IMPACT OF WEARING A ROBOTIC EXOSKELETON ON GAIT PARAMETERS AND MUSCLE ACTIVITY IN INDIVIDUALS WITH INCOMPLETE SPINAL CORD INJURY

A DISSERTATION

SUBMIITED IN PARTIAL FULFILLMENT OF THE REQUIREMENTS

FOR THE DEGREE OF DOCTOR OF PHILOSOPHY

IN THE GRADUATE SCHOOL OF THE

TEXAS WOMAN'S UNIVERSITY

SCHOOL OF PHYSICAL THERAPY

COLLEGE OF HEALTH AND SCIENCES

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DENTON, TEXAS

MAY 2019

DEDICATION

I dedicate my dissertation to my parents, Mujawwil and Fadah, for making me who I am, and for always motivating and supporting me.

My special gratitude to my wife, Ahood, for her patience, love, and understanding the ups-and-downs during this journey.

To my children, Mohammed, Albara, and Jwana, who have made me stronger and more fulfilled than I could ever been.

To all of my family, without your love and support this work would not have been possible.

ACKNOWLEDGEMENTS

I would like to acknowledge everyone who contributed to my dissertation. To Dr. Chad Swank, as my former dissertation committee chair for most of my dissertation period, thank you for your kindness, generosity, and understanding. To Dr. Sharon Wang-Price, as my current dissertation committee chair, thank you for your support from the beginning of this dissertation. Both of you always stood beside me, when I needed guidance. To Dr. Fan Gao, Dr. Ann Medley, and Dr. Hui-Ting Goh as my dissertation committee members, thank you for your support, guidance, and valuable feedback on my dissertation.

I would also like to thank all the Ph.D. students at Texas Woman's University, School of Physical Therapy-Dallas, who helped me with data collection and participant recruitment for my dissertation study.

Special thanks to my country, Saudi Arabia, for their financial support. Without you, I would not be able to complete my Ph.D. studies in the United States.

Special thanks to the United States for giving me opportunity to pursue my doctoral education.

ABSTRACT

SATTAM ALMUTAIRI

IMPACT OF WEARING A ROBOTIC EXOSKELETON ON GAIT PARAMETERS AND MUSCLE ACTIVITY IN INDIVIDUALS WITH INCOMPLETE SPINAL CORD INJURY

MAY 2019

BACKGROUND: Robotic exoskeleton devices, such as the Ekso GT™ robotic exoskeleton (EKSO), enable individuals with lower extremity weakness to stand up and walk over ground. Research relevant to the effects of the EKSO on gait parameters and muscle activity in patients with incomplete spinal cord injury (SCI) is limited. Therefore, the purpose of this study was to evaluate whether people with incomplete SCI would walk differently when they wore the EKSO. Specifically, the temporospatial gait parameters as well as kinematics and muscle activity of the lower extremities during level walking were compared between two conditions: with and without wearing the EKSO.

SUBJECTS: Ten ambulatory adults (age: 39.3 ± 11.7 years, 9 men and 1 woman) with incomplete SCI completed the study. Average time since injury was 7 ± 5.5 years with a Walking Index for Spinal Cord Injury II (WISCI II) score of 14.5±2.8.

METHODS: A 10-camera motion analysis system, synchronized with a surface electromyographic (EMG) unit, was used to obtain temporospatial gait parameters, range of motions (ROMs) of hip flexion-extension, knee flexion-extension and ankle dorsiflexion-plantarflexion, and muscle activity of the lower extremities. Each participant performed walking under two conditions: with and without wearing the EKSO.

RESULTS: There were significant differences between the two conditions in gait speed (p = 0.006, no EKSO: 0.56 ± 0.32m/s, EKSO: 0.20 ± 0.03m/s), stride length (p = 0.001, no EKSO: 1.04 ± 0.24m, EKSO: 0.65 ± 0.07m), step length (p = 0.001, no EKSO: 0.51 ± 0.12m, EKSO: 0.33 ± 0.03m) and swing time (p = 0.006, no EKSO: 0.61 ± 0.17s, EKSO: 0.80 ± 0.13), but not in double-limb-support time (p = 0.474, no EKSO: 0.73 ± 0.75s, EKSO: 0.90 ± 0.25s) and stance time (p = 0.413, no EKSO: 2.03 ± 01.66s, EKSO: 2.44 ± 0.42s). When wearing the EKSO, ankle dorsiflexion-plantarflexion ROM was significantly reduced during the stance phase (p = 0.020, no EKSO: 35.90 ± 26.18, EKSO: 11.34 ± 4.55), but there was no difference between conditions in the knee (p = 0.211, no EKSO: 23.92 ± 9.87, EKSO: 27.58 ± 0.4.99) or hip (p = 0.425, no EKSO: 34.14 ± 8.04, EKSO: 32.25 ± 2.64) ROMs. In addition, there were no significant differences between conditions in all three ROMs in the swing phase. Lastly, there were no significant differences in EMG activity of the lower extremities between conditions in both the stance and swing phases.

CONCLUSION: Participants walked slower with shorter stride length when walking with a robotic exoskeleton. Although kinematics were unaffected in hip and knee ROM, ankle dorsiflexion-plantarflexion ROM was decreased primarily due to the rigid design of the ankle joint in the EKSO, which limited ankle motions. The finding of no significant change of muscle activity when wearing the EKSO could be due to the reduced walking speed and the heterogeneity of the participants.

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CHAPTER I

INTRODUCTION

Spinal cord injury (SCI) notably affects thousands of people annually, leading to significant impairments, activity limitations, and participation restrictions. SCI refers to damage to any part of the spinal cord. When this damage is caused by an external force, (e.g., car accident), it is known as "*traumatic SCI*". When the damage is caused by an internal condition (e.g., cancer, stroke), it is known as "*non-traumatic SCI*". This damage to the spinal cord can result in a complete or incomplete loss of the ability of the brain to receive and send signals across the level of the injury. Consequently, an SCI results in impairments and a loss of function below the level of injury.

The estimated annual incidence of SCI in the United States (U.S.) is approximately 17,700 new cases each year. In 2018, the prevalence of SCI in the U.S. was estimated to be 288,000 persons (National Spinal Cord Injury Statistical Center [NSCISC], 2018). A person with SCI may have impairments, which often affect mobility and walking ability, thus significantly affecting the level of disability and overall quality of life (Seeman, Merkin, Crimmins, & Karlamangla, 2010). Therefore, a primary goal of SCI rehabilitation intervention is to promote a maximal level of functional recovery while minimizing physical impairments and functional limitations, and improving quality of life (Harkema et al., 2012).

During rehabilitation, patients with SCI often are instructed to use compensatory strategies to perform functional activities. However, recent studies have shown that new

rehabilitation strategies, such as locomotor training, are able to enhance and facilitate the recovery of locomotor function (Behrman & Harkema, 2007; Harkema, Behrman, & Barbeau, 2011; Harkema et al., 2012). For example, one intervention used for locomotor training of people with incomplete SCI is body-weight-supported treadmill training (BWSTT) (Senthilvelkumar, Magimairaj, Fletcher, Tharion, & George, 2015). Despite the benefits of BWSTT, this intervention places some burden on therapists. BWSTT often requires the effort of multiple physical therapists (generally up to three) to assist the patient with leg movement and controlled trunk movement, particularly for those patients who require substantial walking assistance (Dietz, Nef, & Rymer, 2011).

Due to the high demand of effort needed from therapists to use BWSTT, several robotic assistive gait devices, such as the robotic exoskeleton, have been developed for automating locomotor training of people with SCI. Robotic exoskeletons are designed to mimic human body design and movement. Some robotic exoskeletons are more sophisticated in terms of detecting the brain signals by non-invasive electroencephalographic (EEG) electrodes and translating these signals to movement (King, 2014; Kwak, Müller, & Lee, 2015). Other robotic exoskeletons are preprogrammed to mimic human movement with sensors or a hand control manually to trigger the movement.

Several robotic exoskeletons are commercially available for use in treatment, including Lokomat (Jezernik, Colombo, Keller, Frueh, & Morari, 2003), lower-extremity powered exoskeleton (LOPES) (Veneman et al., 2007), and Ekso GT™ exoskeleton (EKSO) (Contreras-Vidal et al., 2016). In particular, the EKSO enables individuals with

lower extremity weakness to stand up and walk using a walker or forearm crutches. Additionally, the EKSO allows full weight-bearing and reciprocal gait depending on the level of paralysis or hemiparesis.

Several studies have shown that the use of an EKSO is a safe and well-tolerated intervention for locomotor training of people after SCI (Kolakowsky-Hayner, Crew, Moran, & Shah, 2013; Zeilig et al., 2012). They also reported no adverse or safety events, such as serious skin breakdowns, pain and spasticity, fractures and loss of balance during gait training with the EKSO (Kolakowsky-Hayner et al., 2013). Other studies reported treatment effects of the EKSO such as improved seated balance and trunk muscle activation, (Chisholm, Alamro, Williams, & Lam, 2017), as well as metabolic changes, reduced spasticity, and improved bowel function (Miller, Zimmermann, & Herbert, 2016).

In a pilot study (Swank, Wang-Price, Gao, & Almutairi, 2019), the differences in temporospatial gait parameters, kinematics, and muscular activity in healthy adult people with and without using the EKSO was investigated in order to understand mechanisms of gait changes when wearing the EKSO. Participants demonstrated significant differences between conditions (no EKSO and EKSO) in temporospatial gait parameters, and in lower extremity kinematics during both the stance and swing phases, except hip joints in the swing phase. When wearing the EKSO, participants walked slower with shorter steps and greater double-limb- support time as compared to when walking without the EKSO. The results also showed that the participants used less hip flexion-extension and ankle plantarflexion-dorsiflexion range of motion (ROM), but used greater knee flexion-

extension ROM when they walked in the EKSO. However, a different gait pattern change was observed during the swing phase. During the swing phase, the participants had similar hip flexion-extension ROM between the no EKSO and EKSO conditions, but had less knee flexion-extension and ankle plantarflexion-dorsiflexion ROM in the EKSO condition. In addition, the electromyographic (EMG) analysis showed a significant reduction in muscle activity of lower extremities during the stance phase when the participants walked wearing the EKSO. During the swing phase, muscle activity was different only for distal muscle groups (bilateral soleus and tibialis anterior and left medial hamstrings). However, it remains unknown how the use of the EKSO would influence temporospatial gait parameters, kinematics of the lower extremities, and neuromuscular activity of the lower extremities in people with incomplete SCI.

Statement of Problem

Human walking is a repeated sequence of motions of the lower limbs, which move the body forward with a large base of support to maintain the body's balance. This repeated sequence of motions depends on the degree of freedom (DOF) in joint mobility and on muscle activity. Walking ability often is affected in people with SCI. Different interventions of gait rehabilitation are used in rehabilitation centers to improve walking ability and reduce compensatory movements during walking. Manually- assisted over-ground or over-treadmill gait training commonly is used by physical therapists for locomotor training in people with incomplete SCI (Behrman & Harkema, 2007; Kirshblum, 2004). However, these techniques are physically demanding on therapists. Therefore, the desired dosage of gait training may not be achieved simply because

therapists can become fatigued quickly and, as a result, are unable to maneuver the patients.

Alternative modalities emerged in the late 1990s, such as robotics-assisted gait training. Robotics-assisted gait training omits the burden of using manual assistance for over-ground or over-treadmill gait training. In addition, studies of people with incomplete SCI showed that participants improved their walking ability after robotic-assisted gait training (Milia et al., 2016; Shin, Kim, Park, & Kim, 2014). Therefore, robotic exoskeletons have been recommended for gait training to improve mobility for people with a walking disorder, such as SCI. However, Lokomat and LOPES are suspended over the treadmill, and therefore do not accurately mimic the real walking environment. Additionally, the patient is partially unloaded by a suspension system, thus not allowing full weight-bearing on the lower extremities. These limitations often impede optimal gait training effects and possibly delay plastic change in the central nervous system. Conversely, the EKSO enables individuals with lower extremity weakness, such as that caused by SCI, to stand up and walk over level surfaces using a walker or forearm crutches. As mentioned earlier, the EKSO allows a full weight-bearing and reciprocal gait while providing variable assistance depending on the needs of the individual.

Purposes of the Study

This pilot study indicated that there were significant changes between walking with and without wearing the EKSO in healthy adult participants (Swank et al., 2019). However, after people have an episode of neurological insult, their neural system, and its distribution become disrupted, resulting in altered walking characteristics (Stein, Yang,

Belanger, & Pearson, 1993). Therefore, people with SCI may respond differently than healthy adults while walking in the EKSO. Before investigating the effects of long-term EKSO training on gait, a study investigating immediate changes of gait when patients with SCI wear the EKSO is warranted. Therefore, the purpose of this study was to evaluate whether people with incomplete SCI would walk differently when they wear the EKSO. Specifically, the temporospatial gait parameters as well as kinematics and muscle activity of lower extremities during level walking were compared between the two conditions: with and without wearing the EKSO. The temporospatial gait parameters included walking speed, stride length, double limb support, step length, stance time, and swing time. Because the EKSO only allows motions in the sagittal plane, kinematics of hip flexion-extension, knee flexion-extension, and ankle dorsiflexion-plantarflexion were examined between the two conditions.

Research Questions

The following research questions were investigated:

1. When people with incomplete SCI walk while wearing the EKSO, would the temporospatial gait parameters be different as compared to walking without wearing the EKSO?
2. When people with incomplete SCI walk while wearing the EKSO, would the kinematics of the lower extremities be different as compared to walking without wearing the EKSO?

3. When people with incomplete SCI walk while wearing the EKSO, would the neuromuscular activity of their lower extremities be different as compared to walking without wearing the EKSO?

Hypotheses

Null Hypotheses

The following were the null hypotheses:

1. There would be no significant differences in the temporospatial gait parameters in people with incomplete SCI between walking while wearing the EKSO and without wearing the EKSO.
2. There would be no significant differences in the kinematics of the lower extremities in people with incomplete SCI between walking while wearing the EKSO and without wearing the EKSO.
3. There would be no significant differences in the neuromuscular activity of the lower extremities in people with incomplete SCI between walking while wearing the EKSO and without wearing the EKSO.

Research hypotheses

To investigate the effect of the EKSO on people with incomplete SCI, the following research hypotheses were stated:

1. There would be significant differences in the temporospatial gait parameters in people with incomplete SCI between walking while wearing the EKSO and without wearing the EKSO.

2. There would be significant differences in the kinematics of the lower extremities in people with incomplete SCI between walking while wearing the EKSO and without wearing the EKSO.
3. There would be significant differences in the neuromuscular activity of the lower extremities in people with incomplete SCI between walking while wearing the EKSO and without wearing the EKSO.

Operational Definitions

The following terms were defined for the purpose of this study:

1. *Locomotor training* is a therapeutic intervention designed to promote the recovery of postural control, balance, standing, walking, health, and quality of life. The goal is to retrain the neuromuscular system below the level of the lesion to restore its maximum function (Harkema et al., 2012).
2. *Robotic exoskeleton* is a battery powered, motor driven, robotic device with a pair of legs. It generates active motion at the hip and knee joints in a properly sequenced manner (Karelis, Carvalho, Castillo, Gagnon, & Aubertin-Leheudre, 2017).
3. *Temporospatial gait parameters* refer to the variables related to time and space during walking (Sale et al., 2016).
4. *Step length* is the anterior-posterior distance from the heel of one foot in contact with the ground to the heel of the opposite foot in contact with the ground (Sale et al., 2016).

5. *Double limb support* is a phase of the gait cycle during which both feet are in contact with the ground simultaneously (Sale et al., 2016).
6. *Walk speed* is measured by dividing the distance walked by the ambulation time (Sale et al., 2016).
7. *Kinematics* describes the motions of a body or a segment of the body (e.g., lower limb), regardless of the force, which causes that motion (Kadaba, Ramakrishnan, & Wootten, 1990).
8. *The neuromuscular system* is a combination of central nervous system links to a neural circuit, including motor neurons in the spinal cord, sensory neurons in the dorsal root ganglion, and skeletal muscle fibers throughout the peripheral nervous system in order to initiate the body movement and postural control (Ko, 2001).
9. *Walking Index for Spinal Cord Injury II (WISCI II)* is a gait performance scale using an ordinal score which is assigned based on the type of assistive devices and/or braces that used and the amount of the physical assistance needed during walking (Ditunno et al., 2000; Ditunno & Ditunno, 2009).

Assumptions

The assumptions for the study were as follows:

1. The participants with SCI enrolled in this study have been diagnosed appropriately with incomplete SCI and are representative of the population of people with incomplete SCI.
2. Participants would maintain their normal walking pattern and speed during the evaluation.

3. The electrode placements were applied following the recommendations of Surface ElectroMyoGraphy for the Non-Invasive Assessment of Muscles (SENIAM) (Hermens, Freriks, Disselhorst-Klug, & Rau, 2000), thus allowing the recording of the optimal muscle activity of the target muscle.

Limitations

The limitations for this study were as follows:

1. A sample of convenience may not be representative of the population of interest, thus limiting the generalizability of the study.
2. A small sample size may limit the generalizability of the study.
3. Participants were novice EKSO users, and could have different learning ability of EKSO walking, thus affecting the results of the dissertation study.
4. The EMG and kinematic data could be affected because the EMG electrodes and marker locations were adjusted for the EKSO condition.
5. Only the kinematic data in the sagittal plane was examined in this dissertation study. Although kinematics in both the frontal and transverse planes were restricted by the EKSO, there still could be kinematics in these two planes.

Significance of Study

Persons with incomplete SCI can have challenges when walking (Stein et al., 1993). One locomotor retraining intervention approach is the use of robotic exoskeletons. Previous studies reported that the EKSO is safe and tolerated by people with incomplete SCI (Kolakowsky-Hayner et al., 2013). Other studies evaluated the treatment effects of the EKSO on seated balance and trunk muscle activation (Chisholm et al., 2017), as well

as metabolic changes, spasticity level, and bowel function (Miller et al., 2016). However, recent systematic reviews show insufficient evidence to determine the benefit of the EKSO due to limited research in this area (Contreras-Vidal et al., 2016; Esquenazi, Talaty, & Jayaraman, 2017).

To date, the effects of the use of EKSO on variables such as temporospatial gait parameters, kinematics, and neuromuscular activity have not been examined in people with incomplete SCI. Appropriate kinematics and muscle activation are essential for walking function. Therefore, the results of this study could provide evidence for the use of the EKSO for gait training with regard to temporospatial gait parameters, kinematics, and muscle activity for people with incomplete SCI. The findings also could shed light on how people with incomplete SCI perceive gait training with an exoskeleton, and how robotic-assisted gait training affects their quality of intervention.

CHAPTER II

LITERATURE REVIEW

The purpose of this chapter is to review the literature on the following topics: (1) prevalence and incidence of SCI, (2) classifications of SCI, (3) gait abnormalities in SCI, (4) assessment of gait, (5) neurorehabilitation interventions for SCI, and (6) gait training in SCI.

Prevalence and Incidence of SCI

In 2018, the National Spinal Cord Injury Statistical Center (NSCISC) estimated the prevalence of SCI in the U.S. to be approximately 288,000 persons with an annual incidence of SCI to be approximately 54 cases per million or approximately 17,700 new SCI cases each year (NSCISC, 2018). The life expectancy for people with SCI has increased over the past decade due to advances in medical knowledge and technology, although it remains lower than that of the general healthy population. Particularly, people with incomplete SCI have experienced an increase in life expectancy since 1996. However, modest improvements were observed in complete SCI (Jha, Charlifue, & Lammertse, 2010). During World War I, 90 % of people with SCI died within one year of the injury, and only 1% survived more than 20 years (Swan & Grundy, 2002). The estimated cost of living for a person with SCI throughout his or her lifetime varies, depending on the severity of the injury and disability. The cost is highest during the first year after the injury and then decreases the following years. In 2017, the average annual

cost is $949,493 for the first year of tetraplegia, $537,271 for the first year of paraplegia, and $359,783 at any other levels of SCI (NSCISC, 2018).

In the US, males account for approximately 80% of new SCI cases. Most people with SCI are young adults. The average age at injury was 29 years during the 1970s, but currently the average age at the injury is 42 years. Children represent 5% of people with SCI primarily due to traffic accidents or falls (Swan & Grundy, 2002). About 22% of injuries occurred among non-Hispanic blacks, which is higher than the proportion of non-Hispanic blacks in the general population (12%). More than half (51.4%) of people with SCI are single at the time of injury. The percentage of people with SCI who are married at the time of the injury is 32.8%. However, 9.5% are divorced at the time of injury. The majority of people with SCI (58.1%) are employed and 15.1% of those are students (NSCISC, 2018).

The common causes of SCI include motor vehicle accidents (38%), violence (14%), falls (30%), recreational activities (9%), and medical surgery (5%). Currently, incomplete tetraplegia is the most frequent neurological category, representing 45%, incomplete paraplegia 21%, complete paraplegia 20%, and complete tetraplegia 14% (NSCISC, 2018).

Classifications of SCI

Injury to any part of the spinal cord results in SCI and leads to either complete or incomplete loss of motor, sensory and autonomic body function below the level of the injury depending on the severity and the location of the injury. SCI can be grouped into two general categories: traumatic SCI (TSCI) and non-traumatic SCI (NTSCI). TSCI is a

result from, but not always, external forces such as car accident, fall, gunshot wounds, and stabbing. In addition, TSCI may be caused by surgical procedures or radiation exposure. NTSCI includes cord compressions (e.g., spinal stenosis and spondylolisthesis), circulatory problems (e.g., spinal stroke), demyelinating diseases (e.g., multiple sclerosis and primary lateral sclerosis), inflammation diseases (e.g., spinal meningitis and acquired immunodeficiency induced neuropathy), and abnormal cells growth (e.g., cancer) ("WHO. Spinal Cord Injury," 2019). Demographic characteristics showed clear differences between TSCI and NTSCI. People with an NTSCI are generally older than those with a TSCI and less likely to be male. In addition, the NTSCI group has a shorter length of stay (LOS) in the hospital and is less disabled on admission (New, Simmonds, & Stevermuer, 2011). Other studies reported that people with a TSCI demonstrate a significantly greater functional level, determined by the Functional Independence Measure (FIM), at discharge as compared to those with an NTSCI (McKinley, Seel, & Hardman, 1999; McKinley, Seel, Gadi, & Tewksbury, 2001). This finding could be a result of the TSCI group's longer LOS for more rehabilitation training, thus achieving a higher function level.

The examination of the dermatomes and myotomes is useful for determining the level the sensory and motor functions and the extent of injury severity of a person with SCI. American Spinal Injury Association (ASIA) and the International Medical Society of Paraplegia (IMSOP) published the International Standards Scale for Neurological and Functional Classification of Spinal Cord Injury (see Appendix A) in September 1992. It has become the most common classification scale used by clinicians to determine the

level of injury in people with SCI. There are four components in this scale: (1) sensory and motor levels, (2) the completeness of the injury, (3) the ASIA Impairment Scale (AIS), and (4) the zone of partial preservation for complete injuries. The AIS is highly recommended for clinical assessment of people with acute and chronic SCI (Ditunno, Young, Donovan, & Creasey, 1994). The inter-rater reliability of the AIS scores has been shown to be excellent for the AIS total motor scores (ICC = 0.999), AIS pin prick scores (ICC = 0.997), and AIS light touch scores (ICC = 0.988) (Savic, Bergström, Frankel, Jamous, & Jones, 2007). Additionally, an AIS score has been shown to have a significant close correlation ($p < 0.001$) with somatosensory evoked potentials (SSEP) in an electrophysiological study (Curt & Dietz, 1997). A previous study also showed excellent predictive validity of AIS upper extremity motor score (UEMS) ($r = 0.81$), lower extremity motor score (LEMS) ($r = 0.79$), and adequate predictive sensory scores for pin prick and light touch ($r = 0.64$, $r = 0.55$), respectively (Curt, Keck, & Dietz, 1998).

The AIS is a very useful tool; however, it has some limitations. The AIS cannot be used for an unconscious (e.g., due to trauma), uncooperative (e.g., due to drugs or psychiatric disorders) or incomprehensive (e.g., due to language differences or children) person. Therefore, when the AIS cannot be administered due to above-mentioned conditions, electrophysiological examinations are recommended for determining the level of SCI (e.g., SSEP) (Curt & Dietz, 1997). It is important for clinicians who work with people with SCI to know the advantages and the limitations of AIS. Additionally to that, clinicians have to have insight into the most common spinal cord syndromes to help them for decision-making.

Spinal cord lesions often are classified into five categories: 1) transverse cord lesion, 2) hemicord lesion, 3) central cord lesion, 4) posterior cord lesion, and 5) anterior cord lesion (Blumenfeld, 2002). A transverse cord lesion leads to incomplete or complete interruption of all sensory and motor tracts below the level of the lesion. This lesion type results in weakness, reflex loss and diminished sensation in all dermatomes. The most common causes of this syndrome are trauma, tumors, multiple sclerosis, and transverse myelitis. Hemicord lesion (a.k.a. Brown-Séquard Syndrome) is commonly caused from stabbing and penetrating injuries, this syndrome results in damage to the lateral corticospinal tract, causing ipsilateral upper motor neuron-type weakness, and damage to the posterior columns of the spinal cord leads to ipsilateral loss of vibration and joint position sense. Additionally, damage to anterolateral tracts causes contralateral loss of pain and temperature sensation. When the posterior horn cells are damaged, there can be an ipsilateral loss of pain and temperature sensation.

Central cord damage can occur when there are small lesions in spinothalamic fibers, which result in bilateral suspended sensation loss to pain and temperature. When there are large lesions in the anterior horn cells, lower motor neuron deficits occur at the level of the lesion. In addition, corticospinal tracts could be damaged producing upper motor neuron lesion. Common causes of this syndrome include spinal cord contusion, which often occurs with hyperextension injuries of the cervical spine. Another cause of central cord syndrome is intrinsic spinal cord tumors such as hemangioblastoma, ependymoma, or astrocytoma. Posterior cord syndrome occurs when there is damage to the posterior columns of the spinal cord, resulting in loss of vibration and position senses

below the level of the lesion. With a large region of damage, lateral corticospinal tracts may be involved, causing upper motor neuron-type weakness. Common causes of this syndrome are traumatic hyperextension injuries. Anterior cord syndrome is resulting in damage to the anterolateral tracts, which causes loss of pain and temperature sensation below the level of the lesion. The most common cause of anterior cord syndrome is flexion-rotation force to the spine. In addition, damage to the anterior horn cells may result in lower motor neuron weakness at the level of the lesion and below. In larger lesions, the lateral corticospinal tracts may be involved, causing upper motor neuron lesion.

Gait Abnormalities in SCI

People with SCI may experience impairments of voluntary sensory and motor function below the level of the injury. These impairments could lead to a decrease in walking ability, which is the major concern after an SCI, as inability to walk could affect other body functions such as metabolism, bone density, cardiovascular system, muscle strength, and general fitness (Widerström-Noga, Felipe-Cuervo, Broton, Duncan, & Yezierski, 1999).

Walking patterns of people with incomplete SCI differ from those of healthy individuals (Krawetz & Nance, 1996). The quality of walking depends on performing walking tasks in order and on optimal walking patterns. In addition, the quality of walking varies depending on the severity of the injury, the level of the injury, and severity of spasticity (Krawetz & Nance, 1996). People with incomplete thoracic SCI have the most serious muscle hypertonia and hyperreflexia. Consequently, they have the

greatest deviations in kinematic gait parameters as compared to normal adults, particularly in the cadence, double limb support, velocity, knee excursion, and knee angular velocity (Krawetz & Nance, 1996). In contrast, people with incomplete cervical SCI empirically show less deviation from normal gait parameters because they often have less spasticity, and more neurological intact to ambulate due to inability to use crutches or walker because the weakness in the upper extremities (Krawetz & Nance, 1996). Generally, people with SCI with preserved motor function, normal walking speed has been observed. However, in the more severe cases, the person often walks very slowly or cannot walk at all. People with complete SCI ASIA A are not likely to regain mobility due to loss of lower limb muscle strength. Conversely, the majority of people with incomplete SCI are more likely to regain some of mobility functions (Wirz et al., 2005).

It has been reported that people with incomplete SCI after recovery have an average walking speed of 0.80 m/s as compared to normal walking speed (1.36 m/s) (i.e., slower than normal) (van Hedel, Dietz, & Curt, 2007). As people with incomplete SCI walk slower, the duration of their gait cycle (GC) increases and their stride length decreases (Pepin & Barbeau, 1992). Although researchers advocate controlling walking speed to have a meaningful comparison between people with SCI participants and normal healthy participants (Barbeau, Pépin, Norman, Ladouceur, & Leroux, 1998), it is difficult or even impossible to control walking speed in people with incomplete SCI (Whittle, 2007).

In addition, people with SCI demonstrate changes of angular displacements of the lower limb joints in the sagittal plane as compared to that of normal healthy participants.

Barbeau et al. (1998) reported that the hip joint excursion was greater in people with SCI than in normal healthy participants when they walked at the same (0.3 m/s) speed. In another walking study, patients with SCI were found to have a greater knee excursion than a control group (Krawetz & Nance, 1996). Increased joint excursion may result from the compensation for limited motion at another joint. Additionally, Barbeau et al. (1998) reported that people with SCI made foot contact with the knee in a flexed position, and for some cases, the knee remained flexed throughout the stance phase. The ankle joints in some people with SCI were in a more dorsiflexed position than normal healthy controls at the beginning of the stance phase, and remained dorsiflexed even at push-off (Barbeau et al., 1998).

A clinical predication rule has been established by European Multicenter Study on Human Spinal Cord Injury and used to predict which persons with SCI can ambulate independently (van Middendorp et al., 2011). Three predictors were identified for distinguishing between independent home ambulator and non-independent ambulator, including age, light touch sensory test scores at L3 and S1, and motor test scores of the quadriceps and gastrocnemius muscles (van Middendorp et al., 2011). In addition, Kim, Eng, & Whittaker (2004) conducted a study, consisting of 22 incomplete SCI (8 with T2 – T7 SCI, 14 with C5 – C7 SCI), to examine the relationship between the lower extremity muscle strength and walking function. Manual muscle testing was used to score lower extremity muscle strength using a scale from 0 to 5, in half-point increments, with 0 meaning absence of muscle contraction and 5 meaning normal muscle strength. The finding of this study showed strong relationships between walking performance and

lower extremity muscle strength scores. In particular, strength of hip flexors, hip extensors, and hip abductors were significantly associated with gait speed and six-minute walk distance. The hip flexors were also found to play an important role in advancing the swinging limb forward, whereas the hip extensors and abductors provided stability and absorbed loading during the stance phase (Perry, 2010).

Assessment of Gait

Regaining the ability to walk is a priority for most people after an SCI (Ditunno, Patrick, Stineman, & Ditunno, 2008). As mobility is limited in people after SCI, patients' return to walking is the primary goal for rehabilitation, physical therapists should be familiar with the normal gait patterns in order to adequately retrain gait in the people with SCI (Harkema et al., 2011).

In a human being, normal walking is a repeated sequence of motions of the lower limbs that moves the body along a desired path with a base of support to maintain the body's balance. These repeated sequences of motions depend on free joint mobility and muscle activation, which produces forces to advance the limbs and the body forward. During gait, muscles are selectively activated across multiple joints based on the tasks needed to perform walking. Each muscle is activated by a motor control system at a particular time and intensity to control the limb path. Additionally, the sensory system for proprioception and kinesthesia continuously monitors the tendons, muscles, and ligaments at each segment to sense the motion and its surrounding factors that may affect walking (Perry, 2010). However, lacking proprioception often results in walking disorders, even though the joints are intact (Lajoie et al., 1996).

Walking is a pattern of motions which have been learned and stored in the brain and can be performed subconsciously. Normal gait can break down to repetitive cycling process of the stance and swing phases for each leg alternately. A GC starts when the heel contacts ground (i.e., initial contact) and ends at the next initial contact of the ipsilateral limb. However, this may not be applicable for patients who have gait pathological problems. The term of initial contact will be used to designate the onset of the GC.

The GC often is divided into two phases to study human walking: stance phase and swing phase. When walking at a normal speed 82 m/min (1.36 m/s), each leg spends approximately 60% of the time in the stance phase and 40% in the swing phase for each GC (Murray, Drought, & Kory, 1964). In normal walking, double-limb-support accounts for 10% of each GC for one limb. Double limb support occurs twice during one GC, one for each limb, first, during heel strike of ipsilateral limb while the contralateral limb at toe off and the second while the ipsilateral limb at toe off and the contralateral limb at heel strike. The stride length is the distance between two heel contacts of the same foot and sometimes this term used interchangeably with the term GC (Sheffler & Chae, 2015). The durations of the stance and the swing phases are changeable depending on the gait speed. As the gait speed becomes slower, the times for the stance and the swing phases become greater. Likewise, as the gait speed becomes faster, the times for the stance and the swing phases become shorter (Perry, 2010). In this literature review and for this dissertation study, the gait terminology and description as defined by Perry (2010) were

used to identify major events of a GC with a common pathological gait associated with people with SCI.

To accurately and reliably measure and analyze GCs, complex hardware and software programs often are used. However, there are many simple techniques, which are useful for quantifying gait parameters. These techniques often are used in clinical practice due to cost, space, and time. For example, a stopwatch can be used with a pre-determined distance to calculate the walking speed. Generally speaking, higher quality measurement systems cost more, but will perform better in analyzing the gait parameters. Measurements of temporospatial parameters, joint angular motions, and muscle activity during walking are necessary to distinguish between pathological and normal gait, thus helping clinicians in their decision-making. The following paragraphs discuss quantitative methods that are used to measure and analyze GCs.

Kinematic Gait Parameters

Kinematics is defined as the measurement of the joint motion or body segment motion, including linear displacements, velocities, and acceleration variables. Kinematics is not related to the internal or external forces, but rather assesses the orientation of the body segments, and the angles of the joints. A century ago, photography, either a single photograph or a strip of film, was the primary method used to evaluate human movement, including kinematic measurements (Winter, 2009). This method is useful for some purposes, but not for measurement of all kinematic parameters, because it is an un-calibrated system and only measures joint angles in the sagittal plane.

A three-dimensional (3D) motion system, which can capture movements from more than one angle, has been considered a gold standard for performing kinematic and kinetic measurements. Recently, computer-aided motion analysis systems (e.g., VICON Motion System) have been used to capture trajectories of reflective markers placed on key anatomical locations in order to obtain joint angle motions. Data is captured at a series of time intervals known as 'frames'. The time between frames varies depending on the precision of the system used to capture the data (Whittle, 2007). In order to capture precise joint motions, the minimum capture rate should be no less than 60 frames per second (Hz) for activities such as walking. Faster activities (e.g., running) require higher capture rates (e.g., 100 to 1,000 Hz) (Perry, 2010). The capture system links the position of each marker throughout the frames to create continuous trajectories, which represent the paths for each marker during motions (Kapur, Kapur, Virji-Babul, Tzanetakis, & Driessen, 2005).

Some factors may affect the accuracy of capturing 3D joint motions. Human errors of identifying correct landmarks for marker placements lead to systematic errors of determining the joint center. Additionally, errors can come from difficulty in detecting the markers, such as the anterior superior iliac spine (ASIS) markers in an obese person. Additionally, the skin where markers are placed could move during activity, because the skin moves freely over the bony landmarks. Lastly, the standard technique for identifying the neutral (zero) reference is a quiet standing position. However, in a standing position, some joints are not in neutral position but the 3D system will read this as a zero

reference. For example, the neutral position form the ankle joint is at 90°, yet the balance position requires 5° of dorsiflexion (Perry, 2010).

Temporospatial Gait Parameters

Common temporospatial gait parameters in biomechanical studies include cadence, stride length, step length, speed, and double limb support. One way to measure the GC is by using a pressure sensor mat underneath the foot insole within the shoe. As the participant walks with the shoe, force sensors record ground reaction force and the center of the pressure (Whittle, 2007). Clinically, this measurement technique is inexpensive and is easy to use, but it has some limitations. Each person has a different foot size, which makes it difficult to ensure that the sensor is sized correctly for each person. In addition, the pressure sensor measures the pressure, which is different from force.

A portable mat walkway (e.g., GAITRite) is another instrument used to obtain temporospatial gait parameters by tracking footsteps with embedded pressure sensors. The walkway is connected to a personal computer so that electronic recordings of a person's foot steps can be digitalized and stored in the computer. Using custom software, the digital data of the steps then can be used to obtain temporal and spatial parameters of gait. This method of measuring gait parameters has been shown to be a validated and reliable method of analyzing gait patterns (McDonough, Batavia, Chen, Kwon, & Ziai, 2001).

Kinetic Gait Parameters

Kinetics is the study of forces acting on mechanisms (Tao, Liu, Zheng, & Feng, 2012). Common kinetic gait parameters include ground reaction force, joint moments, torque, powers, and work. A force platform is a device, which helps to quantify kinetic gait parameters. Within the platform, there are a number of sensitive transducers used to measure tiny displacements of the upper surface of the platform in all three axes when the force is applied to the platform. Force platforms should be mounted below the floor or walkway level, with the upper surface of the platform flush with the floor or walking surface. To obtain a valid walking trial, many motion analysis systems require the entire foot to land on the platform. In order to achieve contact of the entire foot with the force platform, adjusting the starting position often is required in every walking trial. If it is possible, in order to avoid artificial gait patterns, the force platforms should not be noticeable from the rest of the floor, and therefore, it is preferable to cover the floor, including the force plates, with a continuous flexible material (Whittle, 2007).

Electromyographic Activity during Gait

To further study gait or a functional activity, a 3D motion analysis system often is synchronized with other systems, such as an EMG system for studying muscle activation. Electromyography is the measurement of the electrical activity within a muscle (i.e., muscle action potential), which creates a force that moves the body part. Two EMG methods, surface electromyography and intramuscular electromyography, have been used to study muscle activity. For intramuscular electromyography, fine-wire or needle electrodes are inserted directly into the muscle. This method is uncomfortable and

sometimes painful. Therefore, it is less feasible than surface electromyography for clinical research. Surface EMG electrodes are electrodes placed on the skin over the target muscle with adhesive tape. Some surface EMG electrodes have a small preamplifier to amplify the signal, which comes from the muscle within 25 mm of the skin's surface. Therefore, surface electromyography is appropriate for superficial muscles but not for deep muscles (Whittle, 2007). Another limitation of surface EMG electrodes is the cross-talk effect, in which surface electrodes can detect signals from muscles adjacent to the muscle of interest (Perry, Easterday, & Antonelli, 1981). Perry (2010) recommended that surface EMG electrodes should be limited to studying group muscle action.

For this dissertation study, a 3D motion analysis system synchronized with force plates and a surface EMG system was selected to capture temporospatial gait parameters, kinematics, and muscle activity, because the 3D motion system is consider to be the gold standard for capturing joint motions.

Neurorehabilitation Interventions for SCI

The process of neurorehabilitation is to examine and evaluate a patient who has a disorder and creates a plan of treatment that helps the patient to restore function, decrease pain, and promote movement and independence. Gait and mobility are one of the fundamental components of independent function usually affected by SCI. In people with SCI, recovering gait function is considered an important goal in their lives (Harkema et al., 2011). Consequently, the primary goal of rehabilitation is to achieve the maximum level of function possible for each patient (Ditunno et al., 2008). Neurological

rehabilitation has evolved through different models of the motor control theories and therapeutic interventions for treating people with a neurological movement disorder (e.g., SCI). This section reviews the development of these models and neural regulation of the human movement and how to implement these theories in the clinical practice.

Traditional Models

According to Shumway-Cook and Woollacott (2007), traditional theories of neurorehabilitation include a hierarchical model (see Figure 1) of motor control. In this model, the higher centers of the CNS always control the lower centers. It is a rigid top-down unidirectional process, which considers the higher centers within the CNS, such as the motor cortex, as the commanders for all movement tasks during execution of the movement (Shumway-Cook & Woollacott, 2007).

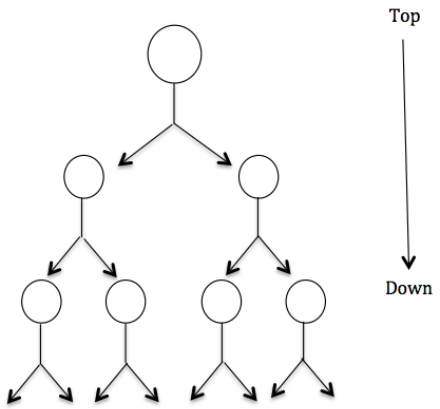


Figure 1. The hierarchical model is the top-down structure, in which the higher centers always control the lower centers (unidirectional communication).

In these theories, the CNS is hard-wired, non-malleable, and incapable of fixing itself (Cajal, 1959). Based on this assumption, therapists select a compensation approach as a rehabilitation strategy for people who have movement disorders due to CNS impairment, and that rehabilitation strategy requires the patient to use compensatory modifications to accomplish a task rather than restoring functions (Behrman, Ardolino, & Harkema, 2017). One example of the compensatory approaches of neurorehabilitation used for people with SCI is to educate patients on how to use a new movement or other parts of his/her body to accomplish the task.

A classic example would be to replace gait training with wheelchair training for mobility. Alternatively, the goal of gait training with parallel bars and a walker for people with SCI who cannot walk is that the patient can load his or her weight through the arms rather than through the legs. Visintin and Barabeau (1994) compared the outcomes of weight bearing on the upper extremities versus weight bearing on the lower extremities with 40% body weight support (BWS). The finding was that the upper extremities weight bearing resulted in decreased muscle activity in the lower limbs and altered limb kinematics (Visintin & Barbeau, 1994). During walking on the parallel bar, a participant compensated for his/her lack of hip, knee, and ankle flexion during the right swing phase by using his arms to push on the parallel bar in order to come up onto his left toes. Therefore, providing an assistive device for gait training can lead to a reliance on compensatory approaches.

Neurofacilitation approaches were developed based on hierarchical and reflex theories, which assume that due to injury, there is a lack of control of the movement

system from the higher-level, thus releasing abnormal reflexes at a lower level (Horak, 1991). Neurofacilitation approaches dramatically change neurorehabilitation interventions and still dominate clinical practice today (Shumway-Cook & Woollacott, 2007). The focuses of the neurofacilitation model on retraining motor control are to facilitate normal movement pattern and to inhibit abnormal reflexes. Using neurofacilitation approaches, therapists promote and inhibit movements by providing proprioceptive feedback with their hands on the patients to guide the movement (Horak, 1991).

Neurofacilitation approaches include the Bobath approach, the Rood approach, Brunnstrom's approach, proprioceptive neuromuscular facilitation (PNF), and the sensory integration approach. These approaches start therapeutic intervention at the lowest level by stimulating reflex responses followed by automatic response and lastly by voluntary fractionalization of isolated movements (Horak, 1991).

Contemporary Theories

Researchers have been working constantly to understand the nature of movement. Therefore, newer models arise with a modification upon understanding the cause and nature of movement. Contemporary theories of motor control explain movement regulation as a distributed process that results from the interaction of multiple factors and human systems working together to generate and control movement (Shumway-Cook & Woollacott, 2007). In other words, movement is controlled by various factors depending on the task, and the higher centers are no longer considered the only ones that control movement.

Recently, new evidence has emerged to determine the role of the spinal cord in controlling movement, particularly in locomotion. There are several studies suggesting that the spinal cord may contain a central pattern generator (CPG), which is a complex basic neural circuit, which can learn and generate walking motions in animals without interaction from the higher centers. In animals, spinal cord transection procedures have been performed to observe locomotion and neuroplasticity. In 1981, Grillner et al. showed that neuronal networks in the spinal cord are able to activate and produce a locomotor rhythm in the absence of afferent input and descending control from the higher centers in the cortex of a cat (Grillner, McClellan, Sigvardt, Wallén, & Wilen, 1981). The study showed that the cat was able to perform basic rhythms and patterns for stepping even after a complete spinal cord transection in response to walking training. Five years later, Lovely et al. (1986) performed a study on cats with complete spinal cord transections. The study was performed on adult cats that were spinalized between T12 and T13. One month later, 14 of 16 cats were able to perform full weight bearing on their hind limbs while stepping on a treadmill. Additionally, they generated a hind-limb forward stepping when they received truncal support and manual assistance from a trainer who held the tail below the level of the hip joints to support lateral stability and lifted each limb during its respective swing phase. Moreover, the cats increased their cadence and step length when the treadmill speed was increased. Similar studies have been implemented on rats, mice, and dogs, which had their spinal cords, transected, and were able to stand and to be induced to walk (Bouyer, 2005; Johnson, Joy, Altevogt, & Liverman, 2005; Leblond, L'Esperance, Orsal, & Rossignol, 2003). These studies support

the idea that the capability of the spinal cord to respond to afferent feedback, whether it is proprioception, muscle length, cutaneous feedback, or all together.

In recent human experiments, the researchers applied epidural electrical to the spinal cord in people with complete SCI (Angeli, Edgerton, Gerasimenko, & Harkema, 2014; Rejc, Angeli, & Harkema, 2015). Stimulating the spinal cord provides a better understanding about the capacity of the spinal cord in generating rhythmic movements and stepping patterns without supraspinal interaction. With implant epidural electrical stimulation in the spinal cord, individuals with SCI were able to stand independently of physical assistance within the first week of stimulation (Rejc et al., 2015). Moreover, voluntary movements of the legs and trunk could be elicited, but only with continuous epidural electrical stimulation (Angeli et al., 2014). It is believed that CPG exists within the lower thoracic and lumbar regions of the human spinal cord (Sheffler & Chae, 2015). Consequentially, this result supports the view of the neural network of spinal cord in that the CPG has the ability to generate basic rhythms and coordinates locomotor movements (Guertin, 2013).

Neuroplasticity

Much of the scientific foundation for the contemporary theories of motor control can be attributed to the understanding of the central nervous system's ability to adapt through a process called neuroplasticity. Neuroplasticity refers to many aspects of the nervous system (e.g., spinal cord) to recognize itself and the ability to form new neural connections by reorganizing its structure, function and connections. In other words, the CNS has the ability to adapt and relearn functions previously lost as a result of cellular

death by trauma or disease. Neuroplasticity can be observed at many levels, from molecular to cellular to systems to behavior. Neuroplasticity can develop during childhood, in response to the environment, in support of learning, in response to disease, or in relation to therapy (Cramer et al., 2011). Adaptive plasticity can occur during nervous system recovery (Cohen et al., 1997), and the degree of recovery varies considerably from person to person, suggesting that there are limitations to adaptive plasticity (Raineteau & Schwab, 2001). Neuroplasticity occurs not only at cortical sites but also at the brainstem and spinal cord sites (van Hedel & Dietz, 2010). The principle of neuroplasticity can occur because cortical, subcortical, and some of the spinal cord circuitry remain intact and partially interconnected. Thus, the ability of recreation of new synaptic connections to supply the area that is partially interconnected in order to restore function is more (Wall, Feinn, Chui, & Cheng, 2015). Therefore, neuroplasticity plays a critical rule in enhancing motor output after SCI through rehabilitation training.

Rehabilitative gait training after incomplete SCI depends on an understanding of theories that attempt to explain the basis of motor control and how the nervous system controls movement. Motor control defined as an area of study of understanding the neural, physical, and behavioral aspects of biological (e.g., human) movements (Schmidt & Lee, 2005). Therapists need to understand how movement is regulated and how neural signals stimulate movements. These theories construct the framework of clinical practice and guide therapists to examine and treat people who have motor control problems and functional movement disorders (Shumway-Cook & Woollacott, 2007). The assumptions behind these theories are constantly evolving with time due to more research and

experiments exploring the area of neurophysiology, biomechanics, motor control, motor learning, motor development, and cognitive training. For the past 30 years, the paradigm of evaluating and treating people with movement disorders, particularly people with SCI, has shifted from hierarchical models of motor control to the new models of motor control.

Gait Training in SCI

The primary goal of rehabilitation following SCI is to promote a maximal level of functional recovery while minimizing physical impairments and functional limitations. Recovery of ambulation in individuals with SCI has been promoted by implementing locomotor training, which is a therapeutic intervention designed to promote the recovery of postural control, balance, standing, walking, health, and quality of life achieved by neuromuscular activation below the level of the lesion after a neurologic injury or disease (Barbeau, 2003; Harkema et al., 2011; Harkema et al., 2012; van Hedel & Dietz, 2010; Wernig, Müller, Nanassy, & Cagol, 1995). Therefore, gait training is critical for recovery after SCI because the spinal cord is capable to learn and adapt after injury.

Body Weight–Supported Treadmill Training

An early neuroscience animal model with implications for people with SCI for locomotor training is the work achieved by Barbeau, Wainberg, and Finch (1987). Modeling hind limb suspension in training a cat with spinal cord transection, these authors used BWSTT to avoid compensatory movement and to provide a true feeling of walking in terms of loading and kinematics (Barbeau, Wainberg, & Finch, 1987). Additionally, BWSTT provides an environment in which to practice stationary walking.

BWSTT has been used as a useful tool to train locomotor function after motor incomplete SCI (Finch, Barbeau, & Arsenault, 1991). Use of BWSTT has been shown to improve coordination of muscle activity of lower extremities and postural alignment (Visintin & Barbeau, 1989), and allows for increased practice of stepping behaviors in people with diminished or absent supraspinal control (Dietz, Wirz, Colombo, & Curt, 1998; Dobkin, Harkema, Requejo, & Edgerton, 1995; Harkema et al., 1997). Senthilvelkumar et al. (2015) examined the effectiveness of body weight-supported over-ground training for improving gait and strength in people with traumatic incomplete tetraplegia. In this study, 16 participants within two years of injury were randomly assigned to one of two groups: body weight-supported over-ground training and BWSTT. Both groups participated in 30 minutes of gait training per day, five days a week for eight weeks. Additionally, both groups received regular rehabilitation, which included flexibility, strength, balance, self-care, and functional training. Both groups had a significant improvement in walking ability and lower extremity muscle strength as compared to the baseline measurements. However, there was no significant difference between groups. The finding suggests that BWSTT and body weight-supported over-ground training may have the potential for improvement of ambulatory function without superiority of one approach over the other.

Effing, Van Meetern, Van Asbeck, and Prevo (2006) completed a pilot study examining the effects of BWSTT on ambulation, balance, and quality of life on three participants with chronic incomplete SCI. All participants were greater than two years post injury and were classified as ASIA C or D. After a baseline was established, the individuals participated in 30-minute BWSTT sessions, five times a week, for 12 weeks.

The finding was that locomotor training on the treadmill actually improved over-ground ambulation function in all three individuals as evidenced by improvements in gait speed and the Timed Up and Go (TUG) scores. Positive trends were also noted in balance as evidenced by improved scores on the Berg Balance Scale (BBS). These balance improvements were noted only in the two individuals with chronic ASIA D incomplete SCI. There were no changes in the quality of life in the three individuals as measured by the Schedule for the Evaluation of Individual Quality of Life (SEIQoL). Although there were improvements in each of the individual's performances, as noted by the authors, there were no consistent trends in all three individuals. However, each individual had different levels of improvements. Each individual also had a customized intervention plan, which limits generalization of outcomes (Effing, Van Meeteren, Van Asbeck, & Prevo, 2006).

To compare BWSTT with over-ground training for walking after incomplete SCI, Dobkin et al. (2006) recruited 146 patients with ASIA B, C, or D in their study. These participants were randomized into two groups: BWSTT or control group. Both groups had 12 weeks of walking training, one hour each day for five sessions weekly. Treadmill speeds were adjusted to enable training at speed greater than 0.72 m/s with a goal of over 1.07 m/s. During BWSTT, the therapists assisted the participant's trunk and lower extremity joint motions required for reciprocal stepping and provided the cutaneous and proprioceptive feedback. Participants in the control group practiced walking in parallel bars or over ground while wearing braces or other devices with assistance provided by one or two therapists during their walking training. If the controls were not able to walk,

they practiced standing and balance. The control group was not allowed to use BWSTT or treadmill. The two approaches were carried out at different locations so that the participants were blinded to the other treatment approach. The primary outcome measures were the Functional Independence Measure (FIM) score and walking speed. There were no significant differences at six months between two groups in the FIM score and walking speed. Therefore, the findings did not support the BWSTT as being more effective than over-ground training or vice versa (Dobkin et al., 2006).

A systematic review investigated whether or not the BWSTT is superior to over-ground gait training in people with traumatic SCI. Ten randomized controlled trials, which compared BWSTT to any other form of gait training except robotic-assisted gait training, were included in review. However, only nine trials provided useable data, and were used in data analysis. The outcome measures of interest were walking speed and distance. The findings of this systematic review supported BWSTT in terms of walking distance but not walking speed. However, the author questioned the results of walking distance because the confidence interval (CI) spans from 31 m to 45 m (Mehrholz, Harvey, Thomas, & Elsner, 2017).

Despite the benefits of BWSTT as an intervention, it has some limitations and burdens on patients and therapists. BWSTT often is very physically intensive and requires the effort of multiple physical therapists (generally up to three) to assist the patient with leg movements and to control trunk movements, particularly for those patients who require substantial walking assistance (Dietz et al., 2011). Therefore, the training duration that the patient could have may be limited due to therapist fatigue.

Moreover, BWSTT provides the patients with appropriate sensory feedback but less about providing normal motions to each joint of the lower limb (Bouyer, 2005).

To overcome these limitations, physical therapists must be creative in their interventions to maximize an individual's potential in both the acute and chronic stages of recovery. As BWSTT provides a normal motion with sensory inputs for the lower limbs for the patient, it could provide high-quality walking training and may promote faster recovery. Robotic gait devices may be able to achieve both through automating locomotor training of people with SCI. Robotic assistive gait devices guide lower extremity movements for gait re-education with less assistance required by physical therapists, yet can provide more intensity of training. Additionally, robotic device training focuses on providing symmetrical gait pattern and over-ground walking training. Moreover, robotic assistive gait devices tend to correct or control primary movement deviations, such as knee hyperextension during the stance phase. Accurately controlling joint motions could help in reducing compensatory movements, which usually accompany poor functional outcomes. For instance, hip hiking and limb circumduction associated with limb clearance during the swing phase can be prevented in robotic exoskeletons.

Robotic-Assisted Gait Training

Initial robotic-assistive gait devices were designed based on human and animal locomotion patterns by researchers and engineers. Some robotic-assistive gait devices are tied to treadmills, but others, such as exoskeletons, are designed to walk over ground with or without a harness system. The early leader in the field of robotic neurorehabilitation

was Hocoma AG, which manufactures the Lokomat (see Figure 2). The Lokomat is a locomotion-training device on a treadmill with a robotic gait orthosis and a suspension system to provide body-weight unloading for the patient while passively moving the legs in a rhythmical walking pattern.

Figure 2. Hocoma, Switzerland lokomat robotic device with permission from Hocoma Inc.: Maryellen Walsh.

The Lokomat has four degrees of freedom (DOFs), left and right hip and knee joints, actuated by four linear drives and a parallelogram structure which allows vertical motion of the patient but prevents lateral motion (Jezernik et al., 2003). The device also provides a passive foot lifter, which helps clear the toes during the swing phase. This device constrains movement on one anatomical plane (sagittal) and thus does not facilitate meaningful balance training. Another robotic device is the LOPES (Veneman et al., 2007). It is designed to allow translational movement of the pelvis with rotational joints at each leg: two at the hip (abduction and flexion) and one at the knee (flexion). Passive foot lifters can be added to induce ankle dorsiflexion, if required. This device has

a total number of eight actuated DOFs, two for the horizontal pelvis translation and three rotational joints per leg, and one DOF for the vertical motion of the pelvis. A more detailed description of the exoskeleton design is presented in Veneman et al. (2007).

The Robotic Ekso GT™ exoskeleton (Ekso Bionics, Richmond, CA, USA) is a device that is considered the first generation of the EKSO (see Figure 3). It was designed specifically for use in rehabilitation treatment and research and has not yet been approved for home use. It enables individuals with lower extremity weakness to stand up and walk while using a walker or a pair of forearm crutches. The EKSO provides natural full weight bearing and reciprocal gait training depending on the level of paralysis or hemiparesis. The EKSO is operated using hydraulic actuators with a hydro-cylinder and battery integrated on a backpack. The EKSO is suitable for the user weighing less than 99.7 kg, and it is adjustable to fit the user whose height can range from 157.50 cm – 193 cm.

Figure 3. Ekso Bionics GT™ robotic exoskeleton

The EKSO functionally allows bilateral hip and knee joint movement to move in sagittal direction and restricts movement in other directions. The ankle joint is allowed to move passively in the sagittal direction. In order to step forward, body shifting is required. Currently, the EKSO has four walk modes. In the first two walk modes, sit-to-stand and moving forward is generated by either a physical therapist or the user by pushing a button, whereas in the other two walk modes, sit-to-stand is generated by the therapist using a push button, but walking is generated by forward and lateral movements of the user's hips to accomplish weight shift (Contreras-Vidal et al., 2016).

Relevant Research about the EKSO

As stated above, the fundamental goal of rehabilitation following SCI is to improve the maximal level of functional recovery while minimizing physical impairments and functional limitations. Many researchers have studied the EKSO with regard to its safety and feasibility as well as locomotor training to promote the recovery of postural control, spasticity, pain, balance, standing, walking, and quality of life (Kolakowsky-Hayner et al., 2013; Kozlowski, Bryce, & Dijkers, 2015; Sale et al., 2016; Stampacchia et al., 2016).

Previous studies focus on the safety and feasibility of using the EKSO in locomotor training (Kolakowsky-Hayner et al., 2013; Kozlowski et al., 2015; Milia et al., 2016). Eight individuals with complete SCI ranging from T4 – T12 within two years of injury participated to evaluate the safety and feasibility of using the EKSO in the Kolakowsky-Hayner et al. (2013) study. Each participant received six weekly sessions with a gradual increase in time. In addition, the level of assistance was decreased as the

practice time was increased. The study concluded that the EKSO is safe for people with complete thoracic SCI in a controlled environment (Kolakowsky-Hayner et al., 2013). Kozlowski et al. (2015) also studies the effects of the use of the EKSO for rehabilitation of SCI. Seven participants with SCI, two with tetraplegia and five with paraplegia, were studied in order to quantify the time and assistance needed to learn to use the EKSO. The participants were able to learn to use the EKSO with minimal assistance in eight sessions. The effort rated by the participant was of mild-to-moderate intensity. In addition, there were no adverse events reported (Kozlowski et al., 2015).

In addition to safety and tolerance studies about the EKSO, the researchers investigated the benefit of using the EKSO regarding the walking function. Thirteen individuals with SCI participated in a study using the EKSO for walking training (Milia et al., 2016). The participants trained five days a week for four weeks. The motor recovery was evaluated using the Six Minute Walk Test (6MWT). The participants showed significant improved in the 6MWT in terms of motor recovery and walking. The participants improved 98.2% or to 128.16m from 76.6m at admission after four weeks of EKSO training (Milia et al., 2016). In the same study, two participants were able to walk with a knee-ankle-foot orthosis after training for at least 16 months. In similar studies of training with the EKSO, participants improved in temporospatial gait parameters including speed, cadence, step length, the TUG score, the 6MWT, and the Ten Meter Walk Test (10MWT) (Ramanujam et al., 2017; Sale et al., 2016; Sale et al., 2018). Despite the small sample size of the studies and the small number of studies published,

the EKSO appears to be a promising intervention approach for locomotor training for individuals with SCI.

In order to trigger the EKSO to initiate a forward step, weight shifting is required from the user. This weight shift training could help in improving the balance. It is well known that balance plays a critical role in maintaining upright standing posture, mobility with confidence, the ability to perform daily activities independently, and the quality of life (Newton, 1997). Center of Mass (CoM) has been studied for individuals with SCI during walking with the EKSO. The dynamic stability (i.e., medial-lateral direction of CoM) was more centered toward the midline when walking with the EKSO (Ramanujam et al., 2017). Additionally, due to a high correlation between BBS and mobility measures in individuals with SCI (Wirz, Muller, & Bastiaenen, 2010), several studies have reported improvements in TUG scores, indicating improved balance (Ramanujam et al., 2017; Sale et al., 2016).

Moreover, evidence exists to support the EKSO for improving many secondary complications for people with SCI. Five adults with SCI with ASIA scale A participated in a study to investigate body composition including total body weight, percentage fat mass (FM), lean body mass (LBM), and bone mineral density (BMD) for their legs using dual energy X-ray absorptiometry. Body mass index (BMI) and cross-sectional areas of the calf muscle were assessed (Karelis et al., 2017). The training lasted up to 60 minutes per session, three times a week for six weeks. The study revealed that total body weight, FM, and BMI were improved significantly. More importantly, there was improvement in BMD, although this improvement was not statistically significant. The cross-section of

calf muscle mass was increased significantly after the intervention. One participant started the study with a diagnosis of osteoporosis, and after the training, his severity of osteoporosis was decreased (Karelis et al., 2017).

In summary, different interventions have been used in rehabilitation centers to improve walking. Gait training has evolved from manual assistance over-ground to over treadmill and recently to the use of a robotic exoskeleton. A variety of robotic exoskeletons are available for training people on walking. The use of a robotic exoskeleton as ambulatory training for people with SCI has shown significant reduction in pain (Kressler et al., 2014), spasticity, improved function of bowel and bladder, sleep (Kozlowski et al., 2015), and improved walking ability, quality of life, as well as reduced depression (Milia et al., 2016).

Although changes of gait parameters were found in healthy adults in a pilot study (Swank et al., 2019), it remains unknown how much the use of the EKSO would influence temporospatial gait parameters, kinematics, and neuromuscular activity in people with incomplete SCI. Therefore, identifying gait parameters in people with incomplete SCI during walking in the EKSO can provide insight for clinicians when using the EKSO.

CHAPTER III

METHODS

SCI often results in physical impairments, which affect a person's ability to walk. Walking inability and mobility limitations are significant components affecting the level of disability and, therefore, overall social health (Seeman et al., 2010). Furthermore, walking is one of the most commonly listed goals by persons with an SCI (Lemay & Nadeau, 2010). Consequently, effective therapeutic intervention, including gait training, is needed to reduce disability and thus the need for physical assistance. Gait training over-ground is considered more desirable than the use of a treadmill (Peshkin et al., 2005). Therefore, exoskeleton robotic devices with the capacity for over-ground walking are being used in rehabilitation practice to facilitate gait training and prompt walking function (Stoller, Waser, Stammler, & Schuster, 2012). The EKSO enables individuals with lower extremity weakness to stand up and walk, which cannot be achieved by using a walker or forearm crutches alone. Therefore, the primary purpose of this study was to evaluate whether people with incomplete SCI would walk differently when they wear the EKSO. Specifically, the temporospatial gait parameters as well as kinematics, and EMG activity of lower extremities during level walking were compared between two conditions: with and without wearing the EKSO. In addition, because the EKSO only allows motions in the sagittal plane, kinematics of hip, knee, and ankle joints were examined in the sagittal plane between the two conditions.

Design

This study was a repeated-measure study design to compare two conditions within one group. In the first condition, temporospatial gait parameters, kinematics, and EMG activities of the participants with incomplete SCI were collected when they walked without wearing the EKSO. During the second condition, the same variables were assessed from the same cohort of participants when they walked while wearing the EKSO. The independent variable of this study was the walking condition with two levels: walking while wearing the EKSO and without the EKSO. The dependent variables included measurements of temporospatial gait parameters, measurements of kinematics and neuromuscular activity of lower extremities during both the stance and swing phases of walking. Specifically, the variables of the temporospatial gait parameters were walking speed, stride length, double-limb-support time, left and right step length, stance time and swing time. The kinematic variables were ROMs of hip flexion-extension, knee flexion-extension, and ankle dorsiflexion-plantarflexion. The EMG variables were muscle activities collected from 12 muscles, including the right and left gluteus medius, rectus femoris, medial hamstrings, tibialis anterior, lateral gastrocnemius, and soleus.

Participants

A power analysis was performed using G*Power 3.1.9.2 (Faul, Erdfelder, Lang, & Buchner, 2007). As indicated in our pilot study, there was a large effect size of 2.0 between the two conditions for kinematic and EMG variables in the healthy adult participants (Swank et al., 2019). A more conservative approach, an effect size of 0.50 was used to estimate appropriate sample size for this dissertation study because of a

different population (i.e., patients with incomplete SCI). With an effect size of 0.50 and an alpha level of 0.05, to estimate the appropriate sample size, nine participants were required to achieve a power of 0.80 for a comparison between two conditions for one group. Considering a 10% attrition rate, 10 participants were recruited for this study. A convenience sample of participants was recruited through local advertising and word-of-mouth from community health professionals, family members, or self-referrals in the Dallas-Fort Worth (DFW) area. An approval from the Institution Review Board at Texas Woman's University was obtained for this study before the commencement of data collection. Upon entry into the study, the principal investigator (PI) or the co-investigator explained the research project to the participants and informed them of the risks and benefits of the study. Once the participants agreed to participate in the study, they signed the informed consent form. Next, a questionnaire of demographic and medical history information was completed by each participant.

Adult men and women with an incomplete SCI of any race or ethnicity between the ages of 18 and 70 years were recruited for this study. Participants were screened by a physical therapist for eligibility to participate in this study. The inclusion criteria were: 1) a classification of the ASIA C or D category (see Appendix A) determined by a physician, 2) a diagnosis of incomplete SCI, at least 4 months prior to participating in the study, 3) sufficient upper extremity strength to use forearm crutches or a walker, and 4) the ability to ambulate at least 10 meters with or without an assistive device (e.g., parallel bars, crutches and a cane). Participants were excluded from the study if they had: 1) a WISCI II (see Appendix B) score < 6, 2) anthropometrics outside the EKSO frame

limitations (> 99.79 kg, < 157.50 cm or 193 cm tall, standing hip width > 41.91 cm, abnormal ROM in hips, knees, ankles, and leg length discrepancy > 1.90 cm), 3) uncontrolled spasticity >3 on the Modified Ashworth Scale (MAS) (Charalambous, 2014) (see Appendix C), 4) unresolved deep vein thrombosis, 5) orthostatic hypotension, 6) significant osteoporosis, 7) skin integrity issues on contact surfaces of the device or sitting surfaces, 8) significant cognitive impairments (unable to follow 3 step commands), and 9) pregnancy.

Outcome Measures

A 3D Vicon motion analysis system (Vicon Motion Systems Ltd., Oxford, UK) was used to obtain kinematic data, including peak-to-peak ROMs of the hip, knee and ankle joints in the sagittal plane. The system consists of 10 infrared cameras with the frame rate set at 120 Hz. These cameras can track the trajectory of the retro-reflective markers during walking trials. The Vicon Plug-In-Gait (PIG) model was used to build a customized model for each participant. The PIG model was selected because it is widely used in gait analysis labs and has been validated (Cappozzo, 1984; Kadaba et al., 1990).

A Delsys EMG system with wireless surface electrodes (Delsys Inc., Natick, MA) (see Appendix E) was used to obtain muscle activity of the right and left the gluteus medius, rectus femoris, medial hamstrings, tibialis anterior, lateral head of the gastrocnemius, and medial aspect of the soleus (see Figure 4). The bandwidth of the EMG system was set at 20 to 450 Hz with a gain of 1,000. The EMG signal was recorded at a sampling rate of 2,000 Hz and timed-synchronized with the VICON Motion Analysis System. Each surface electrode or sensor is rectangular shaped, with a size of 26 × 37 ×

15 mm. Each sensor includes a built-in tri-axial accelerometer with a transmission range of 20m. In addition, each sensor contains an amplifier and two sets of parallel silver contact bars with a fixed distance of 1 cm between the recording bars. One set of bars serves as the active electrode and the other set of bars serves as the reference electrode. Surface EMG has been shown to be more reliable than fine-wire or needle electromyography and can detect more motor units because surface EMG electrodes are larger (Winter, 2009). Test-retest reliability of surface electromyography was found to be moderate to high (ICC = 0.65 - 0.97) (Tseng, Liu, Finley, & McQuade, 2007), and has been shown to be valid for an accurate representation of muscle activation (Marshall & Murphy, 2003).

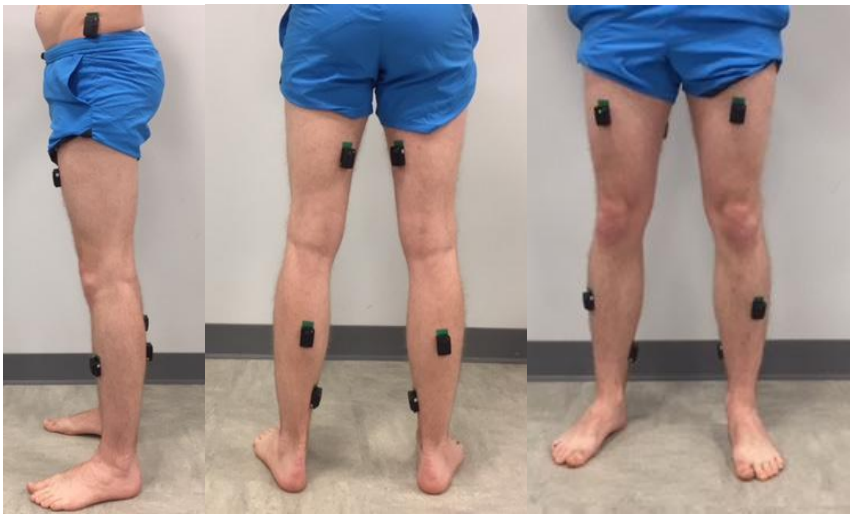


Figure 4. Placements of the electromyographic electrodes on the lower extremities of the participants.

Lastly, six force plates (Advanced Mechanical Technology Inc., Watertown, MA) were time-synchronized with the Vicon Motion System. These force plates were mounted

flush with the floor in the middle of the walkway and were used to determine the gait cycle for each limb during the walking trials. The sampling rate of each force plate is set at 960 Hz, and the threshold for heel-strike and toe-off was set at 10 N.

Ekso GT™ Robotic Exoskeleton

The powered Ekso GT™ robotic exoskeleton or EKSO (Ekso Bionics, Richmond, CA, USA) (see Appendix D) is intended for use in gait training to improve walking function in people with a neurological disorder. The device consists of two legs connected to a torso backpack, which contains an electric motor, a computer, a hand control, and batteries. In addition, two shoulder straps are connected to the torso backpack. When donning the EKSO, the torso backpack is aligned to the user's back, and the legs of the EKSO are aligned with the user's legs by the hook-and-loop straps for the thighs and shins. In the EKSO, the user stands on two non-adjustable foot plates. The length of the legs of the exoskeleton can be adjusted for each user in order to have mechanical joints aligned with the user's anatomical joints (i.e., hip, knee, and ankle joints) to facilitate movements such as sitting, standing up, and walking. The device has four motors, which actuate the hip and knee joint motions in the sagittal plane. The design of the ankle joints only allows passive movements in the sagittal plane.

The EKSO allows for four different modes of step initiation: (1) FirstStep, in which the investigator triggers steps, (2) ActiveStep, in which participants activate their own steps via push buttons on the hand control, (3) ProStep, in which participants start the subsequent step by moving their hips forward and shifting laterally upon which the device triggers the step, and (4) ProStep+, which is similar to ProStep in addition to

lifting foot off the ground to release pressure from forefoot sensors. Additionally, the EKSO allows two different variable assistance of walking: (1) fixed assistance, in which the EKSO provides a value can be programmed from 0 to 100% assistance, and (2) adaptive assistance in which the EKSO allows the patient to voluntarily assist (Gad et al., 2017). For this study, both the ProStep+ mode and the adaptive assistance were used because our participants had incomplete SCI, and they were able to contribute to some movements.

Procedures

Before walking trials, a calibration was performed for the Vicon Motion Analysis system according to manufacturer's recommendations. After providing consent, the potential participants completed an intake form for their demographic data (age, sex, occupation), past medical history, past surgical history, and activity level. Next, anthropometric measurements, including height, weight, standing hip width, ROMs in hips, knees, ankles and leg length, were taken from each participant and used to ensure that the participant could fit the EKSO suit. Once the participant was determined to be eligible for the study, a caliper was used to measure ankle width (between medial and lateral malleoli) and knee width (length between the medial and lateral femoral condyle) for building a PIG model.

Following the anthropometric measurements, participants were asked to walk without wearing the EKSO. Fifteen retro-reflective markers, 14 mm in diameter, were placed bilaterally at the following locations: base of the second toe, calcaneus, lateral malleoli, lateral femoral epicondyle, lateral shank in line with knee and ankle joint

centers, lateral thigh in line with the hip and knee joint centers, anterior superior iliac spine, and on sacrum between the posterior superior lilac spines (see Figure 5).

Figure 5. Placements of the 15 reflective markers on the lower extremities of the participant.

The markers were placed on the bilateral lower extremities for each participant following the Vicon's lower-body PIG model (Duffell, Hope, & McGregor, 2014; Guide, Vicon Plug-in Gait Product, 2010). Table 1 lists the anatomical placements of the 15 markers positioned on the lower extremities of each participant.

Table 1

Position of Markers on the Participants' Lower Extremities

Definition	Position on Participant
Sacral	Mid-way between the posterior superior iliac spines and positioned to lie in the plane formed by the anterior superior iliac spine and posterior superior iliac spine points.

Left ASIS	Left anterior superior iliac spine
Right ASIS	Right anterior superior iliac spine
Left thigh	Over the lower lateral 1/3 surface of the left thigh in line with the hip and knee joint centers
Left knee	On the flexion-extension axis of the left knee
Left tibia	Over the lower 1/3 surface of the left shank
Left ankle	On the lateral malleolus along an imaginary line that passes through the trans-malleolar axis
Left heel	On the calcaneus at the same height above the plantar surface of the foot as the toe marker
Left toe	Over the second metatarsal head, on the mid-foot side of the equinus break between fore-foot and mid-foot
Right thigh	Over the upper lateral 1/3 surface of the right thigh
Right knee	On the flexion-extension axis of the right knee.
Right tibia	Over the upper 1/3 surface of the right shank
Right ankle	On the lateral malleolus along an imaginary line that passes through the trans-malleolar axis
Right heel	On the calcaneus at the same height above the plantar surface of the foot as the toe marker
Right toe	Over the second metatarsal head, on the mid-foot side of the equinus break between fore-foot and mid-foot

Next, the EMG electrodes were affixed to the skin over the targeted muscles with self-adhesive tape. The placements of the electrodes followed the recommendations by the SENIAM (Hermens et al., 2000). Prior to electrode placement, the participant's skin in the areas of electrode placement was cleaned with isopropyl alcohol to improve the quality of the EMG recording. If there was excessive hair, it was shaved using a new disposable electric razor. The electric razor was selected rather than a regular razor due to possible sensory loss of the participants. Once the skin had been appropriately prepared, electrodes were placed on the skin over each muscle parallel to muscle fibers after adjustments to accommodate the exoskeleton's straps. Visual inspection of the EMG signals was performed for each muscle in order to ensure the validity and quality.

Once all markers and EMG electrodes were placed, participants were instructed to stand in a "T" pose for a static trial and this static trial was used to establish zero degrees of joint motion. Following the static trial, participants were instructed to perform 10 walking trials without wearing the EKSO at a self-selected pace with shoes on.

For the EKSO walking trial, EMG electrodes were not removed between conditions, but the markers of the ankle, mid-shank of the tibia, knee and mid-thigh were removed and placed on the EKSO at the level of the corresponding anatomical locations (see Figure 6).

Figure 6. Placements of the 15 reflective markers when the participant wore the Ekso GT™ exoskeleton.

As makers were placed on the EKSO, the knee width and ankle width were re-measured to locate the knee and ankle joint centers for the participants (see Figure 7) using these equations (1) and (2):

$$kw = \{[(Ekw - Tkw) \times 2] + Tkw\} \quad (1)$$

where

kw: knee width

Ekw: knee width when wearing EKSO

Tkw: True knee width

$$aw = \{[(Eaw - Taw) \times 2] + Taw\} \quad (2)$$

where

aw: ankle width

Eaw: ankle width when wearing EKSO

Taw: True ankle width

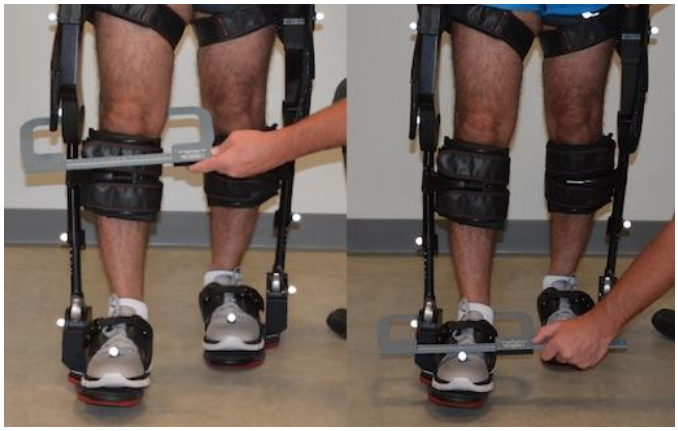


Figure 7. Knee and ankle width measurements while the participant wore the Ekso GT™ exoskeleton.

After each participant donned the EKSO, they were allowed to practice walking in the EKSO for 15 minutes in order to become familiar with the technique necessary to initiate steps in the ProStep+ mode. Following the walking practice, participants were instructed to stand in a "T" pose for a static trial to establish zero degrees of joint motion. Then, participants were asked to walk in the EKSO for 10 good-quality walking trials.

A physical therapist trained in using the EKSO provided supervision and assistance to each participant in the EKSO to prevent a loss of balance. In addition to the outcome measures (i.e., temporospatial gait parameters, kinematics, and EMG activity), data of EKSO use was collected and saved at the end of the visit, including number of

steps, up time, and walking time. The up time for the EKSO started when the participant stood up and ended when the participant sat down.

Signal Acquisition and Processing

Temporospatial, kinematic, and EMG data during gait was collected simultaneously as participants walked with and without wearing the EKSO bionic suit within a motion capture space (see Figure 8). The heel-strike and toe-off events of a gait cycle were determined using a threshold of 10 N vertical force on the force plate. The peak-to-peak ROM in the sagittal plane for the hip, knee, and ankle joints was determined by subtracting the minimum joint angle from the maximum joint angle for the stance and swing phases and for the no EKSO and EKSO conditions, respectively. Lower limb EMG activity was collected from the gluteus medius, rectus femoris, medial hamstrings, tibialis anterior, lateral head of the gastrocnemius, and medial aspect of the soleus, bilaterally. The Vicon Nexus software was used to label markers, interpolate, as necessary, for missing data points, determining the gait events, process the collected data, and finally generate c3d files. Temporospatial, kinematic, and EMG data was extracted from the c3d files using custom MATLAB scripts (Mathworks Inc., MA, USA).

Root-mean-squared (RMS) EMG values were determined using a window size of 120 samples with 60 samples overlapping. Each walking trial was divided into individual gait cycles, which began and ended with the heel strike of the same foot. Each complete gait cycle was further divided into a stance phase (%) and a swing phase (%). The EMG RMS data in each phase was normalized (100%) with 101 data points over corresponding phase. Then, for each walking trial, the collected mean of EMG RMS amplitude was

normalized across the phase of the gait cycle with respect to the peak of the corresponding phase of the gait cycle for each specific muscle across all trials. We elected to use the peak EMG normalization approach because it is the best approach for normalized EMG in neurologically impaired patient populations when obtaining maximal voluntary isometric contractions from 12 muscles was not feasible (Halaki & Ginn, 2012).

Figure 8. Gait Motion Analysis Laboratory.

Statistical Analysis

All statistical analyses were performed using SPSS, version 24.0 (IBM Corp., Armonk, NY). Descriptive statistics, including means, and standard deviations were calculated for the demographic variables. Temporospatial gait parameters were analyzed using paired *t*-tests. If there were no differences in the temporospatial, kinematic and EMG data between the right and the left sides, the averages of the two sides were used for statistical analysis. If there were differences between sides, separate statistical

analyses were performed for each lower extremity. When there was a difference between sides, the collected kinematic data was analyzed using two separate 2 (condition) x 3 (ROM) repeated measures ANOVAs for each lower extremity. Therefore, there would be two for the stance phase and two for the swing phase. Similarly, two separate 2 (condition) x 6 (muscle) repeated measures ANOVAs were used to analyze the normalized EMG data for each lower extremity. There would be two for the stance phase and the other two for the swing phase. If there was a significant interaction in an ANOVA, pairwise comparisons were performed to determine between-condition differences for each dependent variable using the Bonferroni tests. The α level was set at 0.05 for all statistical analyses.

CHAPTER IV

RESULTS

The purpose of this study was to evaluate whether people with incomplete SCI would walk differently when they wear the EKSO compared to their typical walking pattern. Specifically, the temporospatial gait parameters as well as the kinematics and EMG activity of lower extremities during level walking were compared between two conditions: with and without wearing the EKSO. In addition, because the EKSO only allows motions in the sagittal plane, kinematics of hip, knee, and ankle joints were examined in the sagittal plane between the two conditions. This chapter provides a description of the study participants with incomplete SCI as well as descriptive and inferential statistical results of temporospatial gait parameters as well as kinematics, and EMG activity of the lower extremities in the swing and stance phases of gait.

Participants

Fourteen individuals with a history of incomplete SCI were screened for eligibility from February 2017 to September 2018. Three participants were excluded from the study as one participant was classified as ASIA A, one was unable to ambulate, and one was older than 70 years old. Consequently, 11 individuals were enrolled in this study. One participant was unable to complete the study assessment activities and withdrew due to fatigue and increased back pain. Therefore, 10 participants (9 men and 1 woman) with incomplete SCI (5 cervical and 5 lumbar) completed the study and were included for analysis. The participants in this dissertation study had an average age of 39.3 ± 11.7

years, and average weight of 72.5 ± 18.7 kg. Figure 9 illustrates the flow diagram of participant enrollment. A summary of participants' characteristics is provided in Table 2.

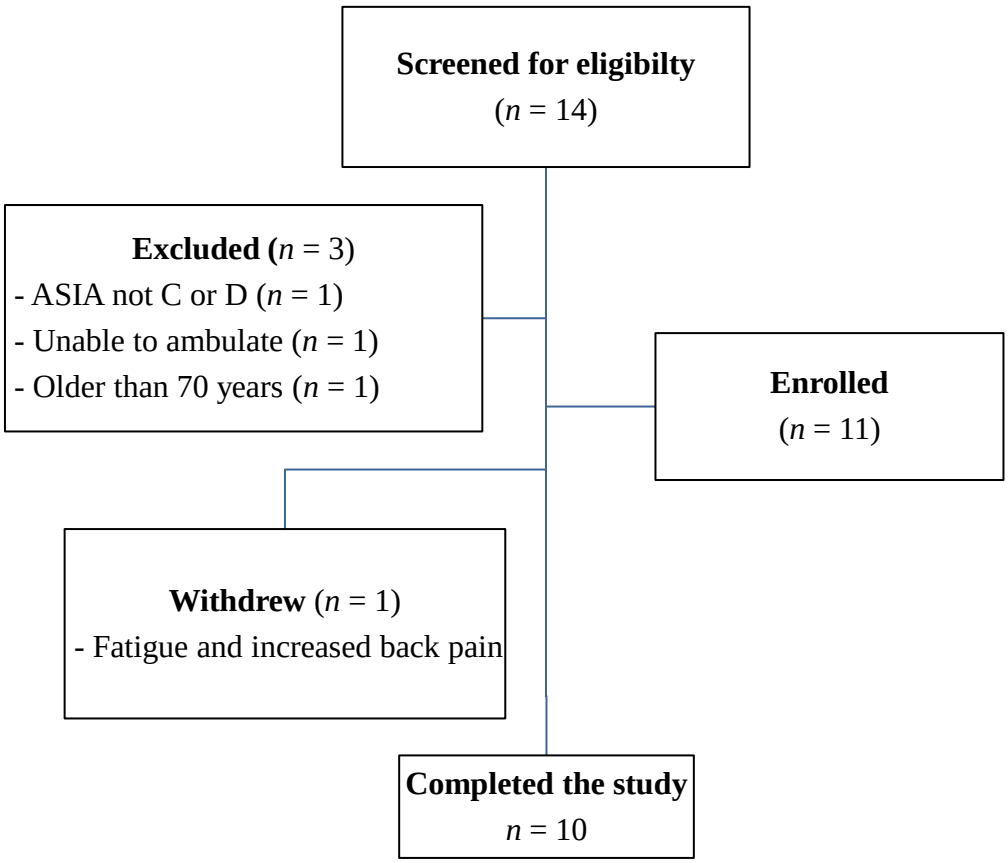


Figure 9. Flow diagram of participant enrollment.

Table 2

Participants' Characteristics

Participant	Time Since Injury (mo)	ASIA Classification	WISCI II Score	AFO	Walking Aids
1	5	D	15	No	Crutches
2	86	C	16	No	Crutches
3	28	C	12	No	Crutches
4	159	D	15	No	Cane
5	175	C	12	Yes	Crutches
6	55	C	9	Yes	Walker
7	14	D	19	No	Cane
8	118	D	16	No	Crutches
9	165	C	15	No	Cane
10	32	D	16	No	Crutches
Average/ Summary	83.7	5C, 5D	14.5	Yes: 2 No: 8	Cane: 3 Crutches: 6 Walker: 1

Note: (mo) = months, ASIA = American Spinal Injury Association, AFO = Ankle-Foot Orthosis, WISCI II = Walking Index for Spinal Cord Injury II.

A licensed physical therapist certified as an EKSO user provided supervision and assistance as needed to the participant during walking with the EKSO trials. All participants were able to walk in the EKSO safely with minimum assistance. Participants were provided

with a 15-minute training session prior to the EKSO walking trials in order to become familiar with the weight shift, which is necessary to elicit the EKSO to make a step. The EKSO's walking session data are presented in Table 3. The EKSO data for Participant #3 was not extracted due to technical problems with the EKSO software.

Table 3

Data of the Ekso GT ™ Exoskeleton Use for All Participants

Participant	Total Number of Steps	Walking Time (Minutes: Seconds)	Up Time (Minutes: Seconds)	Forward Assistance	
				Left	Right
1	256	6: 23	36: 49	77	89
2	185	6: 45	26: 00	96	97
3	-	-	-	-	-
4	255	7: 44	33:28	90	92
5	284	8: 00	44: 19	95	97
6	274	9: 07	24: 40	95	96
7	215	8: 08	24: 12	92	92
8	326	10: 40	29: 00	96	96
9	218	6: 34	25: 41	31	45
10	187	5: 13	17: 51	18	26

Temporospatial Gait Parameters

Because there were no differences in all of the temporospatial gait parameters between the right and left lower extremities during walking for the no EKSO condition (p = 0.360 for stride length, p = 0.439 for step length, p = 0.138 for stance time, and p = 0.247 for swing time), and walking with wearing the EKSO (p = 0.535 for stride length, p = 0.367 for step length, p = 0.976 for stance time, and p = 0.759 for swing time), the average of left and right temporospatial gait parameters were used for statistical analysis.

Participants demonstrated significant differences between walking without wearing the EKSO and with wearing the EKSO on all temporospatial gait parameters (see Table 4) except double-limb-support time and stance time. Overall, participants in the EKSO walked slower with shorter steps and took longer time to complete the gait cycle.

Table 4

Temporospatial Parameters (Mean ± SD) during Gait with and without Wearing the Ekso GT ™ Exoskeleton (EKSO)

Parameter	No EKSO	EKSO	
	Mean ± SD	Mean ± SD	*p* value
Speed (m/s)	0.56 ± 0.32	0.20 ± 0.03	0.006*
Stride length (m)	1.04 ± 0.24	0.65 ± 0.07	0.001*
Double-limb-support time (s)	0.73 ± 0.75	0.90 ± 0.25	0.474

Step length (m)	0.51 ± 0.12	0.33 ± 0.03	0.001*
Stance time (s)	2.03 ± 1.66	2.44 ± 0.42	0.413
Swing time (s)	0.61 ± 0.17	0.80 ± 0.13	0.006*

Note: m = meters, s = seconds, m/s = meters per second, SD = standard deviation,

* Statistically significant at $p < 0.05$.

Kinematics

Tables 5 and 6 show the means and standard deviations for peak-to-peak ROMs of the left and right hip, knee and ankle joints during the stance and swing phases of gait without wearing the EKSO and while wearing the EKSO, respectively.

Table 5

Sagittal Range of Motions (Mean ± SD) of Right and Left of Lower Extremities during Walking without Wearing the Ekso GT™ Exoskeleton (EKSO)

	No EKSO (Stance)			No EKSO (Swing)		
	Right	Left	*p*-value	Right	Left	*p*-value
Hip (°)	33.12 ± 6.50	35.17 ± 11.23	0.483	29.85 ± 6.41	32.42 ± 9.82	0.359
Knee (°)	19.20 ± 8.9	28.64 ± 16.81	0.138	33.96 ± 16.78	47.14 ± 22.01	0.153
Ankle (°)	39.64 ± 38.03	32.16 ± 15.24	0.365	26.39 ± 38.48	21.84 ± 14.07	0.597

Note: SD = standard deviation

Table 6

Sagittal Range of Motions (Mean ± SD) of Right and Left of Lower Extremities during Walking while Wearing the Ekso GT™ Exoskeleton (EKSO)

	EKSO (Stance)			EKSO (Swing)		
	Right	Left	*p*-value	Right	Left	*p*-value
Hip (°)	31.53 ± 3.10	32.97 ± 3.09	0.195	33.01 ± 5.93	36.06 ± 6.98	0.070
Knee (°)	27.94 ± 4.25	27.21 ± 6.45	0.616	35.24 ± 3.86	36.28 ± 3.98	0.454
Ankle (°)	12.33 ± 5.70	10.34 ± 4.02	0.134	6.92 ± 4.16	5.58 ± 2.87	0.480

Note: SD = standard deviation.

Because there were no significant differences in the peak-to-peak sagittal ROMs at the hip, knee, and ankle joints between the right and left lower extremities during walking for both no EKSO and the EKSO conditions, the averages of both lower extremity ROMs were used for statistical analysis. Therefore, two separate 2 (condition) x 3 (ROM) repeated measure ANOVAs were performed to examine the effects of wearing the EKSO during walking on kinematics, one for the stance phase and the other for the swing phase. The ANOVA results showed a significant interaction for the stance phase ($F = 5.99$, $p = 0.034$, $\eta_p^2 = 0.40$), but not for the swing phase ($F = 2.95$, $p = 0.119$, $\eta_p^2 = 0.24$). Post-hoc pairwise comparisons resulted in significant differences between walking conditions (no EKSO vs. EKSO) in the sagittal ROM in the stance phase at the ankle joint ($p = 0.020$), but not for the knee ($p = 0.211$) or for the hip joints ($p = 0.425$). The results indicated that the participants demonstrated similar kinematics at the hip and knee joints between the two walking conditions. However, ankle ROM during walking with the EKSO was significantly less than that during walking without the EKSO. Table

7 summarizes the means and standard deviations for the sagittal ROMs of hip, knee, and ankle joints between walking conditions.

Table 7

Summary of the Mean and Standard Deviation for the Sagittal Range of Motions (Mean ± SD) of Lower Extremities during Walking without the Ekso GT ™ *Exoskeleton (EKSO) and while Wearing the EKSO*

	(Stance)		(Swing)	
	No EKSO	EKSO	No EKSO	EKSO
	Mean ± SD	Mean ± SD	Mean ± SD	Mean ± SD
Hip (°)	34.14 ± 8.04	32.25 ± 2.64	31.14 ± 7.15	34.53 ± 6.04
Knee (°)	23.92 ± 9.87	27.58 ± 4.99	40.55 ± 14.30	35.76 ± 3.31
Ankle (°)	35.90 ± 26.18	11.34 ± 4.55	24.12 ± 25.83	6.25 ± 2.12

Note: SD = standard deviation.

Muscle Activity

Mean amplitude of normalized EMG RMS (%) was calculated bilaterally for the gluteus medius, rectus femoris, medial hamstrings, tibialis anterior, lateral head of the gastrocnemius, and medial aspect of the soleus. Tables 8 and 9 show normalized EMG RMS values for the left and right lower extremities in the stance and swing phases during walking without and with the EKSO, respectively. Because there were differences in most of the EMG RMS values between the right and left lower extremities under each walking condition, four separate repeated measure 2 (condition) x 6 (muscle) ANOVAs were performed to determine the effects of wearing the EKSO during walking on muscle activation, two for the stance phase and two for the swing phase for each side. The ANOVA results showed no significant interactions between the two walking conditions (no EKSO vs. EKSO) for both the right (F = 1.53, p = 0.197, η_p^2 = 0.14), and the left (F = 1.51, p = 0.204, η_p^2 = 0.14) lower extremity muscles in the stance phase, and for both the right (F = 1.45, p = 0.262, η_p^2 = 0.24) and left (F = 1.28, p = 0.286, η_p^2 = 0.12) lower extremity muscles in the swing phase. Table 10 lists the normalized EMG RMS values (%) for the 12 muscles and the differences between the no EKSO and EKSO conditions during the stance and the swing phases. In addition, Figures 10 and 11 illustrate lower extremity muscle activity across the gait cycle.

Table 8

Normalized Electromyographic Amplitudes (%) of the Right and Left Lower Extremities (Mean ± SD) during Walking without Wearing the Ekso GT ™ *Exoskeleton (EKSO)*

	No EKSO (Stance)			No EKSO (Swing)		
	Right	Left		Right	Left	
Muscle	Mean ± SD	Mean ± SD	*p*-value	Mean ± SD	Mean ± SD	*p*-value
GM	0.64 ± 0.12	0.66 ± 0.11	0.312	0.75 ± 0.09	0.75 ± 0.10	0.967
RF	0.65 ± 0.14	0.66 ± 0.09	0.662	0.74 ± 0.12	0.78 ± 0.10	0.108
MH	0.71 ± 0.15	0.63 ± 0.13	0.034*	0.77 ± 0.19	0.74 ± 0.17	0.031*
TA	0.65 ± 0.16	0.71 ± 0.12	0.005*	0.78 ± 0.08	0.84 ± 0.07	0.017*
LG	0.64 ± 0.11	0.61 ± 0.61	0.238	0.79 ± 0.04	0.76 ± 0.07	0.258
SOL	0.68 ± 0.12	0.71 ± 0.12	0.105	0.81 ± 0.04	0.82 ± 0.05	0.442

Note: SD = standard deviation; R = right; L = left; GM = gluteus medius; RF = rectus femoris; MH = medial hamstring; TA = tibialis anterior; LG = lateral gastrocnemius; SOL = soleus.

Table 9

Normalized Electromyographic Amplitudes (%) of the Right and Left Lower Extremities (Mean ± SD) during Walking while Wearing the Ekso GT ™ Exoskeleton (EKSO)

	EKSO (Stance)			EKSO (Swing)		
	Right	Left		Right	Left	
Muscle	Mean ± SD	Mean ± SD	*p*-value	Mean ± SD	Mean ± SD	*p*-value
GM	0.57 ± 0.05	0.59 ± 0.05	0.183	0.75 ± 0.06	0.76 ± 0.08	0.310
RF	0.58 ± 0.07	0.60 ± 0.05	0.350	0.75 ± 0.07	0.77 ± 0.07	0.044*
MH	0.68 ± 0.12	0.58 ± 0.08	0.002*	0.78 ± 0.11	0.73 ± 0.16	0.044*
TA	0.62 ± 0.12	0.69 ± 0.06	0.026*	0.75 ± 0.10	0.82 ± 0.05	0.003**
LG	0.59 ± 0.04	0.56 ± 0.07	0.249	0.75 ± 0.05	0.77 ± 0.06	0.142
SOL	0.64 ± 0.07	0.66 ± 0.09	0.180	0.79 ± 0.06	0.83 ± 0.02	0.050*

Note: SD = standard deviation; R = right; L = left; GM = gluteus medius; RF = rectus femoris; MH = medial hamstring; TA = tibialis anterior; LG = lateral gastrocnemius; SOL = soleus.

Table 10

Normalized Electromyographic Amplitude (%) of Lower Extremities (Mean ± SD) during Gait with and without Wearing the Ekso GT™ Exoskeleton (EKSO) and the Differences Between the Two Walking Conditions

	Stance Phase			**Swing Phase**		
Muscle	No EKSO	EKSO	% Difference	No EKSO	EKSO	% Difference
R GM	0.64 ± 0.12	0.57 ± 0.05	-10.94	0.75 ± 0.09	0.75 ± 0.06	0
R RF	0.65 ± 0.14	0.58 ± 0.07	-10.77	0.74 ± 0.12	0.75 ± 0.07	1.35
R MH	0.71 ± 0.15	0.68 ± 0.12	-4.23	0.77 ± 0.19	0.78 ± 0.11	1.30
R TA	0.65 ± 0.16	0.62 ± 0.12	-4.62	0.78 ± 0.08	0.75 ± 0.10	-3.85
R LG	0.64 ± 0.11	0.59 ± 0.04	-7.81	0.79 ± 0.04	0.75 ± 0.05	-5.06
R SOL	0.68 ± 0.12	0.64 ± 0.07	-5.88	0.81 ± 0.04	0.79 ± 0.06	-2.47
L GM	0.66 ± 0.11	0.59 ± 0.05	-10.61	0.75 ± 0.10	0.76 ± 0.08	1.33
L RF	0.66 ± 0.09	0.60 ± 0.05	-9.09	0.78 ± 0.10	0.77 ± 0.07	-1.28
L MH	0.63 ± 0.13	0.58 ± 0.08	-7.94	0.74 ± 0.17	0.73 ± 0.16	-1.35
L TA	0.71 ± 0.12	0.69 ± 0.06	-2.82	0.84 ± 0.07	0.82 ± 0.05	-2.38
L LG	0.61 ± 0.12	0.56 ± 0.07	-8.20	0.76 ± 0.07	0.77 ± 0.06	1.32
L SOL	0.71 ± 0.12	0.66 ± 0.09	-7.04	0.82 ± 0.05	0.83 ± 0.02	1.22

Note: SD = standard deviation; R = right; L = left; GM = gluteus medius; RF = rectus femoris; MH = medial hamstring; TA = tibialis anterior; LG = lateral gastrocnemius; SOL = soleus.

Figure 10. Normalized Electromyographic Muscle Activity (%) of the Left Lower Extremity Across the Gait Cycle (solid lines represent means and shaded areas represent one standard error).

Note: Lat = lateral, Med Hams = medial hamstrings; % = percentage; Y-axis = Volt

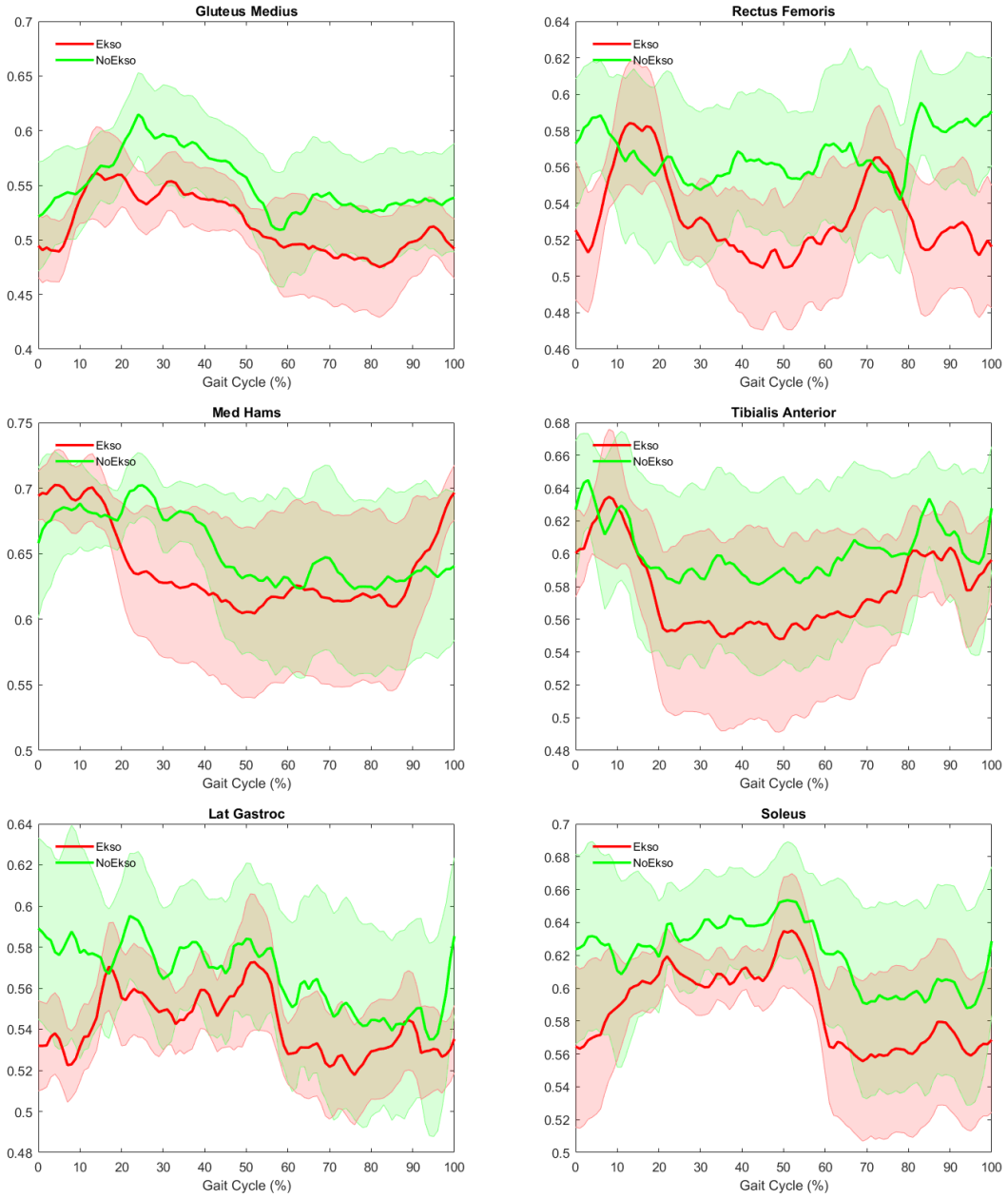


Figure 11. Normalized Electromyographic Muscle Activity (%) of the Right Lower Extremity Across the Gait Cycle (solid lines represent means and shaded areas represent one standard error).

Note: Lat = lateral, Med Hams = medial hamstrings.

CHAPTER V

DISCUSSION

The purpose of this study was to evaluate whether people with incomplete SCI would walk differently when they were wearing the EKSO. Specifically, the temporospatial gait parameters as well as the kinematics and muscle activity of lower extremities during level walking were compared between two conditions: with and without wearing the EKSO. Because the EKSO only allows motions in the sagittal plane, kinematics of hip, knee, and ankle joints were examined in the sagittal plane between the two conditions. A total of 10 participants completed the study. All participants walked without wearing the EKSO first, and then walked while wearing the EKSO. This chapter summarizes and discusses the results, limitations, recommendations for future research, and offers concluding thought.

Research Hypothesis 1

There would be significant differences in the temporospatial gait parameters in people with incomplete SCI between walking while wearing the EKSO and without wearing the EKSO.

Paired *t*-tests revealed that the participants demonstrated significant differences in all temporospatial gait parameters except double-limb-support and stance time when walking wearing the EKSO as compared to walking without wearing the EKSO. On average, participants in the EKSO walked slower with shorter steps and took a longer time to complete the swing phase of the gait cycle. Therefore, Research Hypothesis 1 is

accepted, and the null hypothesis is rejected for the temporospatial gait parameters of speed, stride length, step length, and swing time.

The results indicate that participants with incomplete SCI walked slower while wearing the EKSO than without wearing it. Further, step length decreased and swing time increased while wearing the EKSO. Overall, the participants walked slower by 64% with a 35% shorter step length on both lower extremities when they walked while wearing the EKSO. These results are in agreement with Swank et al. (2019) and Ramanujam et al. (2018), in which both healthy adult participants and participants with incomplete SCI demonstrated a decrease in walking velocity and step length when walking while wearing the robotic exoskeleton. The decreased walking speed may be a result of the inherent limitations of the robotic exoskeleton device (Swank et al., 2019). In order to provide stability during gait, the EKSO does not allow motion in the frontal and the transverse planes at the hip and knee joints and prevents any ankle motions in all directions with a rigid ankle joint. These restrictions may have altered the kinematics, thus reducing walking speed (Veneman, Menger, van Asseldonk, van der Helm, Frans, & van der Kooij, 2008). In addition, the absence of the foot push-off in the terminal stance because of a rigid ankle joint might have prevented participants from achieving greater walking speeds (Pepin, Norman, & Barbeau, 2003). Alternatively, a rigid ankle joint motion in the EKSO may be helpful in preventing foot drop during gait training and therefore decreasing compensatory motion at the knee and hip joints.

Moreover, the EKSO ProStep+ adaptive mode requires participants to shift their weight in order to initiate a step. The learning curve for walking in the EKSO (Kozlowski

et al., 2015) may result in a slower initial walking speed for new EKSO users, such as the participants of this study. Previous studies have shown that experienced users of a robotic exoskeleton had faster walking speed and were able to walk a longer distance after six weeks of training (Kressler et al., 2014; Ramanujam et al., 2018). The decrease in walking speed while wearing the EKSO also might be in part due to the EKSO's weight. The EKSO weights 24kg which is considered to be heavy as compared to other robotic exoskeletons (e.g., ReWalk and Indego; Sale et al., 2018). Further, a previous study showed that faster walking speed increases afferent input and efferent activity during walking in people with SCI (Beres-Jones & Harkema, 2004). Therefore, increased walking speed may play an essential role for rehabilitation of people with SCI, and the slow speed associated with the EKSO may be a barrier for gait rehabilitation.

The results also indicated that the slower walking speed in the EKSO condition appeared to be a result of an increased swing time. Although the stance time and double-limb-support time were increased by 20% and 23%, respectively, these increases were not statistically significant. However, the 32% increase in swing time was statistically significant. It is surprising to find a significant increase in swing time but not in stance and double-support-limb times. The non-significant findings are likely due to the fact that the standard deviations for both the stance time and double-limb-support time were three to four times higher in the no EKSO condition than in the EKSO condition. Therefore, an outlier analysis was performed for the no EKSO data. This analysis showed that an outlier spent much more time in the stance phase and double-limb-support phase of a gait cycle when walking without wearing the EKSO. The statistical analyses without this

outlier showed that all temporospatial gait parameters changed significantly, including the stance time (p = 0.009) and double-limb-support time (p = 0.017). However, all data was analyzed for this study, as it is anticipated that the temporospatial gait parameters in people with SCI are likely to vary, depending on the deficits of motor functions and the severity of abnormal spinal reflexes (Stein et al., 1993).

The participants in this study walked slower (0.56 m/s) without wearing the EKSO as compared to normal walking speed (1.32 m/s; Swank et al., 2018). In addition, the stance phase accounted for 72% of a gait cycle, which is much larger than the average stance phase (approximately 60%) of a normal gait cycle in a person without injury (Murray et al., 1964; Ramanujam et al., 2017). This result is consistent with a previous study by Gil-Agudo et al. (2011), who also found a larger percent stance phase of a gait cycle (approximately 68%) in patients with mild incomplete SCI (Gil-Agudo et al., 2011). However, the stance phase was similar when the participants walked while wearing the EKSO (75% of a gait cycle). The adaptive mode was used in this study; therefore, the EKSO relied on the participants' volitional input to initiate a step, thus resulting in no change of the percentage of the gait cycle for the stance phase for our novice EKSO users.

Research Hypothesis 2

There would be significant differences in the kinematics of the lower extremities in people with incomplete SCI between walking while wearing the EKSO and without wearing the EKSO.

The results showed that participants demonstrated a significant decrease in the

ankle ROM only in the stance phase while walking with the EKSO as compared to the no EKSO condition. Therefore, Research Hypothesis 2 is accepted, and the null hypothesis is rejected for ankle ROM only. It is not surprising to have significant differences in ankle motion, because the EKSO limits motion at the ankle joint. In addition, high variation of ankle dorsiflexion-plantarflexion ROM when the participants walked without EKSO could have contributed the difference. When compared to the ankle ROM of healthy participants without injury during walking without the EKSO, the participants with incomplete SCI in this current study also appeared to have a greater ankle ROM in the stance phase (Swank et al., 2019). The ankle joint of people with SCI has been reported to be in a more dorsiflexed position at the beginning of the stance phase, and remains dorsiflexed even at push-off (Barbeau et al., 1998). Additionally, some people with SCI make initial contact with the forefoot instead of the heel (Perry, 2010). However, because the ankle joint is relatively fixed in the EKSO, ankle ROM in both the healthy adults (Swank et al., 2019) and participants with incomplete SCI in this current study was almost identical (11.74° in healthy adults vs. 11.34° in participants with incomplete SCI) in the stance phase of the gait cycle while wearing the EKSO.

Given that there was significantly reduced ankle ROM in the stance phase while the participants were walking in the EKSO, it was anticipated that there would be a change in ROM at the knee or hip joint. As shown by Swank et al. (2019), there was increased knee ROM with concomitant decreased ankle and hip ROMs when healthy adults walked with EKSO as compared to their normal walking pattern. However, there were no significant differences between conditions at the hip and knee joints in this

current study. Upon a close examination of the data, a large standard deviation was noted for the no EKSO condition, but a relatively small standard deviation for the EKSO condition. A possible explanation is that each participant likely had unique compensatory strategies to accomplish walking after SCI, resulting in diverse kinematic effects, thus accounting for the large standard deviation. However, walking in EKSO provides a relatively standard experience of gait patterning due to the hardware mechanics and software programming, thus accounting for the small standard deviation. Potentially, this standardized trajectory of a gait cycle helps minimize abnormal compensatory strategies and promotes a normalized gait pattern for the individual undergoing robotic exoskeleton gait training. Moreover, the participants in this study were in the chronic stage of SCI, by which they likely have established their abnormal gait patterns (Hornby, Campbell, Zemon, & Kahn, 2005).

It also is speculated that the limited hip and knee ROMs while walking without wearing the EKSO could be due to spasticity. Because spasticity is a velocity-dependent phenomenon, the ROMs of high-velocity joints (e.g., knee) are more likely be limited (Krawetz & Nance, 1996). Additionally, spasticity is often associated with joint stiffness in people with SCI (Barbeau et al., 1998), and may limit joint motion during gait. Lastly, one may argue that the lack of significant findings was due to small sample size. Nonetheless, a small to moderate effect size and power were noted in hip ROM and knee ROM in the stance phase ($\eta_p^2 = 0.40$, $\beta = 0.61$) and in the swing phase ($\eta_p^2 = 0.24$, $\beta = 0.34$).

When the participants of this study (i.e., people with incomplete SCI) walked

while wearing the EKSO, not all kinematics decreased significantly as compared to their typical walking pattern except ankle motion. On the contrary, when healthy adults walked in the EKSO, the kinematics significantly decreased as compared to their typical walking pattern except the hip joint in the swing phase (Swank et al., 2019). This difference may be attributed to the slower gait speed while walking in the EKSO. Nonetheless, both the findings of this study and a study of healthy adults (Swank et al., 2019) question whether the robotic exoskeleton is able to promote normative kinematics. Although the EKSO does not appear to promote normal kinematics, it may be a helpful intervention during early neurorehabilitation. It is often helpful to provide structural support and limit the degree of freedom in early gait training due to motor and sensory deficits of patients with neurological disorders in order to avoid compensatory movement and to ensure patient safety (Hussain, 2014). Further, several sessions of practice may allow participants to walk faster while wearing the EKSO, and therefore may alter lower extremity kinematics (Ramanujam et al., 2018). Precise control of joint kinematics during locomotor training is a desired strategy and can improve patient safety (Tyson, Connell, Busse, & Lennon, 2009). However, previous studies reported that larger improvements were observed in temporospatial gait parameters, kinematics, and performance following completion of training using unconstrained kinematics and assistance when needed as compared to controlling joint kinematics during training (Lewek et al., 2009; Schmidt & Wulf, 1997; Winstein, Pohl, & Lewthwaite, 1994).

Research Hypothesis 3

There would be significant differences in the neuromuscular activity of the lower

extremities in people with incomplete SCI between walking while wearing the EKSO and without wearing the EKSO.

The results showed no significant differences in all lower extremity muscle groups between the two conditions for both the stance phase and the swing phase. Therefore, Research Hypothesis 3 is rejected, and the null hypothesis is accepted. One possible explanation for no significant difference between conditions in muscle activity could be that participants still were required to activate muscles to shift their weight to initiate the step, and participants still were required provide further contribution in order to complete the gait cycle. Therefore, walking while wearing the EKSO did not lower EMG amplitudes or alter muscle activation for participants with incomplete SCI. Similarly, a previous study showed that providing manual assistance for people with incomplete SCI at the lower limbs during walking did not change muscle activity as compared to walking without manual assistance (Domingo, Sawicki, & Ferris, 2007). However, Domingo et al. (2007) reported that muscle activity increased with an increase of speed (1.07 m/s). In this current study, the participants in the EKSO walked with a speed of 20 m/s. Therefore, it may be important to boost the EKSO's speed in order to enhance muscle recruitment.

In this current study, the EKSO ProStep^{+} mode was used, thus increasing the interaction between the participant and the robotic exoskeleton. Although the kinematic trajectory is guided in the EKSO ProStep^{+} mode, the EKSO ProStep^{+} mode allows the participant to contribute to the completion of the gait cycle voluntarily based on their ability. In addition, the EKSO ProStep^{+} mode only provides the missing force when the

participant cannot perform the gait cycle independently, thus improving motor learning during gait training (Pennycott, Wyss, Vallery, Klamroth-Marganska, & Riener, 2012). It is conceivable that when participants in this study (i.e., people with incomplete SCI) walked in EKSO in the ProStep+ mode, they exhibited muscle activity similar to that exhibited during walking without the EKSO. Emerging evidence suggests volitional effort of the participant is essential. Robotic assisted training in which full assistance is provided (i.e., Lokomat), such that stepping is produced without volitional effort from the participant, is less effective than methods requiring volitional effort (Field-Fote & Roach, 2011; Lam et al., 2015). Therefore, the EKSO ProStep+ mode appears to promote a desired effect, namely volitional muscle activity and effort during gait training.

Although the differences between the two conditions were not statistically significant, there was decreased muscle activity (2.82 - 10.94%) in all muscle groups during the gait cycle in the stance phase while walking in EKSO. Particularly, the gluteus medius and rectus femoris muscles demonstrated the largest reduction in muscle activity during stance phase. Although there was no significant reduction in muscle activity during walking while wearing the EKSO, a decrease in right and left gluteus medius muscle activity was observed in the stance phase. It can be speculated that the decrease was due to the structural support from the EKSO. In contrast to healthy adults who demonstrated significant decreased muscle activation during the stance phase while walking in the EKSO (Ramanujam et al., 2018; Swank et al., 2019), this current study showed no significant decrease in muscle activation of all of the lower extremity muscles between conditions. Although the EKSO provides assistance as needed to complete the

gait cycle, the finding suggests that the EKSO may be helpful for maintaining muscle activation during gait training for people with incomplete SCI.

Similarly, decreased muscle activity (0 - 5.06%) also was observed in all muscle groups during the swing phase. One explanation could be that the participants of this study were allowed to use walking aids when walking without wearing the EKSO and with wearing the EKSO as needed, and this may have allowed them to rely more on their arms to assist the leg movement. Specifically, although participants were allowed to use an assistive device, they may have been more reliant on the assistive device during walking in an unfamiliar condition (i.e., walking in EKSO). A previous study (Visintin & Barbeau, 1994) showed that muscle activity of the lower extremities in people with SCI decreased when they walked using walking aids (Visintin & Barbeau, 1994). The level of body loading on the upper extremities could have affected afferent input to the lower extremities, thus decreasing muscle activation (Dietz, Müller, & Colombo, 2002). In this current study, muscle activity of upper extremities was not assessed. Therefore, it is not certain if there was increased muscle activity of upper extremities. Additionally, participants walked slower when wearing the EKSO, and this too could have resulted in decreased muscle activity (Domingo et al, 2007).

In contrast to healthy adults who demonstrated significant decreased muscle activation in the EKSO (Ramanujam et al., 2018; Swank et al., 2019), this current study showed no significant decrease in muscle activation of all of the lower extremity muscles between conditions. Although the EKSO provides assistance as needed to complete the

gait cycle, the finding suggests that the EKSO may be helpful with maintaining the muscle active during the gait training for people with incomplete SCI.

This finding is in contrast to healthy individuals who demonstrated an increase in muscle activity during the swing phase (Swank et al., 2019). Potentially, the participants with incomplete SCI were more compliant than healthy adults in following the EKSO's desired trajectory during the swing phase and did not fight against the robotic exoskeleton as much. Rather, the muscle activity generated by the participants with incomplete SCI simply completed a swing phase of gait that more closely matched their typical kinematic gait pattern. Another potential reason for a lack of increased muscle activity during the swing phase in people with incomplete SCI is the presence of abnormal muscle tone and reflexes (Barbeau et al., 1998). The increase of abnormal reflexes and related reduction of inhibition during the swing phase may have been lessened because of a reduced gait speed. In addition, there may have been less spasticity and overall muscle activity during walking with the EKSO.

Although there was a significant decrease in gait speed while walking wearing the EKSO, muscle activity did not significantly decrease accordingly. Previous studies showed that walking speed might influence muscle activity and kinematics of the lower extremity (Chung & Wang, 2010; Swank et al., 2019). However, several factors other than walking speed also can contribute to reduction of muscle activity. Muscle atrophy combined with impaired neural drive, reduced muscle force and muscle torque can significantly influence muscle activation (Horstman et al., 2008). However, muscle force and torque were not formally assessed in this dissertation study. In addition, the

participants in this study had incomplete SCI; therefore, it is possible that their ability to alter muscle activation was limited because their injured spinal cord may not have been able to interpret different loading during stepping (Beres-Jones & Harkema, 2004). Another possible reason for non-significant findings could be the small effect size and the statistical power (stance phase: $\eta_p^2 = 0.12$, $\beta = 0.41$, swing phase: $\eta_p^2 = 0.26$, $\beta = 0.24$).

Limitations

This study has several limitations impacting our ability to draw firm conclusions. First, this study has a small sample size. A larger sample size is needed in order to increase the statistical power and to obtain a better representation of people with incomplete SCI. The generalizability of the findings of this study is limited to people with the specific criteria applied; for example, only for people classified with ASIA C and D. Furthermore, the amount of gait training in EKSO might not have been sufficient for participants to become familiar with the EKSO. Moreover, there was a variety in the level and time post injury of the participants in this study, which may have affected the results, as these factors may have different influences on the walking characteristics (Krawetz & Nance, 1996). Additionally, many participants used a walking aid for their typical walking, which may have altered muscle activity and, therefore, affected the finding of this study. Lastly, muscle activity and kinematics may have been affected because the EMG electrodes and marker locations were adjusted for the EKSO condition.

Future directions

This study evaluated the EKSO robotic device; however, other robotic devices, such as ReWalk and Indego, are also commercially available and each device may

uniquely impact kinematics, muscle activity, and temporospatial variables. Moreover, each of these devices has different hardware and software features, applies different methods to initiate and complete steps during the gait cycle, and has different levels of FDA approval. For future studies, it is recommended to investigate the ability of robotic exoskeleton devices to help patients replicate the gait of healthy individuals. Furthermore, investigating additional outcome measures such as kinetics and energy consumption cost while walking in a robotic exoskeleton could be clinically meaningful for this patient population. Lastly, determination of the impact of wearing EKSO on gait parameters of individuals with other neurological insult (e.g., stroke) is recommended.

Conclusions

Wearing the EKSO significantly altered temporospatial gait parameters during gait in people with incomplete SCI. Participants walked slower with shorter stride length when walking while wearing the EKSO. However, the EKSO did not make significant changes in the kinematics except at the ankle joint, and muscle activity on people with incomplete SCI, which was possibly due to the reduced walking speed and the heterogeneity of the participants. Future studies with a larger sample size are needed to determine the long-term effect of using the EKSO for gait training.

REFERENCES

Angeli, C. A., Edgerton, V. R., Gerasimenko, Y. P., & Harkema, S. J. (2014). Altering spinal cord excitability enables voluntary movements after chronic complete paralysis in humans. *Brain, 137*(5), 1394-1409.

Barbeau, H., Wainberg, M., & Finch, L. (1987). Description and application of a system for locomotor rehabilitation. *Medical and Biological Engineering and Computing, 25*(3), 341-344.

Barbeau, H., Pépin, A., Norman, K. E., Ladouceur, M., & Leroux, A. (1998). Walking after spinal cord injury: Control and recovery. *The Neuroscientist, 4*(1), 14-24.

Barbeau, H. (2003). Locomotor training in neurorehabilitation: emerging rehabilitation concepts. *Neurorehabilitation and Neural Repair, 17*(1), 3-11.

Behrman, A. L., Ardolino, E. M., & Harkema, S. J. (2017). Activity-based therapy: from basic science to clinical application for recovery after spinal cord injury. *Journal of Neurologic Physical Therapy, 41*(Suppl 3), S39-S45.

Behrman, A. L., & Harkema, S. J. (2007). Physical rehabilitation as an agent for recovery after spinal cord injury. *Physical Medicine and Rehabilitation Clinics of North America, 18*(2), 183-202, v. doi:S1047-9651(07)00018-6 [pii]

Beres-Jones, J. A., & Harkema, S. J. (2004). The human spinal cord interprets velocity-dependent afferent input during stepping. *Brain, 127*(10), 2232-2246.

Blumenfeld, H. (2002). *Neuroanatomy through clinical cases.* Sinauer Sunderland, MA: Oxford University Press.

Bouyer, L. J. (2005). Animal models for studying potential training strategies in persons with spinal cord injury. *Journal of Neurologic Physical Therapy, 29*(3), 117-125.

Cajal, R.Y. (1959). *Degeneration & regeneration of the nervous system*. New York, NY: Hafner Publishing Co.

Cappozzo, A. (1984). Gait analysis methodology. *Human Movement Science, 3*(1-2), 27-50.

Charalambous, C. P. (2014). Interrater reliability of a modified ashworth scale of muscle spasticity. *Classic papers in orthopaedics* (pp. 415-417). Springer, London.

Chisholm, A. E., Alamro, R. A., Williams, A. M., & Lam, T. (2017). Overground vs. treadmill-based robotic gait training to improve seated balance in people with motor-complete spinal cord injury: A case report. *Journal of Neuroengineering and Rehabilitation, 14*(1), 27.

Chung, M., & Wang, M. J. (2010). The change of gait parameters during walking at different percentage of preferred walking speed for healthy adults aged 20–60 years. *Gait & Posture, 31*(1), 131-135.

Cohen, L. G., Celnik, P., Pascual-Leone, A., Corwell, B., Faiz, L., Dambrosia, J., Catala, M. D. (1997). Functional relevance of cross-modal plasticity in blind humans. *Nature, 389*(6647), 180-183.

Contreras-Vidal, J. L., Bhagat, N. A., Brantley, J., Cruz-Garza, J. G., He, Y., Manley, Q., Zhu, F. (2016). Powered exoskeletons for bipedal locomotion after spinal cord injury. *Journal of Neural Engineering, 13*(3), 031001.

Cramer, S. C., Sur, M., Dobkin, B. H., O'Brien, C., Sanger, T. D., Trojanowski, J. Q., Vinogradov, S. (2011). Harnessing neuroplasticity for clinical applications. *Brain : A Journal of Neurology, 134*(Pt 6), 1591-1609. doi:10.1093/brain/awr039

Curt, A., & Dietz, V. (1997). Ambulatory capacity in spinal cord injury: Significance of somatosensory evoked potentials and ASIA protocol in predicting outcome. *Archives of Physical Medicine and Rehabilitation, 78*(1), 39-43.

Curt, A., Keck, M. E., & Dietz, V. (1998). Functional outcome following spinal cord injury: Significance of motor-evoked potentials and ASIA scores. *Archives of Physical Medicine and Rehabilitation, 79*(1), 81-86.

Dietz, V., Wirz, M., Colombo, G., & Curt, A. (1998). Locomotor capacity and recovery of spinal cord function in paraplegic patients: A clinical and electrophysiological evaluation. *Electroencephalography and Clinical Neurophysiology/Electromyography and Motor Control, 109*(2), 140-153.

Dietz, V., Müller, R., & Colombo, G. (2002). Locomotor activity in spinal man: Significance of afferent input from joint and load receptors. *Brain, 125*(12), 2626-2634.

Dietz, V., Nef, T., & Rymer, W. Z. (2011). *Neurorehabilitation technology*. New York: Springer Science & Business Media.

Ditunno, J., Young, W., Donovan, W., & Creasey, G. (1994). The international standards booklet for neurological and functional classification of spinal cord injury. *Spinal Cord, 32*(2), 70.

Ditunno, J. F., Ditunno, P., Graziani, V., Scivoletto, G., Bernardi, M., Castellano, V., Greve, J. D. (2000). Walking index for spinal cord injury (WISCI): An international multicenter validity and reliability study. *Spinal Cord, 38*(4), 234.

Ditunno, P., & Ditunno Jr, J. (2009). Walking index for spinal cord injury (WISCI II): Scale revision. *Spinal Cord, 47*(4), 349.

Ditunno, P., Patrick, M., Stineman, M., & Ditunno, J. (2008). Who wants to walk? preferences for recovery after SCI: A longitudinal and cross-sectional study. *Spinal Cord, 46*(7), 500.

Dobkin, B., Apple, D., Barbeau, H., Basso, M., Behrman, A., Deforge, D., & Spinal Cord Injury Locomotor Trial Group. (2006). Weight-supported treadmill vs over-ground training for walking after acute incomplete SCI. *Neurology, 66*(4), 484-493. doi:66/4/484 [pii]

Dobkin, B. H., Harkema, S., Requejo, P., & Edgerton, V. R. (1995). Modulation of locomotor-like EMG activity in subjects with complete and incomplete spinal cord injury. *Journal of Neurologic Rehabilitation, 9*(4), 183-190.

Domingo, A., Sawicki, G. S., & Ferris, D. P. (2007). Kinematics and muscle activity of individuals with incomplete spinal cord injury during treadmill stepping with and without manual assistance. *Journal of Neuroengineering and Rehabilitation, 4*(1), 32.

Duffell, L. D., Hope, N., & McGregor, A. H. (2014). Comparison of kinematic and kinetic parameters calculated using a cluster-based model and vicon's plug-in gait.

Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine, 228(2), 206-210.

Effing, T., Van Meeteren, N., Van Asbeck, F., & Prevo, A. (2006). Body weight-supported treadmill training in chronic incomplete spinal cord injury: A pilot study evaluating functional health status and quality of life. *Spinal Cord, 44*(5), 287-296.

Esquenazi, A., Talaty, M., & Jayaraman, A. (2017). Powered exoskeletons for walking assistance in persons with central nervous system injuries: A narrative review. *PM&R, 9*(1), 46-62.

Faul, F., Erdfelder, E., Lang, A., & Buchner, A. (2007). G* power 3: A flexible statistical power analysis program for the social, behavioral, and biomedical sciences. *Behavior Research Methods, 39*(2), 175-191.

Field-Fote, E. C., & Roach, K. E. (2011). Influence of a locomotor training approach on walking speed and distance in people with chronic spinal cord injury: A randomized clinical trial. *Physical Therapy, 91*(1), 48-60.

Finch, L., Barbeau, H., & Arsenault, B. (1991). Influence of body weight support on normal human gait: Development of a gait retraining strategy. *Physical Therapy, 71*(11), 842-55; discussion 855-6.

Gad, P., Gerasimenko, Y., Zdunowski, S., Turner, A., Sayenko, D., Lu, D. C., & Edgerton, V. R. (2017). Weight bearing over-ground stepping in an exoskeleton with non-invasive spinal cord neuromodulation after motor complete paraplegia. *Frontiers in Neuroscience, 11*, 333.

Gil-Agudo, A., Pérez-Nombela, S., Forner-Cordero, A., Pérez-Rizo, E., Crespo-Ruiz, B., & del Ama-Espinosa, A. (2011). Gait kinematic analysis in patients with a mild form of central cord syndrome. *Journal of Neuroengineering and Rehabilitation, 8*(1), 7.

Grillner, S., McClellan, A., Sigvardt, K., Wallén, P., & Wilen, M. (1981). Activation of NMDA- receptors elicits "Fictive locomotion" in lamprey spinal cord in vitro. *Acta Physiologica, 113*(4), 549-551.

Guertin, P. A. (2013). Central pattern generator for locomotion: Anatomical, physiological, and pathophysiological considerations. *Frontiers in Neurology, 3*, 183.

Guide, Vicon Plug-in Gait Product. (2010). Foundation notes revision 2.0. *For use with Plug-in Gait Version 2.0 in Vicon Nexus.*

Halaki, M., & Ginn, K. (2012). Normalization of EMG signals: To normalize or not to normalize and what to normalize to? *Computational intelligence in electromyography analysis-a perspective on current applications and future challenges*. IntechOpen.

Harkema, S. J., Schmidt-Read, M., Behrman, A. L., Bratta, A., Sisto, S. A., & Edgerton, V. R. (2012). Establishing the NeuroRecovery network: Multisite rehabilitation centers that provide activity-based therapies and assessments for neurologic disorders. *Archives of Physical Medicine and Rehabilitation, 93*(9), 1498-1507.

Harkema, S., Harkema, S. J., Behrman, A. L., & Barbeau, H. (2011). *Locomotor training: Principles and practice.* New York, NY: Oxford University Press, USA.

Harkema, S. J., Hurley, S. L., Patel, U. K., Requejo, P. S., Dobkin, B. H., & Edgerton, V. R. (1997). Human lumbosacral spinal cord interprets loading during stepping. *Journal of Neurophysiology, 77*(2), 797-811.

Hermens, H. J., Freriks, B., Disselhorst-Klug, C., & Rau, G. (2000). Development of recommendations for SEMG sensors and sensor placement procedures. *Journal of Electromyography and Kinesiology, 10*(5), 361-374.

Horak, F. B. (1991). Assumptions underlying motor control for neurologic rehabilitation. Paper presented at the *Contemporary Management of Motor Control Problems: Proceedings of the II STEP Conference,* 11-28.

Hornby, G., Campbell, D., Zemon, D., & Kahn, J. (2005). Clinical and quantitative evaluation of robotic-assisted treadmill walking to retrain ambulation after spinal cord injury. *Topics in Spinal Cord Injury Rehabilitation, 11*(2), 1-17.

Horstman, A. M., Beltman, M. J., Gerrits, K. H., Koppe, P., Janssen, T. W., Elich, P., & De Haan, A. (2008). Intrinsic muscle strength and voluntary activation of both lower limbs and functional performance after stroke. *Clinical Physiology and Functional Imaging, 28*(4), 251-261.

Hussain, S. (2014). State-of-the-art robotic gait rehabilitation orthoses: Design and control aspects. *NeuroRehabilitation, 35*(4), 701-709.

Jezernik, S., Colombo, G., Keller, T., Frueh, H., & Morari, M. (2003). Robotic orthosis lokomat: A rehabilitation and research tool. *Neuromodulation: Technology at the Neural Interface, 6*(2), 108-115.

Jha, A., Charlifue, S. B., & Lammertse, D. P. (2010). Spinal cord injury and aging. *Spinal Cord Medicine: Principles & Practice, 91*(1), 80-93.

Johnson, R. T., Joy, J. E., Altevogt, B. M., & Liverman, C. T. (2005). *Spinal cord injury: Progress, promise, and priorities.* Washington, D.C: National Academies Press.

Kadaba, M. P., Ramakrishnan, H., & Wootten, M. (1990). Measurement of lower extremity kinematics during level walking. *Journal of Orthopaedic Research, 8*(3), 383-392.

Kapur, A., Kapur, A., Virji-Babul, N., Tzanetakis, G., & Driessen, P. F. (2005). Gesture-based affective computing on motion capture data. Paper presented at the *International Conference on Affective Computing and Intelligent Interaction,* 1-7.

Karelis, A. D., Carvalho, L. P., Castillo, M. J. E., Gagnon, D. H., & Aubertin-Leheudre, M. (2017). Effect on body composition and bone mineral density of walking with a robotic exoskeleton in adults with chronic spinal cord injury. *Journal of Rehabilitation Medicine, 49*(1), 84-87.

Kim, C. M., Eng, J. J., & Whittaker, M. W. (2004). Level walking and ambulatory capacity in persons with incomplete spinal cord injury: Relationship with muscle strength. *Spinal Cord, 42*(3), 156.

King, C. E. (2014). *Brain-Computer Interface Systems for Neurorehabilitation.* (Doctoral dissertation, UC Irvine).

Kirshblum, S. (2004). *New Rehabilitation Interventions in Spinal Cord Injury.* The Journal of Spinal Cord Medicine, 27:4,342-350, doi: 10.1080/10790268.2004.11753772.

Kirshblum, S. C., Burns, S. P., Biering-Sorensen, F., Donovan, W., Graves, D. E., Jha, A., . . . Mulcahey, M. (2011). International standards for neurological classification of spinal cord injury (revised 2011). *The Journal of Spinal Cord Medicine, 34*(6), 535-546.

Ko, C. (2001). Neuromuscular system. In Neil J. Smelser, Paul B. Baltes (Eds.), *International Encyclopedia of the Social & Behavioral Sciences.* Amsterdam: Elsevier.

Kolakowsky-Hayner, S. A., Crew, J., Moran, S., & Shah, A. (2013). Safety and feasibility of using the ekso™ bionic exoskeleton to aid ambulation after spinal cord injury. *Journal of Spine, 4*, 003.

Kozlowski, A., Bryce, T., & Dijkers, M. (2015). Time and effort required by persons with spinal cord injury to learn to use a powered exoskeleton for assisted walking. *Topics in Spinal Cord Injury Rehabilitation, 21*(2), 110-121.

Krawetz, P., & Nance, P. (1996). Gait analysis of spinal cord injured subjects: Effects of injury level and spasticity. *Archives of Physical Medicine and Rehabilitation, 77*(7), 635-638.

Kressler, J., Thomas, C. K., Field-Fote, E. C., Sanchez, J., Widerström-Noga, E., Cilien, D. C., . . . Martinez, A. (2014). Understanding therapeutic benefits of overground bionic ambulation: Exploratory case series in persons with chronic, complete spinal cord injury. *Archives of Physical Medicine and Rehabilitation, 95*(10), 1878-1887. e4.

Kwak, N., Müller, K., & Lee, S. (2015). A lower limb exoskeleton control system based on steady state visual evoked potentials. *Journal of Neural Engineering, 12*(5), 056009.

Lajoie, Y., Teasdale, N., Cole, J. D., Burnett, M., Bard, C., Fleury, M., Lamarre, Y. (1996). Gait of a deafferented subject without large myelinated sensory fibers below the neck. *Neurology, 47*(1), 109-115.

Lam, T., Pauhl, K., Ferguson, A., Malik, R. N., Krassioukov, A., & Eng, J. J. (2015). Training with robot-applied resistance in people with motor-incomplete spinal cord injury: Pilot study. *Journal of Rehabilitation Research & Development, 52*(1), 113-130.

Leblond, H., L'Esperance, M., Orsal, D., & Rossignol, S. (2003). Treadmill locomotion in the intact and spinal mouse. *The Journal of Neuroscience : The Official Journal of the Society for Neuroscience, 23*(36), 11411-11419. doi:23/36/11411 [pii]

Lemay, J., & Nadeau, S. (2010). Standing balance assessment in ASIA D paraplegic and tetraplegic participants: Concurrent validity of the berg balance scale. *Spinal Cord, 48*(3), 245-250.

Lewek, M. D., Cruz, T. H., Moore, J. L., Roth, H. R., Dhaher, Y. Y., & Hornby, T. G. (2009). Allowing intralimb kinematic variability during locomotor training poststroke improves kinematic consistency: A subgroup analysis from a randomized clinical trial. *Physical Therapy, 89*(8), 829-839.

Lovely, R. G., Gregor, R. J., Roy, R. R., & Edgerton, V. R. (1986). Effects of training on the recovery of full-weight-bearing stepping in the adult spinal cat. Experimental neurology, 92(2), 421-435.

Marshall, P., & Murphy, B. (2003). The validity and reliability of surface EMG to assess the neuromuscular response of the abdominal muscles to rapid limb movement. *Journal of Electromyography and Kinesiology, 13*(5), 477-489.

McDonough, A. L., Batavia, M., Chen, F. C., Kwon, S., & Ziai, J. (2001). The validity and reliability of the GAITRite system's measurements: A preliminary evaluation. *Archives of Physical Medicine and Rehabilitation, 82*(3), 419-425.

McKinley, W. O., Seel, R. T., Gadi, R. K., & Tewksbury, M. A. (2001). Nontraumatic vs. traumatic spinal cord injury: A rehabilitation outcome comparison. *American Journal of Physical Medicine & Rehabilitation, 80*(9), 693-699.

McKinley, W. O., Seel, R. T., & Hardman, J. T. (1999). Nontraumatic spinal cord injury: Incidence, epidemiology, and functional outcome. *Archives of Physical Medicine and Rehabilitation, 80*(6), 619-623.

Mehrholz, J., Harvey, L., Thomas, S., & Elsner, B. (2017). Is body-weight-supported treadmill training or robotic-assisted gait training superior to overground gait training and other forms of physiotherapy in people with spinal cord injury? A systematic review. *Spinal Cord, 55*(8), 722.

Milia, P., De Salvo, F., Caserio, M., Cope, T., Weber, P., Santella, C., . . . Bigazzi, B. (2016). Neurorehabilitation in paraplegic patients with an active powered exoskeleton (ekso). *Digital Medicine, 2*(4), 163.

Miller, L. E., Zimmermann, A. K., & Herbert, W. G. (2016). Clinical effectiveness and safety of powered exoskeleton-assisted walking in patients with spinal cord injury: Systematic review with meta-analysis. *Medical Devices (Auckland, N.Z.), 9*, 455-466. doi:10.2147/MDER.S103102

Murray, M. P., Drought, A. B., & Kory, R. C. (1964). Walking patterns of normal men. *JBJS, 46*(2), 335-360.

National spinal cord injury statistical center (NSCISC). (2018). National spinal cord injury statistical center (NSCISC). Retrieved from https://www.nscisc.uab.edu/Public/Facts 2018.pdf

New, P., Simmonds, F., & Stevermuer, T. (2011). A population-based study comparing traumatic spinal cord injury and non-traumatic spinal cord injury using a national rehabilitation database. *Spinal Cord, 49*(3), 397.

Newton, R. A. (1997). Balance screening of an inner city older adult population. *Archives of Physical Medicine and Rehabilitation, 78*(6), 587-591.

Pennycott, A., Wyss, D., Vallery, H., Klamroth-Marganska, V., & Riener, R. (2012). Towards more effective robotic gait training for stroke rehabilitation: A review. *Journal of Neuroengineering and Rehabilitation, 9*(1), 65.

Pepin, A., & Barbeau, H. (1992). Effects of treadmill speed and incline on the walking pattern of normal and spastic paretic subjects [Abstracts]. *Society for Neuroscience*, 18, 860.

Pepin, A., Norman, K., & Barbeau, H. (2003). Treadmill walking in incomplete spinal-cord-injured subjects: 1. adaptation to changes in speed. *Spinal Cord, 41*(5), 257.

Perry, J., Easterday, C. S., & Antonelli, D. J. (1981). Surface versus intramuscular electrodes for electromyography of superficial and deep muscles. *Physical Therapy, 61*(1), 7-15.

Perry, J. (2010). In Burnfield J. M. (Ed.), *Gait analysis Normal and pathological function* (2nd ed.). Thorofare, NJ: SLACK.

Peshkin, M., Brown, D., Santos-Munné, J. J., Makhlin, A., Lewis, E., Colgate, J. E., . . . Schwandt, D. (2005). KineAssist: A robotic overground gait and balance training device. Paper presented at the *Rehabilitation Robotics, 2005. ICORR 2005. 9th International Conference On*, 241-246.

Raineteau, O., & Schwab, M. E. (2001). Plasticity of motor systems after incomplete spinal cord injury. *Nature Reviews Neuroscience, 2*(4), 263-273.

Ramanujam, A., Spungen, A., Asselin, P., Garbarini, E., Augustine, J., Canton, S., . . . Forrest, G. (2017). Training response to longitudinal powered exoskeleton training for SCI. *Wearable robotics: Challenges and trends* (pp. 361-366). Springer, Cham.

Ramanujam, A., Cirnigliaro, C. M., Garbarini, E., Asselin, P., Pilkar, R., & Forrest, G. F. (2018). Neuromechanical adaptations during a robotic powered exoskeleton assisted walking session. *The Journal of Spinal Cord Medicine, 41*(5), 518-528.

Rejc, E., Angeli, C., & Harkema, S. (2015). Effects of lumbosacral spinal cord epidural stimulation for standing after chronic complete paralysis in humans. *PloS One, 10*(7), e0133998.

Sale, P., Russo, E. F., Russo, M., Masiero, S., Piccione, F., Calabrò, R. S., & Filoni, S. (2016). Effects on mobility training and de-adaptations in subjects with spinal cord injury due to a wearable robot: A preliminary report. *BMC Neurology, 16*(1), 12.

Sale, P., Russo, E. F., Scarton, A., Calabro, R. S., Masiero, S., & Filoni, S. (2018). Training for mobility with exoskeleton robot in spinal cord injury patients: A pilot study. *European Journal of Physical and Rehabilitation Medicine, 54*(5), 745-751. doi:10.23736/S1973-9087.18.04819-0

Savic, G., Bergström, E., Frankel, H., Jamous, M., & Jones, P. (2007). Inter-rater reliability of motor and sensory examinations performed according to american spinal injury association standards. *Spinal Cord, 45*(6), 444.

Schmidt, R. A., & Lee, T. D. (2005). *Motor control and learning: A behavioral emphasis.* Champaign, IL: Human Kinetics.

Schmidt, R. A., & Wulf, G. (1997). Continuous concurrent feedback degrades skill learning: Implications for training and simulation. *Human Factors, 39*(4), 509-525.

Seeman, T. E., Merkin, S. S., Crimmins, E. M., & Karlamangla, A. S. (2010). Disability trends among older americans: National health and nutrition examination surveys, 1988-1994 and 1999-2004. *American Journal of Public Health, 100*(1), 100-107.

Senthilvelkumar, T., Magimairaj, H., Fletcher, J., Tharion, G., & George, J. (2015). Comparison of body weight-supported treadmill training versus body weight-supported overground training in people with incomplete tetraplegia: A pilot randomized trial. *Clinical Rehabilitation, 29*(1), 42-49. doi:10.1177/0269215514538068

Sheffler, L. R., & Chae, J. (2015). Hemiparetic gait. *Physical Medicine and Rehabilitation Clinics, 26*(4), 611-623.

Shin, J. C., Kim, J. Y., Park, H. K., & Kim, N. Y. (2014). Effect of robotic-assisted gait training in patients with incomplete spinal cord injury. *Annals of Rehabilitation Medicine, 38*(6), 719-725.

Shumway-Cook, A., & Woollacott, M. H. (2007). *Motor control: Translating research into clinical practice.* Baltimore, Maryland: Lippincott Williams & Wilkins.

Stampacchia, G., Rustici, A., Bigazzi, S., Gerini, A., Tombini, T., & Mazzoleni, S. (2016). Walking with a powered robotic exoskeleton: Subjective experience, spasticity and pain in spinal cord injured persons. *NeuroRehabilitation, 39*(2), 277-283.

Stein, R., Yang, J., Belanger, M., & Pearson, K. (1993). Modification of reflexes in normal and abnormal movements. *Progress in brain research* (pp. 189-196). Amsterdam: Elsevier.

Stoller, O., Waser, M., Stammler, L., & Schuster, C. (2012). Evaluation of robot-assisted gait training using integrated biofeedback in neurologic disorders. *Gait & Posture, 35*(4), 595-600.

Swan and Grundy, B. (2002). *ABC OF SPINAL CORD INJURY*. London: Blackwell Publishing Ltd.

Swank, C., Wang-Price, S., Gao, F., & Almutairi, S. (2019). Walking in A robotic exoskeleton does not mimic natural gait. *JMIR Rehabilitation and Assistive Technologies,* 6(1), e11023. doi:10.2196/11023

Tao, W., Liu, T., Zheng, R., & Feng, H. (2012). Gait analysis using wearable sensors. *Sensors, 12*(2), 2255-2283.

Tseng, S., Liu, W., Finley, M., & McQuade, K. (2007). Muscle activation profiles about the knee during tai-chi stepping movement compared to the normal gait step. *Journal of Electromyography and Kinesiology, 17*(3), 372-380.

Tyson, S., Connell, L., Busse, M., & Lennon, S. (2009). What is bobath? A survey of UK stroke physiotherapists' perceptions of the content of the bobath concept to treat postural control and mobility problems after stroke. *Disability and Rehabilitation, 31*(6), 448-457.

van Hedel, H. J., & Dietz, V. (2010). Rehabilitation of locomotion after spinal cord injury. *Restorative Neurology and Neuroscience, 28*(1), 123-134.

van Hedel, H. J., Dietz, V., & Curt, A. (2007). Assessment of walking speed and distance in subjects with an incomplete spinal cord injury. *Neurorehabilitation and Neural Repair, 21*(4), 295-301. doi:1545968306297861 [pii]

van Middendorp, J. J., Hosman, A. J., Donders, A. R. T., Pouw, M. H., Ditunno Jr, J. F., Curt, A., . . . EM-SCI Study Group. (2011). A clinical prediction rule for ambulation outcomes after traumatic spinal cord injury: A longitudinal cohort study. *The Lancet, 377*(9770), 1004-1010.

Veneman, J. F., Kruidhof, R., Hekman, E. E., Ekkelenkamp, R., Van Asseldonk, E. H., & Van Der Kooij, H. (2007). Design and evaluation of the LOPES exoskeleton robot for interactive gait rehabilitation. *Neural Systems and Rehabilitation Engineering, IEEE Transactions On, 15*(3), 379-386.

Veneman, J. F., Menger, J., van Asseldonk, E. H., van der Helm, Frans CT, & van der Kooij, H. (2008). Fixating the pelvis in the horizontal plane affects gait characteristics. *Gait & Posture, 28*(1), 157-163.

Visintin, M., & Barbeau, H. (1989). The effects of body weight support on the locomotor pattern of spastic paretic patients. *Canadian Journal of Neurological Sciences/Journal Canadien Des Sciences Neurologiques, 16*(03), 315-325.

Visintin, M., & Barbeau, H. (1994). The effects of parallel bars, body weight support and speed on the modulation of the locomotor pattern of spastic paretic gait. A preliminary communication. *Spinal Cord, 32*(8), 540-553.

Wall, T., Feinn, R., Chui, K., & Cheng, M.S. (2015). The effects of the nintendo (TM) Wii fit on functional gait, balance and quality of life in ambulatory individuals with incomplete spinal cord injury. *The Journal of Spinal Cord Medicine*, 38(6), 777-783.

Wernig, A., Müller, S., Nanassy, A., & Cagol, E. (1995). Laufband therapy based on 'rules of spinal locomotion'is effective in spinal cord injured persons. *European Journal of Neuroscience, 7*(4), 823-829.

Whittle, M. (2007). *Gait analysis an introduction* (4th ed.). Edinburgh, NY: Butterworth-Heinemann.

WHO. Spinal cord injury (2019, April 16). Retrieved from http://www.who.int/mediacentre/factsheets/fs384/en/

Widerström-Noga, E. G., Felipe-Cuervo, E., Broton, J. G., Duncan, R. C., & Yezierski, R. P. (1999). Perceived difficulty in dealing with consequences of spinal cord injury. *Archives of Physical Medicine and Rehabilitation, 80*(5), 580-586.

Winstein, C. J., Pohl, P. S., & Lewthwaite, R. (1994). Effects of physical guidance and knowledge of results on motor learning: Support for the guidance hypothesis. *Research Quarterly for Exercise and Sport, 65*(4), 316-323.

Winter, D. A. (2009). *Biomechanics and motor control of human movement.* Waterloo, Ontario: John Wiley & Sons.

Wirz, M., Zemon, D. H., Rupp, R., Scheel, A., Colombo, G., Dietz, V., & Hornby, T. G. (2005). Effectiveness of automated locomotor training in patients with chronic incomplete spinal cord injury: A multicenter trial. *Archives of Physical Medicine and Rehabilitation, 86*(4), 672-680.

Wirz, M., Muller, R., & Bastiaenen, C. (2010). Falls in persons with spinal cord injury: Validity and reliability of the berg balance scale. *Neurorehabilitation and Neural Repair, 24*(1), 70-77. doi:10.1177/1545968309341059

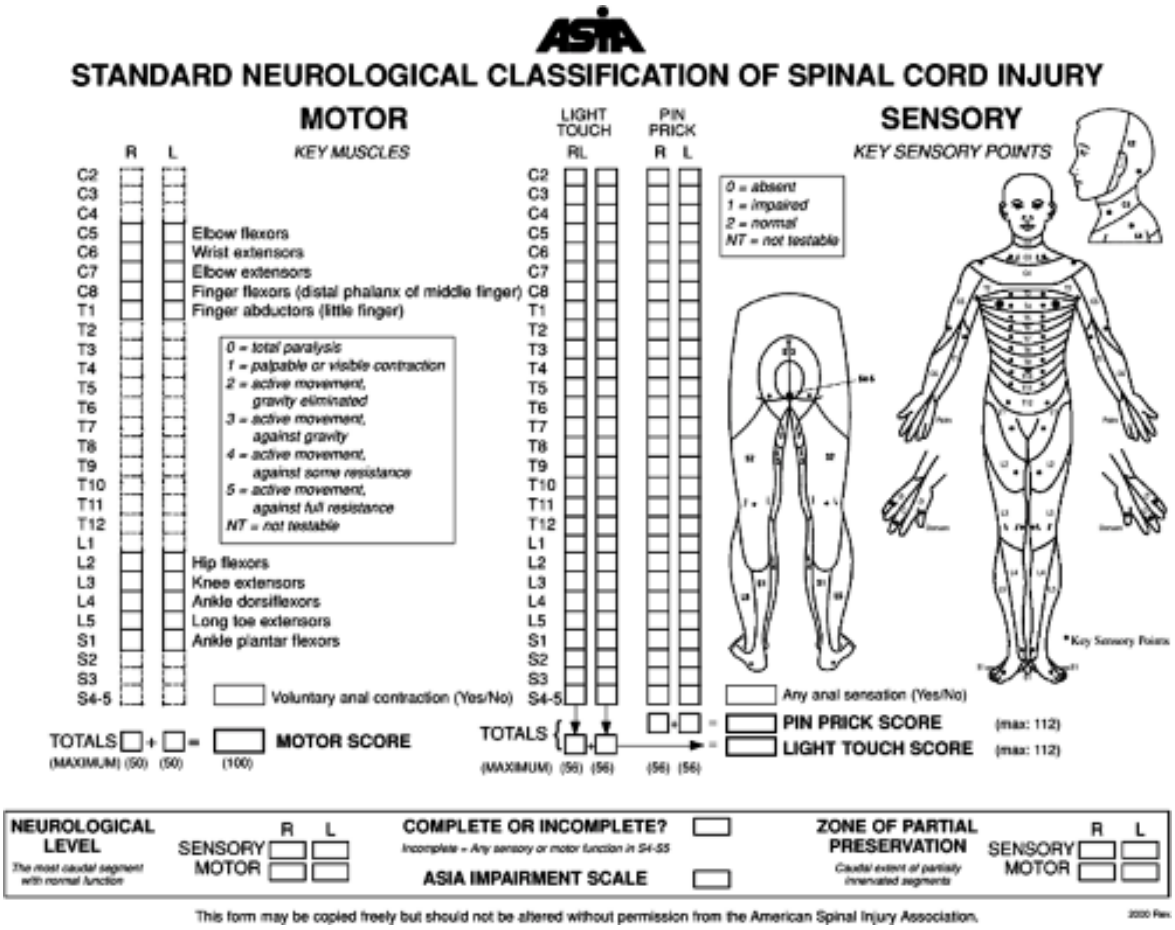
Zeilig, G., Weingarden, H., Zwecker, M., Dudkiewicz, I., Bloch, A., & Esquenazi, A. (2012). Safety and tolerance of the ReWalk™ exoskeleton suit for ambulation by people with complete spinal cord injury: A pilot study. *The Journal of Spinal Cord Medicine, 35*(2), 96-101.

APPENDIX A

American Spinal Injury Association Scale (AIS) (Kirshblum et al., 2011)

- **A = Complete.** No sensory or motor function is preserved in the sacral segments S4-S5.
- **B = Sensory incomplete.** Sensory but not motor function is preserved below the neurological level of injury (NLI) and includes the sacral segments S4-S5, and no motor function is preserved more than three levels below the motor level on either side of the body.
- **C = Motor incomplete.** Motor function is preserved below the NLI, and more than half of key muscle functions below the single neurological level of injury have a muscle grade less than three (Grades 0–2).
- **D = Motor incomplete.** Motor function is preserved below the neurological level, and at least half (half or more) of key muscle functions below the NLI have a muscle grade greater than or equal to 3.
- **E = Normal.** If sensation and motor function as tested with the International Standards For Neurological Classification Of Spinal Cord Injury (ISNCSCI) are graded as normal in all segments, and the patient had prior deficits, then the AIS grade is E. Someone without an SCI does not receive an AIS grade.

Standards for Neurological and Functional Classification of Spinal Cord Injury



APPENDIX B

The Walking Index for Spinal cord Injury (WISCI II)

Level	Description
0	Client is unable to stand and/or participate in assisted walking.
1	Ambulates in parallel bars, with braces and physical assistance of two persons, less than 10 meters.
2	Ambulates in parallel bars, with braces and physical assistance of two persons, 10 meters.
3	Ambulates in parallel bars, with braces and physical assistance of one person, 10 meters.
4	Ambulates in parallel bars, no braces and physical assistance of one person, 10 meters.
5	Ambulates in parallel bars, with no braces and no physical assistance, 10 meters.
6	Ambulates with walker, with braces and physical assistance of one person, 10 meters.
7	Ambulates with two crutches, with braces and physical assistance of one person, 10 meters
8	Ambulates with walker, no braces and physical assistance of one person, 10 meters.
9	Ambulates with walker, with braces and no physical assistance, 10 meters.
10	Ambulates with one cane/crutch, with braces and physical assistance of one person, 10 meters.

11 Ambulates with two crutches, no braces and physical assistance of one person, 10 meters.

12 Ambulates with two crutches, with braces and no physical assistance, 10 meters.

13 Ambulates with walker, no braces and no physical assistance, 10 meters.

14 Ambulates with one cane/crutch, no braces and physical assistance of one person, 10 meters.

15 Ambulates with one cane/crutch, with braces and no physical assistance, 10 meters.

16 Ambulates with two crutches, no braces and no physical assistance, 10 meters

17 Ambulates with no devices, no braces and physical assistance of one person, 10 meters

18 Ambulates with no devices, with braces and no physical assistance, 10 meters.

19 Ambulates with one cane/crutch, no braces and no physical assistance, 10 meters.

20 Ambulates with no devices, no braces and no physical assistance, 10 meters.

APPENDIX C

Modified Ashworth Scale (MAS)

Score	Modified Ashworth Scale Bohannon & Smith (1987)
0	No increase in muscle tone
1	Slight increase in muscle tone, manifested by a catch and release or by minimal resistance at the end of the range of motion when the affected part(s) is moved in flexion or extension
1+	Slight increase in muscle tone, manifested by a catch, followed by minimal resistance throughout the reminder (less than half) of the ROM (range of movement)
2	More marked increase in muscle tone through most of the ROM, but affected part(s) easily moved
3	Considerable increase in muscle tone passive, movement difficult
4	Affected part(s) rigid in flexion or extension

APPENDIX D

Ekso GT™ robotic exoskeleton

APPENDIX E

Delsys electromyographic (EMG)