlimited, Iowa City, Iowa) was placed longitudinally over the soleus muscle. These electrodes are embedded in an epoxy mount with preamplifier circuitry and a 2-cm interelectrode distance. The preamplifier and second-stage amplifier provide a total amplification of 1000× with a low-frequency cut off of 20 Hz.
To evoke H-reflexes, one millisecond current pulses were delivered via a constant-current stimulator (Grass Instruments, model S8800 with a modified CCU1) using a 2 cm 1/2 sphere silver cathode placed in the popliteal fossa and a 10 cm silver anode positioned just superior to the patella. The tibial nerve was localized, in the popliteal fossa by the electrode placement, to evoke a soleus H-reflex at the least current intensity required. Data were acquired at a sample rate of 10 kHz per channel and stored digitally with a commercially available data acquisition system (Data-Pac III by Run Technologies) in a personal computer (Dell Systems, Intel Celeron).

**Subject Preparation**

Subjects were asked to wear tennis shoes and change into appropriate clothing (dark colored cycling shorts and shirt). Skin was shaved and cleaned for application of surface electrodes. For trials using the PAFO, each subject was fitted with an off-the-shelf PAFO. The leg chosen for donning the PAFO and the order of testing with and without it was randomized for each subject. For the PAFO trial, each subject was requested to wear a unilateral, size-fitted PAFO. The insole of the shoe was removed in order to fit the AFO and to even out the limb length on both sides. Fitting of the AFO was assessed by measuring fit inside shoe, length of the calf shell and that of the footplate. Standardized fitting included using an AFO whose length fits an inch to two below the fibular head when donned and whose footplate length extends till the tip of the toes.\(^{244}\) During treadmill walking, footswitches were placed inside the shoes that were helpful in determining the phases of walking. Each subject was fitted with a body weight supporting harness equipped with an additional overhead safety catch. The harness and safety catch when used either with or without BWS provided safety to the person walking on the treadmill and held or caught the person if he or she lost their balance, stumbled or began to fall.
**Procedure**

H-reflexes were elicited in two testing conditions (with and without PAFO) and two parts of gait cycle (mid-stance and mid-swing) during walking. The order of testing was randomized for with and without PAFO conditions (Figure 7-1). Treadmill speed was maintained constant for the both the testing conditions. H-reflexes were first be elicited in static standing position. Collecting, H-reflexes in static position served as a control reference across trials since the reflex is not modulated in a static position. For this purpose, participants were asked to stand quietly and H-reflexes were collected in this position. Stimulus intensity was maximized and three maximum M-waves were recorded in the static standing condition. Fifteen H-reflexes were then elicited at stimulus intensity within a range of 8-12 % of the M-max in the static position. A recruitment curve was constructed in the static stance position to ensure that the H-reflex was on the ascending limb of the curve.

Once the subject began stepping on the treadmill at self-selected speed, three maximum M-waves were recorded in mid-stance phase and mid-swing phase respectively. These recordings were used to determine the stimulus intensity for each tested phase in the gait cycle and were also useful for subsequent normalization of the data. Subsequently, stimulation was delivered at stimulus intensity within a range of 8-12 % of the M-max in the mid-stance and mid-swing phase of the gait cycle. At least 15 H-reflexes were recorded in each of the two selected phases of the gait cycle at each of the time points.

During testing, in both the static and walking condition, the activity in the soleus and TA muscle was recorded over duration of 100 ms prior to electrical stimulation. This activity was normalized to the average of three maximum voluntary contractions of the TA and soleus collected at the start of the experiment. Also, the M-wave was constantly monitored to make readjustments to the stimulus intensity if required. Duration between two consecutive electrical
stimulations were randomly maintained at a minimum of three seconds and maximum 5 seconds to avoid post activation depression and habituation. After recording the H-reflexes in walking, the procedure for the static position was repeated again. After data collection, the data was stored for subsequent analysis. A fifteen-minute sitting break was provided after completion of static and walking testing in one condition before proceeding with the other.

As part of a secondary question examining the effect of the ipsilateral brace on contralateral limb the same procedure was repeated on the contralateral side. Only eight of the above participants participated in this part of the study.

Data Processing

After filtering and rectification of the data, mean peak-to-peak amplitude of 10 H-reflexes for each phase of the examined gait cycle was calculated and compared between the two conditions (with and without PAFO). Prior to this comparison, the H-reflex values were normalized to M-max values procured in the respective phases (H/M ratio).

Data Analysis

For the mid-stance and mid-swing phase on the ipsilateral limb, a paired t-test with bracing condition (with or without PAFO) as the independent variable and the H-reflex amplitude as the dependent variable was performed. The same analysis was repeated for the contralateral limb. Significant changes between the two conditions (with and without PAFO) were identified using the Holm’s step-down correction which corrects for multiple comparisons by adjusting alpha value. Additionally, a repeated measures ANOVA was also performed to compare M-max amplitude, actual M wave amplitude used for stimulation of the H-reflex and electromyographic activity recorded 100ms prior to stimulation in the TA and soleus muscles between the two conditions across the gait cycle. Significance level was set at p<0.05.
Results

Figure 7-2 through Figure 7-4 shows the ipsilateral and contralateral H-reflex modulation in mid-stance and mid-swing phase with and without the PAFO ipsilaterally. After correcting for multiple comparisons, the mean H/M ratio was significant importantly for the ipsilateral mid-swing phase of walking (without PAFO: 0.13±0.10 & with PAFO 0.29±0.14). The values for the H/M ratio for all the tested conditions are reported in Table 7-2 and Table 7-3. The mean EMG of soleus and tibialis anterior muscles 100 ms prior to the electrical stimulation did not change systematically in both conditions (Tables 7-4 through Table 7-7). Furthermore the M-max amplitude and the actual Mwave amplitude used for evoking the soleus H-reflex did not change significantly (Figure 7-5 and Figure 7-6).

Discussion

This is the first study to systematically examine the effect of bracing on soleus H-reflex modulation during the task of walking in individuals with ISCI. The main finding of the study was that, there was a significant increase in soleus H-reflex amplitude ipsilaterally in the mid-swing phase while walking with a PAFO. In the absence of a change in the ankle-foot orientation or stretch at the ankle joint with a PAFO, these results are suggestive of an increase in afferent inflow in the mid-swing phase of walking.

In our study, the background EMG activity in both the walking conditions (with and without the PAFO) was similar despite the changes in H-reflex amplitude reinforcing the fact that the modulation of the reflex is not directly dependent on the excitation level of the alpha-motorneurons.146 Likewise the presence or absence of ankle clonus did not affect reflex amplitude in the two walking conditions. Subject I9 who had clonus had a change in mid-swing reflex amplitude similar to subject I4 who did not demonstrate any clonic activity in the soleus muscle.
In individuals with ISCI, an increase in reflex amplitude with the PAFO during the mid-swing phase could be the result of altered presynaptic inhibition. Presynaptic inhibition is important because during walking sensory input from cutaneous and proprioceptive receptors continuously converges on the spinal circuits.\textsuperscript{263,272} These surplus inputs must either be synergistically combined with the motor commands or be appropriately suppressed to minimize interference. Since only certain input requires selective modulation, presynaptic inhibition of sensory input allows suppression of specific inputs to a neuron without influencing other synaptic inputs. The predominant sources of presynaptic inhibition are peripheral inputs from cutaneous afferents and central descending pathways.\textsuperscript{272,273} During walking, cutaneous input from the foot sole is known to modulate reflex activity and change muscle synergies thereby contributing to adaptive locomotor strategies. For example, Bastiaanse et al (2000) observed that load receptors are involved in the regulation of cutaneous reflex responses in order to adapt the locomotor pattern to the environmental conditions.\textsuperscript{274}

The repertoire of adaptive movement strategies is usually limited in individuals with ISCI because the ability of the spinal cord to modulate sensory input and presynaptic inhibition are usually impaired post-injury.\textsuperscript{263,275} Previous studies, have systematically reported that greater H/M ratios were recorded in post-SCI subjects than recorded in non-injured controls.\textsuperscript{156,263-265} The increase in reflex amplitude during brace walking in our study may have occurred because of the stimulation of plantar cutaneous afferents caused by the PAFO. Excitation of the plantar cutaneous afferents facilitates soleus H-reflex in persons with SCI in sitting.\textsuperscript{266} In the absence of supraspinal modulation of reflex activity, peripheral cutaneous inputs may be beneficial to modulate reflex activity.\textsuperscript{155} Nakajima et al (2006) has shown that reflex connections from cutaneous nerves in the foot on to the lower limb muscles are arranged in a highly topographical
manner and may play an important role in limb loading and ground contact in response to tactile sensation. However, from the results of our study two interesting observations can be made about the PAFO during walking.

An increase in reflex amplitude observed in the mid-swing phase during walking with a PAFO suggests that a simple, off-the-shelf orthosis could potentially increase afferent inflow and modulate reflex activity in high functioning individuals with ISCI. However, in individuals with shallow modulation of reflex activity throughout the gait cycle, the increase in afferent inflow especially during the mid-swing phase might be unfavorable for reflex modulation and ultimately walking. Therefore, prior to its use, the type and purpose of orthotic device, the movement task of interest such as walking or cycling and the targeted population needs to be assessed carefully.

Interestingly, no increase in reflex amplitude was noted with the brace in the contralateral limb in mid-stance or mid-swing. Also motion data collected previously did not show a change in limb kinematics with or without the PAFO contralaterally. Our inference further strengthens the idea of a localized cutaneous response of the PAFO on the reflex modulation ipsilaterally.

**Clinical Implications**

Soleus H-reflexes are exaggerated post-SCI. If reflex dysregulation is secondary to disruption of supraspinal inhibitory control mechanisms, then training strategies inhibiting the hyperactive reflex segmentally in a task and phase specific manner may be beneficial for the proper restoration of locomotion. Locomotor training is one such strategy that works on the above principle of providing optimal sensory input to the nervous system to recover walking ability after ISCI. The training provides sensory cues and phasic information related to locomotion. One of the cues pivotal to training is to minimize sensory stimulation that would conflict with sensory information associated with locomotion. The potency of the
training stems from the fact that a single bout of locomotor training is capable of producing significant depression of the exaggerated soleus H-reflexes and improved walking speed in persons with ISCI. Therefore, research efforts are being directed at systematically determining the critical components of locomotor training, that optimize the provision of critical sensory input that aid walking recovery. Within the realms of this goal, our study demonstrates that the integration of clinically acknowledged stepping aids such as the PAFO’s during locomotor training could be counter productive to the recovery of walking post SCI. Therefore such devices should be chosen only after careful consideration of outcome for training purposes.

Limitations

First, we only examined the effect of the PAFO in the mid-stance and mid-swing phase of walking limiting our inferences to only two specific phases of walking. A thorough examination in different phases of the gait cycle might reveal the unique effects of the PAFO within the entire gait cycle. Second, for standardization purposes, we examined the effect of only one type of AFO which limits generalizability to other types of AFO’s. Based on their impairments individuals with ISCI might use customized AFOs which could yield different results.

Conclusions

In persons with ISCI, soleus H-reflex amplitude increased significantly in the mid-swing phase of walking with an AFO compared to walking without it. Our findings suggest that, in the presence of impaired central and peripheral modulation of reflex activity, an ankle foot orthosis that provides persistent cutaneous inputs from the foot sole might contribute to modulating reflex activity. However, increase in afferent input in certain phases of the gait cycle might not always be favorable to the task of walking. Therefore therapeutic interventions targeted at promoting walking recovery in individuals with ISCI should carefully consider the use of such non-adaptable, compensatory orthotic devices that could potentially hinder retraining or reeducation.
of function. Instead, adaptable strategies that have been documented to provide appropriate phase specific sensory input such as functional cutaneous stimulation of the foot or appropriate cueing using manual assistance should be incorporated during training to assist with walking and promote recovery.
Figure 7-1. Experimental design for testing the effect of walking with and without an ankle foot orthosis (AFO) in individuals with incomplete spinal cord injury (ISCI) at their self-selected (SS) walking speed.
Figure 7-2. Average H/M ratio values with and without an AFO in mid-stance and mid-swing phase of walking relative to static standing in the ipsilateral limb.

Figure 7-3. Average H/M ratio values with and without AFO in mid-stance and mid-swing phase of walking relative to static standing in the contralateral limb.
Figure 7-4. Ipsilateral and contralateral raw soleus H-reflex data while walking with and without AFO during mid-stance (MSt) and mid-swing (MSw) phase of the gait cycle. Vertical red and blue guide bars capture soleus H-reflex event.
Figure 7-5. Ipsilateral [A] and contralateral [B] M-max amplitude with and without the AFO in the mid-stance and mid-swing phase of walking.
Figure 7-6. Ipsilateral [A] and contralateral [B] actual M wave amplitude used to evoke the soleus H-reflex with and without the AFO across the gait cycle.
Table 7-1. Participant demographics.

<table>
<thead>
<tr>
<th>ID</th>
<th>Age</th>
<th>Sex</th>
<th>Height</th>
<th>Orthotic</th>
<th>Injury level</th>
<th>ASIA score</th>
<th>Injury duration in months</th>
<th>Speed (m/sec)</th>
<th>Assitive/Orthotic device</th>
</tr>
</thead>
<tbody>
<tr>
<td>I2</td>
<td>33</td>
<td>M</td>
<td>5'11&quot;</td>
<td>R</td>
<td>C6-7</td>
<td>D</td>
<td>14</td>
<td>0.6</td>
<td>Cane on left</td>
</tr>
<tr>
<td>I3</td>
<td>66</td>
<td>M</td>
<td>6' 3&quot;</td>
<td>L</td>
<td>C7</td>
<td>D</td>
<td>79</td>
<td>0.5</td>
<td>Cane on right</td>
</tr>
<tr>
<td>I4</td>
<td>49</td>
<td>F</td>
<td>5'5&quot;</td>
<td>L</td>
<td>C4-5</td>
<td>D</td>
<td>46</td>
<td>0.7</td>
<td>NA</td>
</tr>
<tr>
<td>I5</td>
<td>59</td>
<td>M</td>
<td>6'0&quot;</td>
<td>R</td>
<td>C5-6</td>
<td>D</td>
<td>58</td>
<td>0.4</td>
<td>Cane on left</td>
</tr>
<tr>
<td>I6</td>
<td>49</td>
<td>F</td>
<td>5'10&quot;</td>
<td>L</td>
<td>C7</td>
<td>D</td>
<td>23</td>
<td>0.7</td>
<td>NA</td>
</tr>
<tr>
<td>I7</td>
<td>40</td>
<td>F</td>
<td>5'8&quot;</td>
<td>L</td>
<td>C2-T1</td>
<td>D</td>
<td>253</td>
<td>0.5</td>
<td>NA</td>
</tr>
<tr>
<td>I8</td>
<td>59</td>
<td>F</td>
<td>5&quot;</td>
<td>R</td>
<td>C6</td>
<td>D</td>
<td>180</td>
<td>0.3</td>
<td>NA</td>
</tr>
<tr>
<td>I9</td>
<td>25</td>
<td>M</td>
<td>5'11&quot;</td>
<td>R</td>
<td>T4-5</td>
<td>D</td>
<td>90</td>
<td>0.4</td>
<td>Solid AFO on right</td>
</tr>
<tr>
<td>I10</td>
<td>57</td>
<td>M</td>
<td>6'2&quot;</td>
<td>L</td>
<td>C5</td>
<td>D</td>
<td>122</td>
<td>0.3</td>
<td>Cane on right</td>
</tr>
</tbody>
</table>

ASIA: American Spinal Injury Association

Table 7-2. Ipsilateral H/M ratio with and without the AFO.

<table>
<thead>
<tr>
<th>Subject</th>
<th>No AFO</th>
<th>AFO</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Standing</td>
<td>Mid-stance</td>
</tr>
<tr>
<td>ID</td>
<td></td>
<td></td>
</tr>
<tr>
<td>I2</td>
<td>0.71</td>
<td>0.59</td>
</tr>
<tr>
<td>I3</td>
<td>0.35</td>
<td>0.20</td>
</tr>
<tr>
<td>I4</td>
<td>0.69</td>
<td>0.53</td>
</tr>
<tr>
<td>I5</td>
<td>0.68</td>
<td>0.51</td>
</tr>
<tr>
<td>I6</td>
<td>0.05</td>
<td>0.04</td>
</tr>
<tr>
<td>I7</td>
<td>0.08</td>
<td>0.10</td>
</tr>
<tr>
<td>I8</td>
<td>0.79</td>
<td>0.66</td>
</tr>
<tr>
<td>I9</td>
<td>0.70</td>
<td>0.58</td>
</tr>
<tr>
<td>I10</td>
<td>0.36</td>
<td>0.48</td>
</tr>
<tr>
<td>Avg</td>
<td>0.49</td>
<td>0.41</td>
</tr>
<tr>
<td>Std.Dev</td>
<td>0.29</td>
<td>0.23</td>
</tr>
</tbody>
</table>
Table 7-3. Contralateral H/M ratio with and without the AFO.

<table>
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<th>Subject ID</th>
<th>No AFO</th>
<th>AFO</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Standing</td>
<td>Mid-stance</td>
</tr>
<tr>
<td>I2</td>
<td>0.31</td>
<td>0.45</td>
</tr>
<tr>
<td>I3</td>
<td>0.40</td>
<td>0.45</td>
</tr>
<tr>
<td>I4</td>
<td>0.51</td>
<td>0.66</td>
</tr>
<tr>
<td>I5</td>
<td>0.22</td>
<td>0.19</td>
</tr>
<tr>
<td>I6</td>
<td>0.44</td>
<td>0.19</td>
</tr>
<tr>
<td>I7</td>
<td>0.06</td>
<td>0.10</td>
</tr>
<tr>
<td>I8</td>
<td>0.36</td>
<td>0.13</td>
</tr>
<tr>
<td>I9</td>
<td>0.70</td>
<td>0.71</td>
</tr>
<tr>
<td>Avg</td>
<td>0.38</td>
<td>0.36</td>
</tr>
<tr>
<td>Std.dev</td>
<td>0.19</td>
<td>0.24</td>
</tr>
</tbody>
</table>

Table 7-4. Normalized soleus EMG amplitude with and without AFO ipsilaterally.

<table>
<thead>
<tr>
<th>Subject ID</th>
<th>Standing without AFO</th>
<th>Standing with AFO</th>
<th>Midstance without AFO</th>
<th>Midstance with AFO</th>
<th>Midswing without AFO</th>
<th>Midswing with AFO</th>
</tr>
</thead>
<tbody>
<tr>
<td>I2</td>
<td>0.12</td>
<td>0.16</td>
<td>0.22</td>
<td>0.13</td>
<td>0.23</td>
<td>0.16</td>
</tr>
<tr>
<td>I3</td>
<td>0.04</td>
<td>0.05</td>
<td>0.05</td>
<td>0.05</td>
<td>0.05</td>
<td>0.05</td>
</tr>
<tr>
<td>I4</td>
<td>0.14</td>
<td>0.13</td>
<td>0.15</td>
<td>0.20</td>
<td>0.16</td>
<td>0.17</td>
</tr>
<tr>
<td>I5</td>
<td>0.13</td>
<td>0.12</td>
<td>0.16</td>
<td>0.14</td>
<td>0.14</td>
<td>0.14</td>
</tr>
<tr>
<td>I6</td>
<td>1.60</td>
<td>1.65</td>
<td>0.28</td>
<td>0.31</td>
<td>1.25</td>
<td>1.22</td>
</tr>
<tr>
<td>I7</td>
<td>4.68</td>
<td>4.73</td>
<td>4.68</td>
<td>4.73</td>
<td>4.69</td>
<td>4.73</td>
</tr>
<tr>
<td>I8</td>
<td>2.17</td>
<td>2.17</td>
<td>2.26</td>
<td>2.19</td>
<td>2.18</td>
<td>2.17</td>
</tr>
<tr>
<td>I9</td>
<td>2.64</td>
<td>2.66</td>
<td>2.65</td>
<td>2.66</td>
<td>2.66</td>
<td>2.67</td>
</tr>
<tr>
<td>I10</td>
<td>NT</td>
<td>NT</td>
<td>NT</td>
<td>NT</td>
<td>NT</td>
<td>NT</td>
</tr>
</tbody>
</table>

Table 7-5. Normalized TA EMG amplitude with and without AFO ipsilaterally.

<table>
<thead>
<tr>
<th>Subject ID</th>
<th>Standing without AFO</th>
<th>Standing with AFO</th>
<th>Midstance without AFO</th>
<th>Midstance with AFO</th>
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<th>Midswing with AFO</th>
</tr>
</thead>
<tbody>
<tr>
<td>I2</td>
<td>0.14</td>
<td>0.12</td>
<td>0.14</td>
<td>0.13</td>
<td>0.21</td>
<td>0.17</td>
</tr>
<tr>
<td>I3</td>
<td>0.10</td>
<td>0.12</td>
<td>0.11</td>
<td>0.10</td>
<td>0.12</td>
<td>0.10</td>
</tr>
<tr>
<td>I4</td>
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<td>0.96</td>
<td>0.96</td>
<td>0.95</td>
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<td>0.96</td>
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<tr>
<td>I5</td>
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<td>0.40</td>
<td>0.38</td>
<td>0.41</td>
<td>0.42</td>
<td>0.44</td>
</tr>
<tr>
<td>I6</td>
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<td>0.03</td>
<td>0.02</td>
<td>0.02</td>
</tr>
<tr>
<td>I7</td>
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<td>0.57</td>
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<td>0.55</td>
</tr>
<tr>
<td>I8</td>
<td>0.46</td>
<td>0.43</td>
<td>0.42</td>
<td>0.44</td>
<td>0.46</td>
<td>0.44</td>
</tr>
<tr>
<td>I9</td>
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<td>0.89</td>
<td>0.82</td>
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</tr>
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<td>NT</td>
<td>NT</td>
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</tr>
</tbody>
</table>
Table 7-6. Normalized soleus EMG amplitude with and without AFO contralaterally.

<table>
<thead>
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<th>Subject ID</th>
<th>Standing</th>
<th>Midstance</th>
<th>Midswing</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>without AFO</td>
<td>with AFO</td>
<td>without AFO</td>
</tr>
<tr>
<td>I2</td>
<td>0.14</td>
<td>0.16</td>
<td>0.16</td>
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<tr>
<td>I3</td>
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<td>0.09</td>
<td>0.09</td>
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<tr>
<td>I4</td>
<td>0.12</td>
<td>0.08</td>
<td>0.19</td>
</tr>
<tr>
<td>I5</td>
<td>0.12</td>
<td>0.13</td>
<td>0.15</td>
</tr>
<tr>
<td>I6</td>
<td>0.10</td>
<td>0.12</td>
<td>0.11</td>
</tr>
<tr>
<td>I7</td>
<td>3.33</td>
<td>3.33</td>
<td>3.33</td>
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<td>I8</td>
<td>1.60</td>
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<tr>
<td>I9</td>
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<td>I10</td>
<td>NT</td>
<td>NT</td>
<td>NT</td>
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</tbody>
</table>

Table 7-7. Normalized TA EMG amplitude with and without AFO contralaterally.

<table>
<thead>
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<th>Subject ID</th>
<th>Standing</th>
<th>Midstance</th>
<th>Midswing</th>
</tr>
</thead>
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<td>without AFO</td>
<td>with AFO</td>
<td>without AFO</td>
</tr>
<tr>
<td>I2</td>
<td>0.15</td>
<td>0.17</td>
<td>0.18</td>
</tr>
<tr>
<td>I3</td>
<td>0.48</td>
<td>0.45</td>
<td>0.43</td>
</tr>
<tr>
<td>I4</td>
<td>0.47</td>
<td>0.46</td>
<td>0.47</td>
</tr>
<tr>
<td>I5</td>
<td>0.57</td>
<td>0.57</td>
<td>0.56</td>
</tr>
<tr>
<td>I6</td>
<td>0.44</td>
<td>0.44</td>
<td>0.43</td>
</tr>
<tr>
<td>I7</td>
<td>1.25</td>
<td>1.24</td>
<td>1.22</td>
</tr>
<tr>
<td>I8</td>
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<td>0.67</td>
</tr>
<tr>
<td>I10</td>
<td>NT</td>
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<td>NT</td>
</tr>
</tbody>
</table>


60. Standards for neurological classification of spinal injury patients. (Chicago, IL, 1984).


77. Ramon y Cajal, S. *Degeneration and regeneration of the nervous system.* (Oxford University Press, London, 1928).


215. Barker, S. P. in *School of Biomedical Engineering, Science and Health Systems* (Drexel University, 2004).


BIOGRAPHICAL SKETCH

Preeti Nair was born in Mumbai, India in 1979. She received her bachelor’s degree in physical therapy from Pune University, India, in 2001. She worked for a year at a government hospital in Mumbai and for the non-profit Multiple Sclerosis Society of India. Her specific interest in neurological rehabilitation developed from two areas of interest; first, the elusive workings of the nervous system for the control of movement and learning. Secondly, her work experience with individuals with neurological impairment and the challenges faced in restoring them to their activities of daily living. The process of disablement unleashed by the disease state and perpetuated by underdeveloped infrastructure and apathy for neurological rehabilitation motivated her to pursue higher education in the United States; a country that has set the mark for its multidimensional approach towards enablement and empowerment of an individual with impairment. Duly, she chose the interdisciplinary, Rehabilitation Sciences Doctoral program offered at the University of Florida in 2002. Under the expert tutelage of Dr. Andrea Behrman and Dr. Steven Kautz, her research deals with examining the neuromechanical control of walking with orthotic devices in individuals after spinal cord injury that integrates the principles of neurological control of walking, motor control and movement mechanics. She is a recipient of the Alumni Fellowship which provided financial support for her doctoral education.