Effects of Physical Guidance on Motor Control and Learning During Human Walking

by

Antoinette R. Domingo

A dissertation submitted in partial fulfillment
of the requirements for the degree of
Doctor of Philosophy
(Kinesiology)
in The University of Michigan
2009

Doctoral Committee:

Associate Professor Daniel P. Ferris, Chair
Research Professor James A. Ashton-Miller
Associate Professor Rachael D. Seidler
Assistant Professor Riann Palmieri

Acknowledgments

I would not have been able to complete this dissertation without the help and support of the following people. I would like to give them my sincere gratitude:

Dan Ferris: Thank you for giving me the opportunity to earn my PhD at an outstanding University with an outstanding advisor. You gave me the freedom to pursue research in the areas that I was interested in. You also never let me settle. Your enthusiasm and support inspired me to keep going in times of frustration and self-doubt. I was so fortunate to have met you at Berkeley. Thanks for helping me see the glass half full.

Rachael Seidler: You showed interest in my research and always helped me to see things from a different perspective. Your questions and suggestions helped me to take my ideas to the next level.

Riann Palmieri-Smith: You have been a great role model for me as a clinician-scientist. I was glad to have you on my committee because you empathized with my clinical side and took into account my personal goals.

James Ashton-Miller: Thank you for exchanging ideas with me and for having an open mind.

Rodger Kram: My first research advisor at UC Berkeley who got me started in science, and who introduced me to Dan Ferris. Thank you for your exceptional mentorship.

Annie: Thanks for being a great friend and office mate. You were always willing to talk about design ideas and subject protocols with me. Your advice was always so helpful. I learned how to be organized by watching you! I'll never forget that first summer in the lab when you taught me how to use Excel, as well as all the fun times we had outside the lab throughout the years. I'm so glad we got to collaborate on the HNL Digital Short–I look at it as one of my greatest accomplishments during grad school ☺.

Kurt: Thanks so much for helping me build and design the devices used for my experiments, as well as for staying help and going on food runs for us late in the evenings. You also made each data collection more enjoyable for me and the subjects with your friendly and outgoing personality. Your enthusiasm is admirable and I'm looking forward to calling you Dr. Sieloff someday.

Greg: A star scientist who was born to be an academic. I'm proud of the fact we're academic siblings. You've been a great mentor to me, even moreso after you graduated. I'm looking forward to hearing about your continued success.

Keith: Thanks for being so willing to help with my data collections for my first manuscript, and more importantly thank you for all the laughs. I'm glad to call you my academic "brother".

Catherine: Thanks for being a great roommate and for all the help in the lab. I will always remember our adventures in Rome and Munich with a smile. Allora!

Steve Cain: Thank you for all the engineering advice you gave me, as well as for the "breakfast for dinners."

Evelyn: I'm so glad we got to work together during the last year. You're always thinking outside of the box and have such great enthusiasm for research and life. I know you have a great time earning your PhD. Thanks for indulging my compulsive calibrations. You helped to keep me laughing during a very stressful time.

Dani: Thanks so much for helping me with my data collections and all the tracking you did. It helped me immensely during the last stages of my dissertation.

Sarah: Thanks for helping me with the data collections. You're a quick learner and helped to make my data collections more efficient.

Peter: Thank you for your help with writing the code for my data processing and analysis, as well as your helpful input at Kuo-Ferris meetings. I'm glad we got to know each other better at the Preparing Future Faculty seminar. It's rare to meet someone who is so kind and also so intelligent.

Steve Collins: Thank you for input on my destabilization device design, as well as for your friendship. I hope you'll always remember "the good old days" at Michigan.

Felix: Thanks for your friendship over the years and for getting me started with Matlab. Thanks for all the good times and visits to our office in the CCRB basement. Texture forever!

Jiro: I'm so glad to have become friends with you. You were always a friendly face at Kuo-Ferris meetings. Thanks also for Matlab club!

Christine: Thanks for being such a good listener and understanding friend. You always were very encouraging. I'm glad we got to "Docstock" together. I'll always remember the holiday parties at your place. You always make me feel like part of the family.

Joaquin: My fellow Californian—thanks for the friendship over the years and for all the free music! I was glad that we could go to concerts together during grad school. I hope we make it a tradition ☺. Word.

Chris Mendias: I'm so fortunate we met at Summer Institute and that we've stayed friends since then. You always know the right things to say and you are a great shopping buddy! Thanks for all the helpful advice over the years and the continued friendship.

Shawn: Thank you for all your help with my dissertation—you helped me on so many levels. You gave me emotional support and encouragement, especially when I needed it most. I was able to talk to you about my research ideas and data analysis at any hour, late at night, or even early in the morning as you were waking up. You also gave me essential engineering advice when I had to design and build my devices. You showed me not to be afraid of Unistrut, and I learned some important engineering skills from you, even though I was resistant to it. I'm looking forward to the rest of our lives together.

Thanks to my family for their support.

Table of Contents

List of Figures

List of Tables

Abstract

Effects of Physical Guidance on Motor Control and Learning during Human Walking

by

Antoinette Domingo

Chair: Daniel P. Ferris

Physical guidance is often used in rehabilitation when teaching patients to re-learn movements. However, the effects of guidance on motor learning of complex skills are not clear. The overall goal of this dissertation is to determine how physical guidance affects the neural control and motor learning of human walking. In the first experiment, I studied the effects of manual assistance on kinematics and muscle activation during body-weight supported treadmill training in subjects with incomplete spinal cord injury. I found that kinematics and muscle activity did not substantially change when subjects were given manual assistance. Manual assistance allowed subjects with spinal cord injury to train at faster speeds and made muscle activation patterns more similar to those in able-bodied subjects. In the next set of experiments, I used a novel treadmill-mounted balance beam (beam-mill) to study learning of walking balance in neurologically intact human subjects. Subjects practiced walking on the beam-mill with different types of physical assistance and were compared to those that practiced without assistance. In the second experiment, physical assistance was provided with a spring-based stabilization device. Results showed that error-reducing physical assistance hindered learning of the unassisted task. In the third experiment, I investigated whether augmenting error during practice would enhance learning of beam-walking since movement errors drive learning. Two groups of subjects practiced with a destabilization device that had springs with medium or high negative stiffness. Another group walked on a narrower beam to augment error, but with more similar dynamics to the evaluation task. Subjects that practiced unassisted had greater performance gains than those that practiced with error augmentation. However, practicing on the narrow beam had the best performance gains of the error augmentation groups. In the last experiment, subjects practiced with a device that permitted normal movement variability but minimized catastrophic error (i.e. stepping off the beam). Subjects that practiced with this device had very small performance gains, demonstrating that catastrophic errors are important for learning walking balance (if they can be made safely). Overall, these studies support using physical guidance during gait rehabilitation but emphasize that task specificity should be maintained during practice.

Chapter 1. Introduction

Motivation

Physical guidance is often given to patients in rehabilitation settings. It can be given to increase safety, minimize fear, or to help a patient complete a movement they otherwise would be unable to do independently (Wulf, Shea et al. 1998). Some studies have analyzed the effect of physical guidance on performing simple upper limb reaching movements and showed that guidance can be detrimental to motor learning (Armstrong 1970; Winstein, Pohl et al. 1994). For more complex motor skills such as an asymmetrical weight bearing task, guidance was also not helpful for learning (Sidaway, Ahn et al. 2008). In contrast, manually assisted body-weight supported treadmill training has been shown to be effective in restoring gait in patients with neurological injury (Visintin, Barbeau et al. 1998; Behrman and Harkema 2000; Dobkin, Apple et al. 2006). The heterogeneity of outcomes of these studies suggest that principles of motor learning for simple tasks may not be the same for complex tasks (Wulf and Shea 2002). There is still much to be learned about how physical guidance should be used to maximize rehabilitation outcomes.

Robotic devices have recently been developed to provide physical guidance during gait rehabilitation. There is great potential for robotic devices to be useful in rehabilitation settings because of their capability to deliver high intensity and dosage of therapy and reliable measurement of performance (Huang and Krakauer 2009; Marchal-Crespo and Reinkensmeyer 2009). An example of this is the Lokomat (Hocoma AG, Volketswil, Switzerland), a robotic exoskeleton used for locomotor training with body weight support on a treadmill (Colombo, Wirz et al. 2001).

Manually assisted body-weight supported treadmill training has been considered very effective for locomotor training in subjects with spinal cord injury and stroke (Visintin, Barbeau et al. 1998; Dobkin 1999; Behrman and Harkema 2000; McCain, Pollo et al. 2008), but it is also extremely labor intensive. Manually-assisted treadmill training may take up to three therapists to administer and may lead to repetitive stress injury in the trainers. The Lokomat was developed to provide automated locomotor training in patients with neurological injury to help reduce the physical demands on the trainers and also to provide more consistent training within and between sessions. However, studies have shown that using this device may not be as or more beneficial than manually assisted treadmill training in subjects with subacute stroke (Hidler, Nichols et al. 2009) and incomplete spinal cord injury (Israel, Campbell et al. 2006). This could be due in part because it is not known how best to provide physical guidance to maximize learning and rehabilitation outcomes.

I want to examine how physical guidance affects the control and learning of walking balance. In these studies, I will be studying able-bodied subjects learning walking on a narrow balance beam. Once I establish how physical guidance can enhance walking balance in healthy subjects, my long-term goal is to extend these principles to relevant patient populations (spinal cord injury, stroke, elderly).

Balance impairments are common in patients with neurological injury and in the elderly (Shumway-Cook, Anson et al. 1988; Woollacott and Tang 1997; Menz, Lord et al. 2003). Adequate balance is needed to maintain stability during walking, move efficiently, and safely negotiate the environment. Designing a therapeutic intervention for improving walking balance in patient populations could greatly improve functional mobility in millions of individuals and inform how best to design robotic devices used for gait rehabilitation. No studies have examined how physical guidance affects motor learning of walking balance. This knowledge could inform more efficient and effective treatment strategies for gait rehabilitation.

My long-term goal is to design better gait rehabilitation interventions based on fundamental motor learning principles. Without knowledge of how physical guidance affects motor learning during gait, it is not possible to optimize its use during the rehabilitation process. The overall aim of the research described in this dissertation is to determine how different types of physical guidance affect motor learning of a specific locomotor task: walking on a narrow balance beam.

Background

Physical guidance and motor learning

Feedback about performance of motor skills can be administered in a variety of ways. The effects of one type of feedback, knowledge of results, on motor learning has been studied quite extensively (Salmoni, Schmidt et al. 1984). Knowledge of results is augmented feedback that gives information about the extent of error or success of performance after a task is completed. It can be administered in different ways (visual, verbal, etc.) and at different frequencies. It has been shown that while knowledge of results improves performance when it is present, the improvements are no longer present once the feedback is removed (Schmidt and Bjork 1992). When knowledge of results is given at high frequencies during practice, the learner may become dependent on it to guide movement and subsequently does not develop their own strategies for error detection and correction (Salmoni, Schmidt et al. 1984; Sidaway, Moore et al. 1991).

Physical guidance is commonly used in rehabilitation settings and can also be viewed as a form of feedback. Physical guidance is different from knowledge of results because it affects performance during a trial, whereas knowledge of results influences performance on subsequent trials (Winstein, Pohl et al. 1994).

It also changes task dynamics (the motion dependent forces experienced during the task) which may also affect motor learning. If task dynamics are changed, sensory feedback will also likely be affected. This has important implications because motor learning is specific to the sensory feedback available during practice (Proteau, Marteniuk et al. 1992; Proteau, Tremblay et al. 1998). There are instances where physical guidance appears to be helpful in regaining motor skills (e.g., body-weight supported treadmill training). It is important that we find how best to use physical guidance since it is sometimes necessary to ensure safety and prevent injury.

Physical guidance and internal models

It is important to understand the impact of using physical guidance in the context of the internal model for motor control. Based on previous sensorimotor experiences, the central nervous system creates an internal model, or neural representation, of limb dynamics and uses it to determine the motor output for a desired movement in an expected environment. When the dynamics of the limb or environment are different than expected, movement errors result. By comparing the expected sensory feedback to the actual sensory feedback, the internal model is updated and motor output is modified to successfully produce the desired movement (Kawato 1999). Over time, these movement errors drive learning of a new model for the new limb dynamics or environment (Shadmehr and Mussa-Ivaldi 1994). Recent studies have shown that walking may be controlled by an internal model (Lam, Wolstenholme et al. 2003; Pang, Lam et al. 2003; Emken and Reinkensmeyer 2005).

Since errors are needed to update the internal model for motor control, we would expect that using physical guidance that reduces errors to hinder learning. Conversely, since errors are critical to motor learning (Rumelhart, Hinton et al. 1986; Lisberger 1988; Dancause, Ptito et al. 2002), then it could be inferred that increasing errors during practice would improve learning. Studies have shown that a proportionality exists between motor errors and motor learning (Thoroughman and Shadmehr 2000; Scheidt, Dingwell et al. 2001). There is also some evidence that augmenting error during practice actually does enhance motor learning in the upper limb of healthy subjects and subjects with stroke (Wei, Bajaj et al. 2005; Patton, Stoykov et al. 2006), as well as in the legs during a novel walking task in healthy subjects (Emken and Reinkensmeyer 2005).

One important consideration when using physical guidance is that task dynamics may be changed to some extent from the desired task. This depends on how the physical guidance is applied and controlled. Changing task dynamics may hinder learning because the internal model for the task is re-calibrated for the new dynamics created by the physical guidance.

Model of Walking Balance

For the majority of the studies described in this dissertation, I examine the effects of physical guidance on learning to walk on a treadmill mounted balance beam (beam-mill) (Figure 1.1) in able-bodied subjects. The beam-mill provides a continuous walking surface where walking balance is challenged. I also designed the beam-mill so that task difficulty could be varied by changing the widths of the beam.

There has been much research on the control of postural stability as well as quantitative assessments of standing balance (Monsell, Furman et al. 1997; Allum and Shepard 1999; Visser, Carpenter et al. 2008). However, there is little correlation between static balance and dynamic balance or standing and walking balance (Ringsberg, Gerdhem et al. 1999; Owings, Pavol et al. 2000; Shimada,

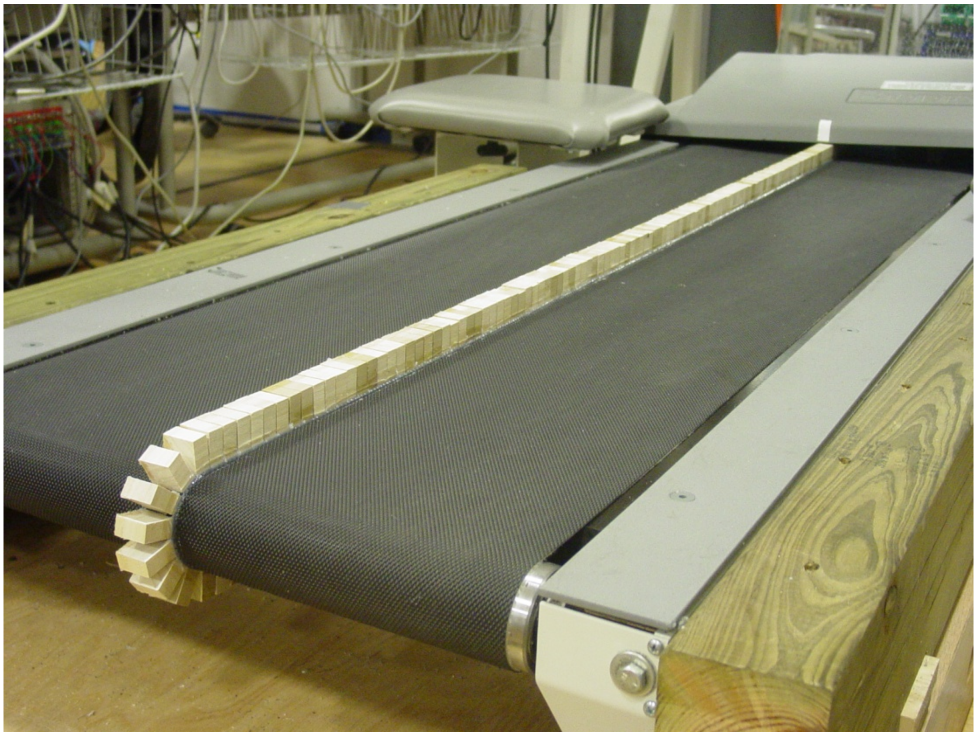

Figure 1. The treadmill mounted balance-beam.
Wood blocks (1.27 or 2.5 cm wide) lined up to make a continuous walking surface. This was used to challenge walking balance in able-bodied subjects.

Obuchi et al. 2003). Since most falls occur during walking, not standing (Blake, Morgan et al. 1988; Niino, Tsuzuku et al. 2000), it is imperative that we devise assessment tools and rehabilitation strategies that specifically target balance during walking.

I chose to study narrow beam walking because it is similar to overground walking but is more challenging to dynamic balance. Beam walking exploits the lateral passive instability of walking (Kuo 1999; Bauby and Kuo 2000). Tandem walking (walking with one foot directly in front of the other) imposes a decreased base of support during walking and is often used in routine neurological testing (Pryse-Phillips and Murray 1992). Examination of reflex modulation in humans has found similar phase-dependency between balance beam walking and treadmill walking (Llewellyn et al, 1990). Past research has found that the central nervous system increases fusimotor drive to increase sensitivity to proprioceptive feedback in cats during beam walking (Prochazka, Hulliger et al. 1987). To compensate for the increased fusimotor drive, the nervous system inhibits H-reflex gain for beam walking compared to normal walking (Llewellyn, Yang et al. 1990). During difficult motor tasks such as beam walking, the modifications in feedback gain likely lead to increased sensory resolution at supraspinal areas and perhaps, sensorimotor instability. These differences and similarities make beam walking a good locomotor task to study motor learning during a challenging locomotor task in healthy individuals.

Dissertation Outline

Chapters 2-5 of this dissertation describe four separate studies. In all of these chapters, I examine the effects of physical guidance on different aspects of walking. In Chapter 2, I look at how manual assistance given during body-weight supported treadmill training affects lower extremity kinematics and muscle activation in subjects with incomplete spinal cord injury. This was an important initial study for my dissertation because body-weight supported treadmill training is considered the “gold-standard” for locomotor rehabilitation in subjects with incomplete spinal cord injury (Dobkin 1999; Behrman and Harkema 2000). In the next three chapters, I examine the effects of different types of physical guidance on learning of narrow beam walking in able-bodied subjects.

Hypotheses

Chapter 2 is a study where I tested subjects with incomplete spinal cord injury (SCI) and measured electromyography and kinematics of the lower extremities while walking with and without manual assistance. I compared these results with data from able-bodied subjects. For this study, there were two competing hypotheses. EMG activity could decrease because the manual assistance given to the subjects would decrease patient effort. EMG activity could also increase because the manual assistance would help to provide more normative joint kinematics and proprioceptive input. This study was published in the *Journal of Neuroengineering and Rehabilitation* in 2007.

In the third chapter, I studied the effects of physical guidance that reduces errors on learning walking balance in able-bodied subjects. Subjects practiced walking on the beam-mill (2.5 cm wide) with or without assistance provided by a spring-based lateral stabilization device. To examine the effects of task difficulty on the relationship between using physical guidance and learning of walking balance, another group of subjects practiced walking on a narrower balance beam (1.27 cm wide) with or without assistance. I hypothesized that subjects who practiced without assistance would have greater performance gains in unassisted beam walking than those that did not. This was based on the idea that error drives motor learning and assistance tends to reduce errors. I also hypothesized the difference in performance gains would be less for those learning the more difficult task (walking on the narrow beam). If a task is too difficult, assistance would be helpful in producing examples of the desired task. For this study, I designed and built the beam-mill. I also built the lateral stabilization device, which was similar to the device used in a study by Donelan and colleagues (2004), in which they examined the mechanical and metabolic cost of lateral stabilization during walking. The data from this study was accepted for publication in *Gait and Posture* in 2009.

In the fourth chapter, I studied how physical guidance that augmented error affected learning of walking balance in able-bodied subjects. Subjects practiced walking on the beam-mill with or without assistance. In this case, assistance was given via a lateral destabilization device that had the properties of a negative stiffness spring. If the subject's hips moved away from the center of the beam, the device applied a proportional force in the same direction as the subject's movement. I hypothesized that subjects that used the destabilization device during practice would have greater performance gains in unassisted walking. This was based on the idea that error is a critical stimulus to learning and that amplifying errors, rather than reducing errors, may enhance motor learning. For this study, I designed and built the destabilization device. This study is being formatted for submission to the journal *Human Movement Science*.

Results from Chapters 3 and 4 suggested that performance gains in beam walking are related to greater movement variability at the pelvis during practice. Greater movement variability may indicate exploration of the subject's limits of stability. The study described in Chapter 5 examines the relative roles of catastrophic error and the exploration of the subjects' state-space in learning to walk unassisted on the beam-mill. One group of subjects practiced walking on the beam with a device that allowed for exploration of the movement space, but minimized catastrophic error (akin to riding a bicycle with training wheels). We compared this group to the group that practiced without assistance. I hypothesized that the group that practiced with catastrophic error (i.e., stepping off the beam) would have greater performance gains than those that practiced without catastrophic error. This was based on the idea that exploration of the movement space is not complete unless catastrophic errors are experienced during practice. This study is being formatted for submission to the *Journal of Neuroengineering and Rehabilitation*.

Chapter 2. Kinematics and muscle activity of individuals with incomplete spinal cord injury during treadmill stepping with and without manual assistance

Abstract

Background

Treadmill training with bodyweight support and manual assistance improves walking ability of patients with neurological injury. The purpose of this study was to determine how manual assistance changes muscle activation and kinematic patterns during treadmill training in individuals with incomplete spinal cord injury.

Methods

We tested six volunteers with incomplete spinal cord injury and six volunteers with intact nervous systems. Subjects with spinal cord injury walked on a treadmill at six speeds (0.18-1.07 m/s) with body weight support, with and without manual assistance. Healthy subjects walked at the same speeds only with body weight support. We measured electromyographic (EMG) and kinematics in the lower extremities and calculated EMG root mean square (RMS) amplitudes and joint excursions. We performed cross-correlation analyses to compare EMG and kinematic profiles.

Results

Normalized muscle activation amplitudes and profiles in subjects with spinal cord injury were similar for stepping with and without manual assistance (ANOVA, $p>0.05$). Muscle activation amplitudes increased with increasing speed (ANOVA, $p<0.05$). When comparing spinal cord injury subject EMG data to control subject EMG data, neither the condition with manual assistance nor the condition without manual assistance showed a greater similarity to the control subject data, except for vastus lateralis. The shape and timing of EMG patterns in subjects with spinal cord injury became less similar to controls at faster speeds, especially when walking without manual assistance (ANOVA, $p<0.05$). There were no consistent changes in kinematic profiles across spinal cord injury subjects when they were given manual assistance. Knee joint excursion was ~5 degrees greater with manual assistance during swing (ANOVA, $p<0.05$). Hip and ankle joint excursions were both ~3 degrees lower with manual assistance during stance (ANOVA, $p<0.05$).

Conclusions

Providing manual assistance does not lower EMG amplitudes or alter muscle activation profiles in relatively higher functioning spinal cord injury subjects. One advantage of manual assistance is that it allows spinal cord injury subjects to walk at faster speeds than they could without assistance. Concerns that manual assistance will promote passivity in subjects are unsupported by our findings.

Introduction

Several investigators have shown that body weight supported treadmill training can improve walking ability in those with incomplete spinal cord injury (Wernig and Muller 1992; Dietz, Colombo et al. 1995; Wernig, Muller et al. 1995; Dietz, Wirz et al. 1998; Wernig, Nanassy et al. 1998; Behrman and Harkema 2000; Hicks, Adams et al. 2005; Dobkin, Apple et al. 2006). During this treatment, the patient is suspended in a standing position above a treadmill by means of a modified parachute harness so that the patient only bears a portion of his weight on their legs. A therapist on each side of the person then manually assists his legs through walking motions while the treadmill belt is moving. A third therapist may also stand behind the patient to help stabilize the trunk. One study showed that 80% of people with incomplete spinal cord injury who used a wheelchair for mobility became functional ambulators after body weight supported treadmill training (Wernig, Nanassy et al. 1998). The effects of this training were maintained long after the intensive treadmill training ended. However, Dobkin et al. performed a multi-center randomized clinical trial that had more equivocal results (Dobkin, Apple et al. 2006). They found that body weight supported treadmill training was no more effective than highly intensive "conventional" physical therapy in improving walking ability. Clearly more research is needed to examine mechanisms and ideal training parameters for body weight supported treadmill training.

Recently, Hidler highlighted the need for more evidence supporting the choice of specific training parameters (Hidler 2005). The amount of body weight support and the walking speed are just a few of the parameters that can greatly vary during treatment. We do not know what is the most effective and efficient manner to set these parameters or how to progress them as a patient makes functional gains. Another factor of training to consider is the use of functional electrical stimulation with locomotor training. Several studies have found therapeutic effects of functional electrical stimulation during gait rehabilitation (Field-Fote 2001; Barbeau, Ladouceur et al. 2002; Field-Fote and Tepavac 2002), but like body weight support and walking speed, it is not clear how to optimize its use.

Another parameter of body weight supported treadmill training that needs to be considered is the amount of mechanical assistance that should be given and the manner in which it is given. One approach is to allow patients to practice stepping on a treadmill with body weight support but no mechanical assistance at all. This could only be done for patients with sufficient motor ability so that body weight support alone facilitated stepping. When this is not possible, the most readily available and most used form of assistance is manual. Unfortunately, this is also very labor intensive and requires a high level of skill to administer. The assistance given could vary from step to step and/or from trainer to trainer. To address these limitations, several groups have developed robotic devices to provide mechanical assistance during stepping (Hesse, Uhlenbrock et al. 2000;

Colombo, Wirz et al. 2001; Emken and Reinkensmeyer 2005; Wirz, Zemon et al. 2005; Sawicki, Domingo et al. 2006).

One possible downside to manual or robotic assistance during body weight supported treadmill training is diminished motor learning. Physical guidance improves performance during the learning phase of an upper limb task while guidance is given, but the improvement in performance is not retained once the guidance is removed (Armstrong 1970; Singer and Pease 1976; Schmidt and Lee 1999). There is no clear evidence on how guidance affects learning in cyclical lower limb tasks. A fundamental assumption of body weight supported treadmill training is that it promotes activity dependent plasticity to improve functional ability. Activity dependent plasticity depends on sufficient and appropriate voluntary drive to promote modifications in synaptic connections (Lotze, Braun et al. 2003; Kaelin-Lang, Sawaki et al. 2005). If manual assistance promotes passivity, then it may be detrimental because diminished neural activation limits the possibility of neural plasticity in relevant circuits.

In contrast, physical guidance may be necessary to learn how to perform a walking movement correctly. Presumably, manual assistance during body weight supported treadmill training helps to ensure that the patient is experiencing the correct kinematics of walking. This could be important because sensory information is an input to the locomotor neural networks. Afferent feedback directly influences the spinal generation of muscle activity that produces human walking (Dobkin, Harkema et al. 1995; Harkema, Hurley et al. 1997; Maegele, Muller et al. 2002; Beres-Jones and Harkema 2004; Ferris, Gordon et al. 2004; Kawashima, Nozaki et al. 2005). Therefore, manual assistance could result in afferent feedback more typical of non-disabled persons during stepping practice. In addition, there are some situations in which learning a movement without physical guidance could be dangerous. When learning to walk after spinal cord injury, manual assistance certainly increases safety, especially when walking at faster speeds.

The purpose of this study was to determine how manual assistance affects lower limb electromyographic (EMG) activity and joint kinematics in subjects with incomplete spinal cord injury during body weight supported treadmill training. There are two competing hypotheses on how EMG activity might be affected by treadmill training with manual assistance. One possibility is that manual assistance decreases the patient’s effort, thereby reducing EMG amplitudes. An alternative possibility is that manual assistance provides more normative kinematic patterns, resulting in more appropriate sensory feedback and increasing EMG amplitudes. We examined individuals with incomplete spinal cord injury that were able to walk with and without manual assistance at multiple speeds during body weight supported treadmill training to compare kinematics and muscle activation. The findings of this study should help to determine if manual assistance affects EMG activity and joint excursions for body weight supported treadmill training.

Methods

Subjects

We tested six adult volunteers with an incomplete spinal cord injury and six neurologically intact adult volunteers. Six subjects with incomplete spinal cord injury (ASIA Impairment Scale Classification of C or D) at the cervical or thoracic level participated in the study. Subjects were at least 12 months post-injury and free of any conditions that would limit their ability to safely complete testing. Five of six subjects were community ambulators with preferred over ground walking speeds of 0.37-0.95 m/s. Of these five subjects, four used canes. Table 2.1 details the cause, classification, level of spinal injury, preferred walking speed, and assistive devices of each subject. Six control subjects (age = 25.8 ± 2.9 years, mass = 66.7 ± 13.4 kg, mean ± SD) without neurological injury also participated in the study. The University of Michigan Institutional Review Board approved this project and all subjects gave informed consent prior to participating.

Procedures

Subjects with spinal cord injury walked on a treadmill with and without manual assistance at six different speeds (0.18, 0.36, 0.54, 0.72, 0.89, 1.07 m/s) with body weight support (Robomedica,Inc., Irvine, CA). All subjects with spinal cord injury underwent one to two training sessions on the treadmill with body weight support prior to data collection to familiarize them with the procedure. The amount of body weight support and stepping speeds achieved varied between subjects due to their different walking abilities. Subjects with spinal cord injury were supported with 30% body weight support unless they required greater support to walk at multiple treadmill speeds. Initially, subjects were asked to walk with 30% body weight support without manual assistance. If they were unable to take steps at this level of support at 0.36 m/s, body weight support was increased in 10% increments until the subject could walk safely at this speed without manual assistance. Three subjects walked with 30% body weight support, two subjects walked with 50% body weight support, and one subject walked with 60% body weight support. The goal of the manual assistance was to minimize gait deviations (e.g., increasing step length, increasing toe clearance and hip flexion during swing). We attempted to collect data at all speeds for all subjects but only two subjects were able to walk at all six speeds with and without assistance. We collected data on the remaining subjects from the trials they were able to safely complete. Table 2.1 shows the stepping speeds each subject was able to achieve. Subjects who normally used lower limb orthoses wore them during testing to ensure their safety (Table 2.1). Control subjects walked on the treadmill without manual assistance at all speeds with 30% body weight support to match the baseline condition of the subjects with spinal cord injury.

The same trainers manually assisted all subjects following the procedures described by Behrman and Harkema for locomotor training with partial body weight support (Behrman and Harkema 2000). The trainers were instructed and supervised by a former trainer who was from the UCLA Human Locomotion

Research Center that directed a large scale clinical trial on body weight supported treadmill training (Dobkin, Apple et al. 2003).

Data acquisition and analysis

While walking under the two experimental conditions, we collected surface electromyographic and kinematic data. We used a Konigsberg Instruments, Inc. (Pasadena, CA) telemetry EMG system to record activity from eight muscles on one lower limb (tibialis anterior, TA; soleus, SO; medial gastrocnemius, MG; lateral gastrocnemius, LG; vastus lateralis, VL; vastus medialis, VM; rectus femoris, RF; and medial hamstring, MH). Inter-electrode distance was 2.5 cm for all subjects and muscles. Electrodes were circular with a diameter of 1.1 cm. We verified that cross-talk was negligible by visual inspection of the EMG signals (Winter, Fuglevand et al. 1994). We also used footswitches to delineate the stance phase and swing phase of gait. We placed electrogoniometers (Biometrics, Ltd., Ladysmith, VA) at the ankle, knee and hip joints on each leg to record joint angles. If the patient wore an ankle foot orthosis, the goniometer was placed on the outside of the orthosis. The computer collected all analog data at 1200 Hz for 15-25 seconds per trial depending on speed (Motion Analysis Corporation, Santa Rosa, CA). Subjects also wore footswitches as insoles to indicate the time each foot was or was not on the ground (B & L Engineering, Tustin, CA). Contacts in the footswitches were at the heel, fifth metatarsal, first metatarsal, and great toe to signify when those areas of the foot were tttbearing weight. Subjects with spinal cord injury performed two trials of each condition (with and without manual assistance) and speed in a randomized order. Between

4 and 19 steps were analyzed per trial depending on speed. The difference in number of steps analyzed across trials and subjects was not likely to artificially alter the results (Arsenault, Winter et al. 1986). Although some subjects could walk at faster speeds with manual assistance than they could without, only trials from speeds at which the subject could walk both with and without manual assistance were included. We only analyzed EMG and kinematic data from speeds that subjects could both walk with and without assistance because EMG amplitudes are a function of walking speed and including the data from the higher walking speeds would skew the results.

We used commercial software (Visual 3D, C-Motion, Inc., Rockville, MD) to process collected EMG and kinematic data. EMG data were high-pass filtered (20 Hz) to remove artifacts while preserving the integrity of the data, and then rectified and low-pass filtered (25 Hz). Kinematic data were low pass filtered at 6 Hz (Winter 1990). Averaged EMG and kinematic profiles were time normalized to the percentage of the stride cycle, beginning and ending with heel strike of the same foot. We calculated the EMG root-mean-square (RMS) for each step cycle within a trial for each muscle, and then averaged these values for an overall RMS value for each trial. We also calculated separate RMS values for the stance and swing phases of gait.

For each muscle, we normalized EMG RMS data to the highest average RMS that occurred in that muscle without manual assistance during one of the two trials at 0.36 m/s. We chose this speed for normalization because it was the highest speed that all subjects with spinal cord injury could achieve. Using JMP statistical software (Cary, NC), we used a repeated measure ANOVA (individual subject by speed by condition) to test for significant differences between normalized RMS values for the stance and swing phases separately. We also used a repeated measure ANOVA (individual subject by speed by condition) to test for significant differences between joint range of motion values. Tukey HSD post-hoc tests were performed to identify differences between specific groups.

We performed cross-correlation analyses using Equation (1) to compare averaged electromyographic waveforms and kinematic profiles of control subjects with the profiles of each spinal cord injury subject with and without manual assistance (Huang and Ferris 2004; Kao and Ferris 2005; Wren, Do et al. 2006).

$$R = \frac{\Sigma x_i y_i}{\left(\Sigma x_i^2\right)^{1/2}\left(\Sigma y_i^2\right)^{1/2}} \qquad (1),$$

where x_i and y_i are two series of data, and i = 0, 1, 2, …, N-1. The first series of data was the averaged control subject data, and the second series was the data from individual subjects with spinal cord injury. Because the data were normalized to the percentage of the gait cycle, N = 101 in all analyses. We used the cross-correlation results to assess if manual assistance altered the shape and timing of muscle activation and kinematic profiles of subjects with spinal cord injury so that it was more similar to control subject profiles. We also performed cross-correlation analyses to compare EMG waveforms and kinematic profiles of subjects with spinal cord injury walking with manual assistance to walking without manual assistance. We performed repeated measure ANOVAs (individual subject by speed by condition) to test for significant differences in R-values and time lags. Tukey HSD post-hoc tests were performed to identify specific differences between groups. Power analyses were also carried out where appropriate.

We calculated coefficients of variation (CV) of EMG activation and joint angle profiles using Equation (2) to quantify variability of the different conditions (Winter 1991).

$$CV = \frac{\sqrt{\frac{1}{N}\sum_{i=1}^{N}\sigma_i^2}}{\frac{1}{N}\sum_{i=1}^{N}\left|X_i\right|} \qquad (2),$$

where N is the number of intervals over the stride, X_i is the mean value of the variable at the ith interval, and σ_i is the standard deviation of variable X about X_i. We performed a repeated measure ANOVA (individual subject by speed by condition) to test for significant differences in the coefficients of variation of the joint angle profiles. We performed post-hoc tests and power analyses as described above.

Results

Three of six subjects with spinal cord injury could walk at faster speeds with manual assistance than without. The average highest walking speed without manual assistance was 0.76 m/s. The average walking highest speed with manual assistance was 0.95 m/s (Table 2.1).

Electromyography

There were clear differences between muscle activation patterns in subjects with spinal cord injury and control subjects. However, muscle activation profiles in subjects with spinal cord injury walking with manual assistance were very similar to profiles while walking without manual assistance (Figures 2.1, 2.2, and 2.3). Cross-correlation analyses of average EMG waveforms between with and without manual assistance produced correlation values greater than 0.89 and phase lags less than 2% (Table 2.2). When comparing spinal cord injury data to control data, neither the condition with manual assistance nor the condition without manual assistance showed a greater similarity to the control subject data (correlation and phase lag, ANOVA, $p > 0.05$). The exception was that when the subjects with SCI were given manual assistance, the profile of the vastus lateralis activation was more similar to the profile of the control subjects ($p = 0.002$, $R = 0.91$ without manual assistance, $R = 0.93$ with manual assistance)..

Muscle activation amplitudes in subjects with spinal cord injury walking with manual assistance were very similar to amplitudes during walking without manual assistance (Figures 2.4 and 2.5). There were no significant differences in normalized EMG RMS between the two conditions for any muscles (ANOVA, $p > 0.05$), except VM during stance (ANOVA, $p = 0.02$).

There were increases in muscle activation amplitudes of subjects with spinal cord injury with speed. Stance EMG RMS increased from slowest to fastest speeds for all experimental conditions in soleus (96%), medial gastrocnemius (120%), vastus lateralis (44%), rectus femoris (48%), and vastus medialis (61%) (all $p < 0.01$) (Figure 2.4). Swing EMG RMS increased in soleus (61%), medial gastrocnemius (33%), vastus medialis (61%), and vastus lateralis (49%) (all $p < 0.04$) (Figure 2.5). The remaining muscles did not have significant increases in EMG RMS ($p > 0.05$).

The shape of muscle activation patterns in subjects with spinal cord injury tended to become less similar to controls at faster speeds, especially when walking without manual assistance. When comparing the without manual assistance condition to controls, R-values became significantly less from the slowest to the fastest speed in TA (0.85 to 0.83), SO (0.87 to 0.80), MG (0.84 to 0.74), LG (0.85 to 0.74), VM (0.94 to 0.90), and VL (0.94 to 0.90) (ANOVA, $p < 0.05$). The phase shift also became larger with increasing speed in LG (5 to –26) ($p < 0.05$). When comparing the manual assistance condition to controls, only the TA had a significantly lower R-value with increasing speed (0.87 to 0.83) (ANOVA, $p < 0.05$).

Kinematics

Kinematic profiles in subjects with spinal cord injury walking with manual assistance were very similar to profiles while walking without manual assistance (Figures 2.1, 2.2, and 2.3). Cross-correlation analyses between with and without manual assistance produced correlation values greater than 0.77 and phase lags less than 3% (ANOVA, $p < 0.05$) (Table 2.2). There were small differences in range of motion between conditions (Table 2.3). During swing, knee joint excursion was ~5 degrees greater with manual assistance (ANOVA, $p < 0.05$). During stance, hip and ankle joint excursion were both ~3 degrees lower with manual assistance (ANOVA, $p < 0.05$).

There were differences in the results of the cross-correlation analyses when we compared the shape and timing of kinematic profiles of spinal cord injury subjects walking with and without manual assistance to control subject data. There was a higher R-value and smaller time shift at the knee joint in the comparison of walking with manual assistance to control data than in the comparison of walking without manual assistance to control data (R, ANOVA $p = 0.003$; time shift, ANOVA $p = 0.011$) (Table 2.2).

Range of motion of the joints increased with increasing speed in the subjects with spinal cord injury. At faster speeds, ankle range of motion over the whole gait cycle increased by 63% (ANOVA, $p = 0.003$). Hip range of motion increased with increasing speed during the stance phase (67%) and swing phase (64%) (ANOVA, $p < 0.001$).

Kinematic Variability

Variability was less at the ankle joint when subjects with spinal cord injury were given manual assistance (CV = 0.46 without manual assistance, CV = 0.34 with manual assistance, ANOVA, $p = 0.03$). There were no clear differences in kinematic variability between the with and without manual assistance conditions at the knee or hip (ANOVA, $p > 0.05$). Figure 2.6 shows mean joint angles ± 1 SD for all six subjects with spinal cord injury during walking at 0.36 m/s both with and without manual assistance.

Discussion

The purpose of this study was to determine how manual assistance affected lower limb electromyographic activity and joint kinematics in higher-level subjects with incomplete spinal cord injury during body weight supported treadmill training. We found that muscle activation amplitudes and patterns generally did not change when subjects with spinal cord injury were given manual assistance. Although we expected altered joint excursions with manual assistance, only small changes occurred. There was a small increase in knee joint excursion with manual assistance during swing phase of gait, but this was accompanied by small decreases in hip and ankle range of motion during stance phase. These changes in the joint range of motion excursions were likely due to the facilitation provided by the trainers during manual assistance. Variability of the kinematic profile at the ankle joint decreased when subjects with spinal cord injury were given manual assistance. We also found significant increases in EMG amplitudes and joint excursions with higher walking speeds. The shape of muscle activation patterns in subjects with spinal cord injury also tended to become less similar to controls at faster speeds, especially when walking without manual assistance.

We observed some differences between EMG profiles of control subjects and SCI subjects (Figures 2.1, 2.2, and 2.3). Interpretation of EMG voltages across subjects is generally limited for reasons such as skin impedance, subcutaneous fat thickness, muscle morphology, and electrode placement (Hogrel 2005). Despite this, it is still worthwhile to note some general differences in EMG voltages between control subjects and subjects with spinal cord injury.

The subjects with spinal cord injury adapted to higher speeds differently than the control subjects. At the slowest speed, EMG voltages in the thigh muscles and TA were generally greater in subjects with spinal cord injury than in control subjects (Figure 2.1). Plantar flexor activation amplitudes were comparable between control subjects and subjects with spinal cord injury at the slowest speed. With faster walking speeds, electromyographic activity in the thigh muscles and TA increased in subjects with spinal cord injury but remained about the same in control subjects (Figure 2.2 and 2.3). The most noticeable EMG amplitude difference with speed between SCI and control subjects was in the plantar flexors. Plantar flexor activation greatly increased in control subjects at faster speeds, but there was only a small increase in subjects with spinal cord injury.

There were concurrent changes in kinematics with increasing speed. Ankle plantar flexion increased at terminal stance phase with higher speed in control subjects, but there was less of an increase in this joint angle with speed in the subjects with spinal cord injury. Full knee extension was not achieved by subjects with SCI, and they also tended to be more flexed at the hip than control subjects throughout the gait cycle. These differences in EMG activity and kinematics between control subjects and subjects with spinal cord injury suggest that there are inherent differences in strategies for walking. Because subjects with spinal cord injury have motor deficits, spasticity, and sensory impairments, they must use different patterns of muscle activation and kinematics to accomplish the same functional movements (Grasso, Ivanenko et al. 2004).

The difference in adaptation to walking at faster speeds by the control subjects and subjects with spinal cord injury is of importance. The control subjects increased ankle plantar flexor muscle activity at terminal stance to increase their walking speed (Figure 2.3). The subjects with spinal cord injury lacked this increase in plantar flexor EMG activity. Normally, the ankle joint contributes more mechanical work during walking than the hip or knee (Meinders, Gitter et al. 1998). Instead, it appeared that the subjects with spinal cord injury compensated for the lack of ankle power by increasing muscle activity in the hip flexors. This may explain the high net cost of gait in individuals with spinal cord injury (Waters and Lunsford 1985). In addition, the inadequacy of ankle push off in terminal stance may prevent patients with spinal cord injury from achieving higher walking speeds (Pepin, Norman et al. 2003). This suggests that providing powered assistance at the ankle joint may be very important when designing robotic devices for rehabilitation (Sawicki, Domingo et al. 2006).

Our findings suggest that manual assistance may help to keep muscle activation patterns more similar to the pattern of control subjects during faster walking speeds. The shape of muscle activation patterns in the subjects with spinal cord injury became less similar to the control patterns at faster speeds, especially when walking without assistance. This is in agreement with previous research that showed walking at fast speeds may be an important part of gait rehabilitation programs in persons with spinal cord injury. Beres-Jones et al. found that faster stepping speeds increase afferent input and efferent activity during walking in individuals with spinal cord injury (Beres-Jones and Harkema 2004). Other studies indicated that step training at faster treadmill speeds is more effective at increasing over ground walking speed than step training at slower treadmill speeds in patients with stroke (Pohl, Mehrholz et al. 2002; Sullivan, Knowlton et al. 2002). Manual assistance may be beneficial because it allows persons with spinal cord injury to more safely achieve higher walking speeds. Half the subjects with spinal cord injury in this study could walk at faster speeds with manual assistance than without (Table 2.1).

There are potential limitations to this study. One limitation to this study was the small number of subjects we tested. The small number of subjects is not a major factor in our outcomes. We found significant differences in several variables. For many of the variables we did not find significant differences between conditions (SO and VL EMG amplitudes during the stance phase, MH EMG amplitude during the swing phase, R-values for TA, SO, LG, VM, VL, and ankle joint profiles, and the time shift for SO EMG profile), power analyses showed that testing more subjects would not likely change the results. Another variable of this study to consider is the ability of the trainers to administer manual assistance. EMG activity and kinematics could vary depending on the ability and experience of the trainers, and how much assistance the trainers give the subjects. In our case, the trainers were under the direct supervision of someone who was trained at a leading center in body weight supported treadmill training (UCLA Department of Neurology). Manual assistance should only provide enough assistance to facilitate normative walking kinematics and not completely overpower the efforts of the patient (Wernig 2005). Therefore, it is likely more assistance was needed and given at higher walking speeds than at slower speeds. When measurement devices are available to quantify the amount of assistance given without altering the manner in which the assistance should be given, this variable may be included in the statistical analysis. Lastly, subjects with spinal cord injury may adapt to walking on the treadmill with manual assistance over time, which may result in different muscle activation patterns and amplitudes (Pearson 2000). This is likely to happen if their walking ability improves with training, as it has been shown in previous studies (Wernig and Muller 1992; Dietz, Colombo et al. 1995; Wernig, Muller et al. 1995; Dietz, Wirz et al. 1998; Wernig, Nanassy et al. 1998; Behrman and Harkema 2000). A training study will be necessary to determine the effects of long-term motor adaptations.

Other future studies should involve testing subjects with different levels of impairment or with different neurological injuries since body weight supported treadmill training is used as treatment for a wide range of patients. All of our subjects were classified on the ASIA Impairment Scale as C or D and most of them were community ambulators. This was a necessary part of the study because the design required that the subjects have some walking ability in order to compare walking with and without manual assistance. However, results may be different for subjects with spinal cord injury that have more or less functional impairments than the ones in our study. Patients with neurological conditions other than spinal cord injury, such as stroke, Parkinson's Disease, or cerebral palsy, should also be tested.

Conclusions

We predicted that EMG activity and joint kinematics would change with manual assistance. The overall result, however, is that EMG amplitudes change little with manual assistance for relatively higher functioning spinal cord injury subjects. There were small but significant differences in joint range of motion with manual assistance. Providing manual assistance is not a detrimental part of body weight supported treadmill training and it allows subjects with spinal cord injury to walk at faster speeds than they could without assistance. In addition, manual assistance helps to keep the muscle activation patterns more similar to control data when walking at higher speeds.

Acknowledgments

The authors would like thank the subjects that participated in our studies, the University of Michigan PM&R staff for screening and recruitment of subjects with spinal cord injury, and the Human Neuromechanics Laboratory members for their assistance with data collection and processing. * A similar version of this chapter is published as: Domingo A, Sawicki GS, and Ferris DP. Kinematics and muscle activity of individuals with incomplete spinal cord injury during treadmill stepping with and without manual assistance. *J Neuroeng Rehabil* 4: 32, 2007.

Figures

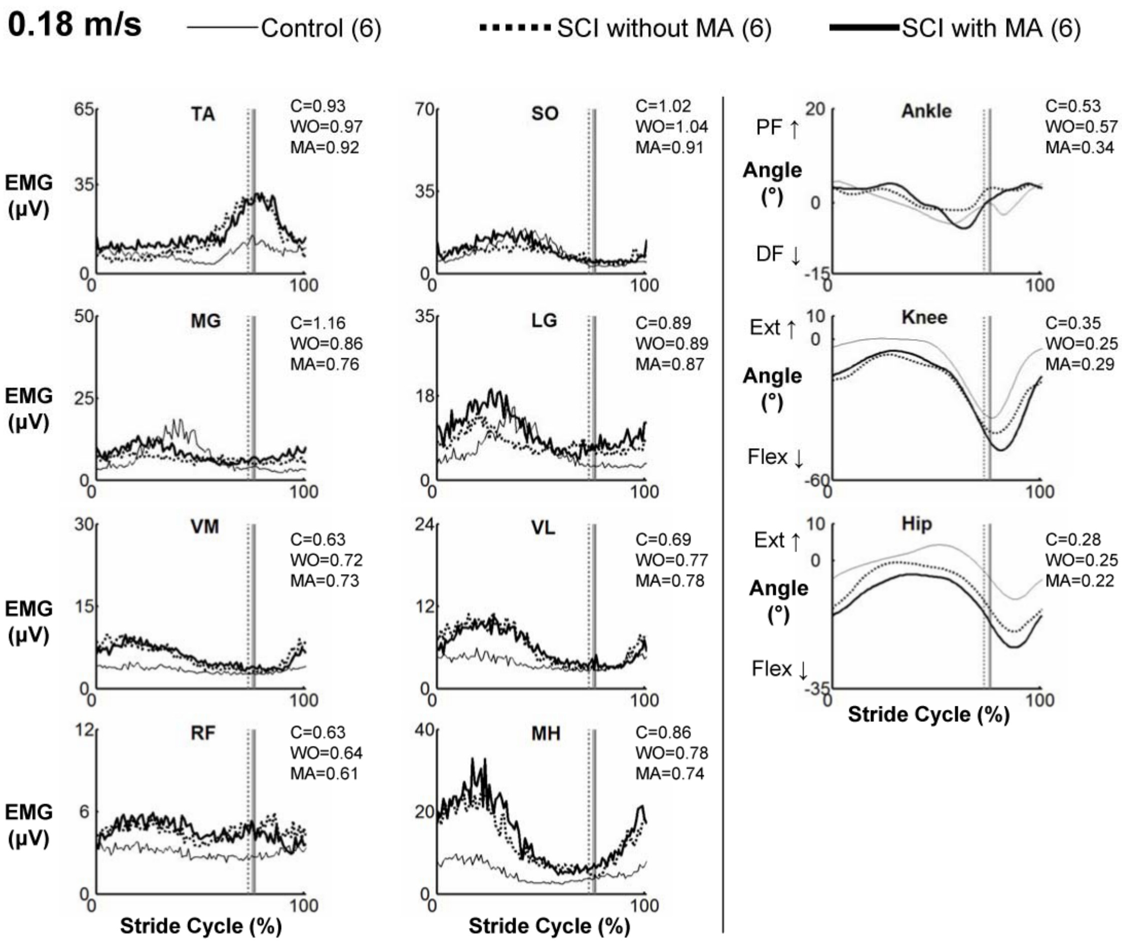

Figure 2.1 EMG profiles for subjects with spinal cord injury walking with (MA) and without (WO) manual assistance and control (C) subjects at 0.18 m/s
Averaged EMG profiles for tibialis anterior (TA), soleus (SO), medial gastrocnemius (MG), lateral gastrocnemius (LG), vastus medialis (VM), vastus lateralis (VL), rectus femoris (RF), and medial hamstring (MH) and averaged kinematic profiles for the ankle, hip and knee. Averages are taken from six subjects with spinal cord injury and six neurologically intact controls. Data from each subject were averaged over several step cycles within a trial, then over two trials of the same condition and speed, and finally averaged across subjects for the same condition and speed. Stride cycles were normalized from heel strike (0%) to heel strike of the same foot (100%). Vertical lines indicate the beginning of swing phase. The average coefficient of variation across subjects over the stride cycle is reported to the right of each plot.

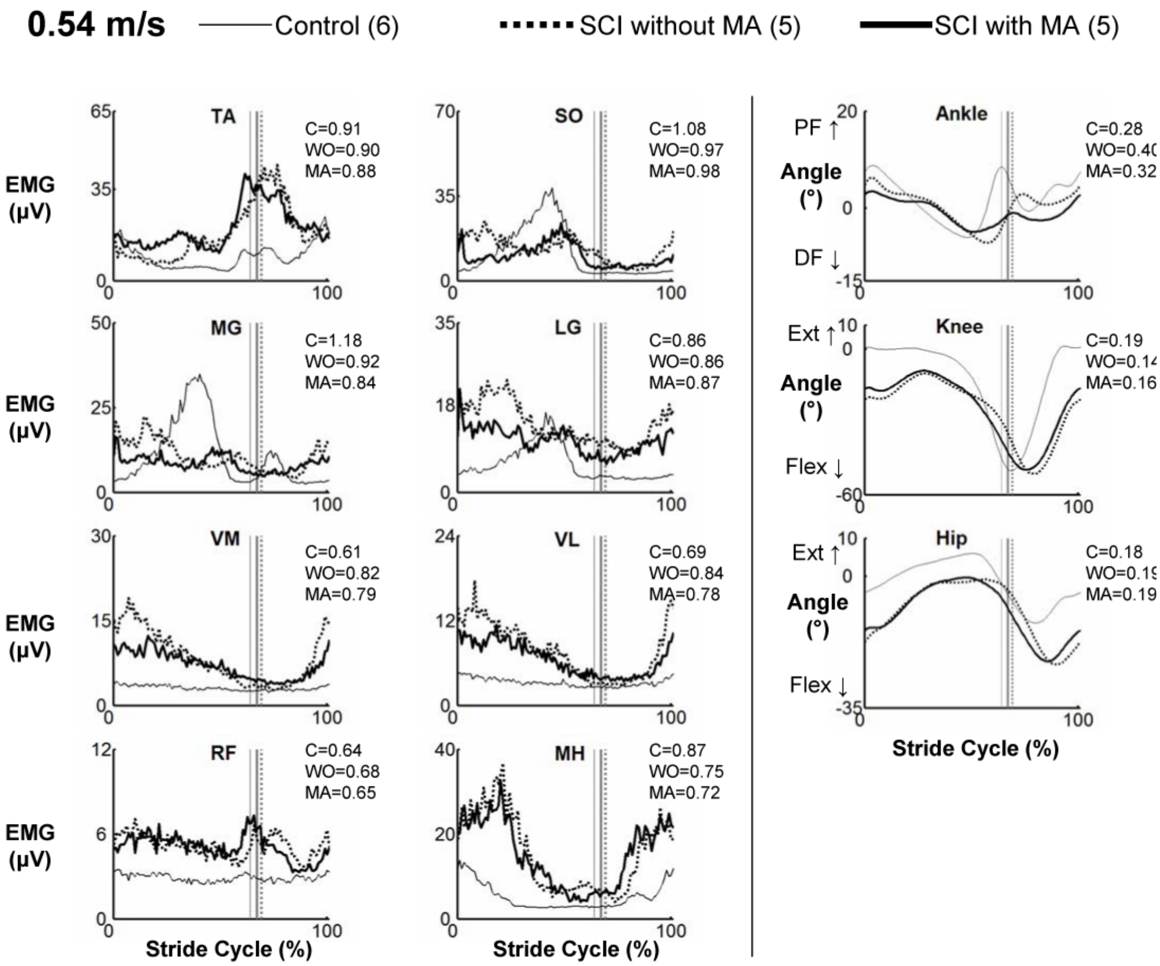

Figure 2.2 EMG profiles for subjects with spinal cord injury walking with (MA) and without (WO) manual assistance and control (C) subjects at 0.54 m/s

Averaged EMG profiles for tibialis anterior (TA), soleus (SO), medial gastrocnemius (MG), lateral gastrocnemius (LG), vastus medialis (VM), vastus lateralis (VL), rectus femoris (RF), and medial hamstring (MH) and averaged kinematic profiles for the ankle, hip and knee. Averages are taken from five subjects with spinal cord injury and six neurologically intact controls. Stride cycles were normalized from heel strike (0%) to heel strike of the same foot (100%). The average coefficient of variation across subjects over the stride cycle is reported to the right of each plot.

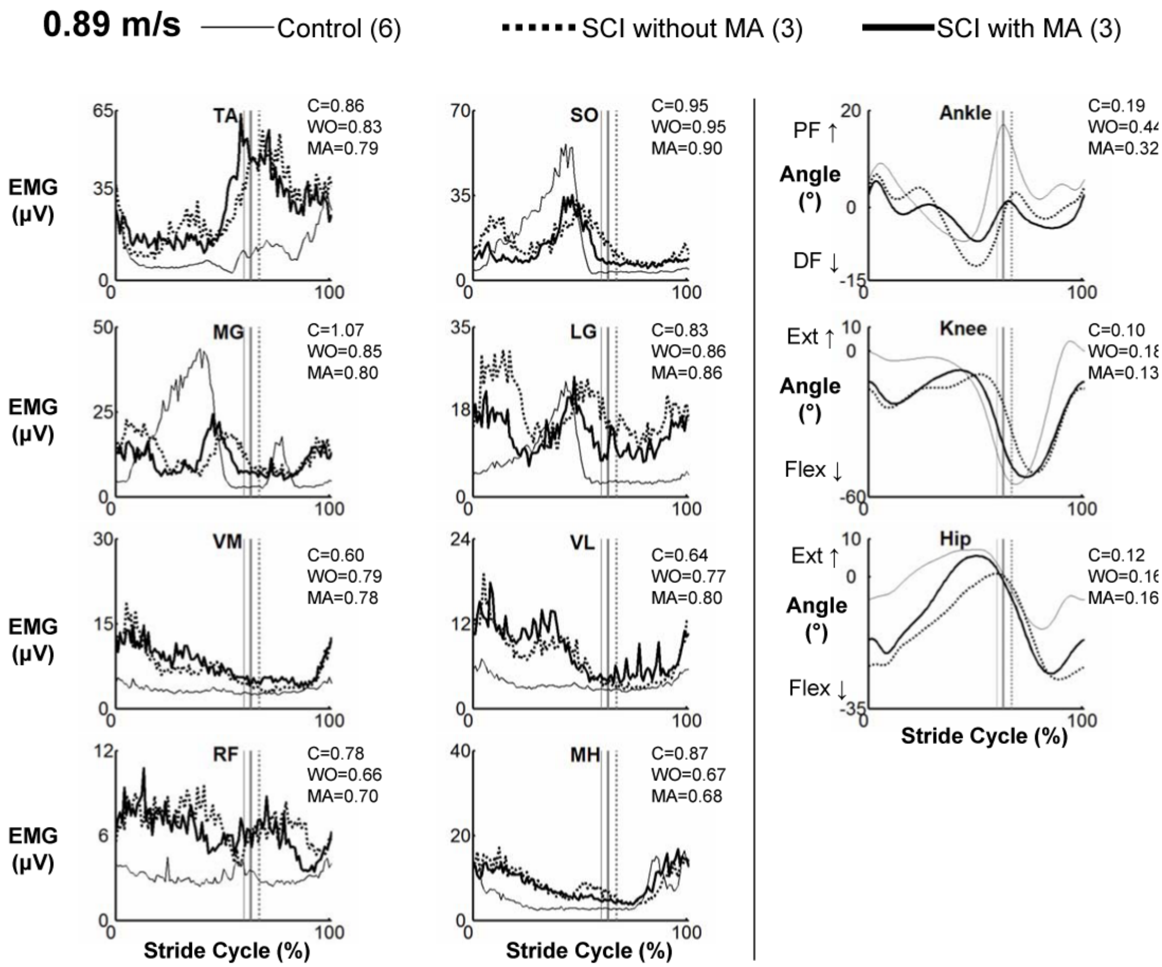

Figure 2.3 EMG profiles for subjects with spinal cord injury walking with (MA) and without (WO) manual assistance and control (C) subjects at 0.89 m/s

Averaged EMG profiles for tibialis anterior (TA), soleus (SO), medial gastrocnemius (MG), lateral gastrocnemius (LG), vastus medialis (VM), vastus lateralis (VL), rectus femoris (RF), and medial hamstring (MH) and averaged kinematic profiles for the ankle, hip and knee. Averages are taken from three subjects with spinal cord injury and six healthy controls. Stride cycles were normalized from heel strike (0%) to heel strike of the same foot (100%). The average coefficient of variation across subjects over the stride cycle is reported to the right of each plot.

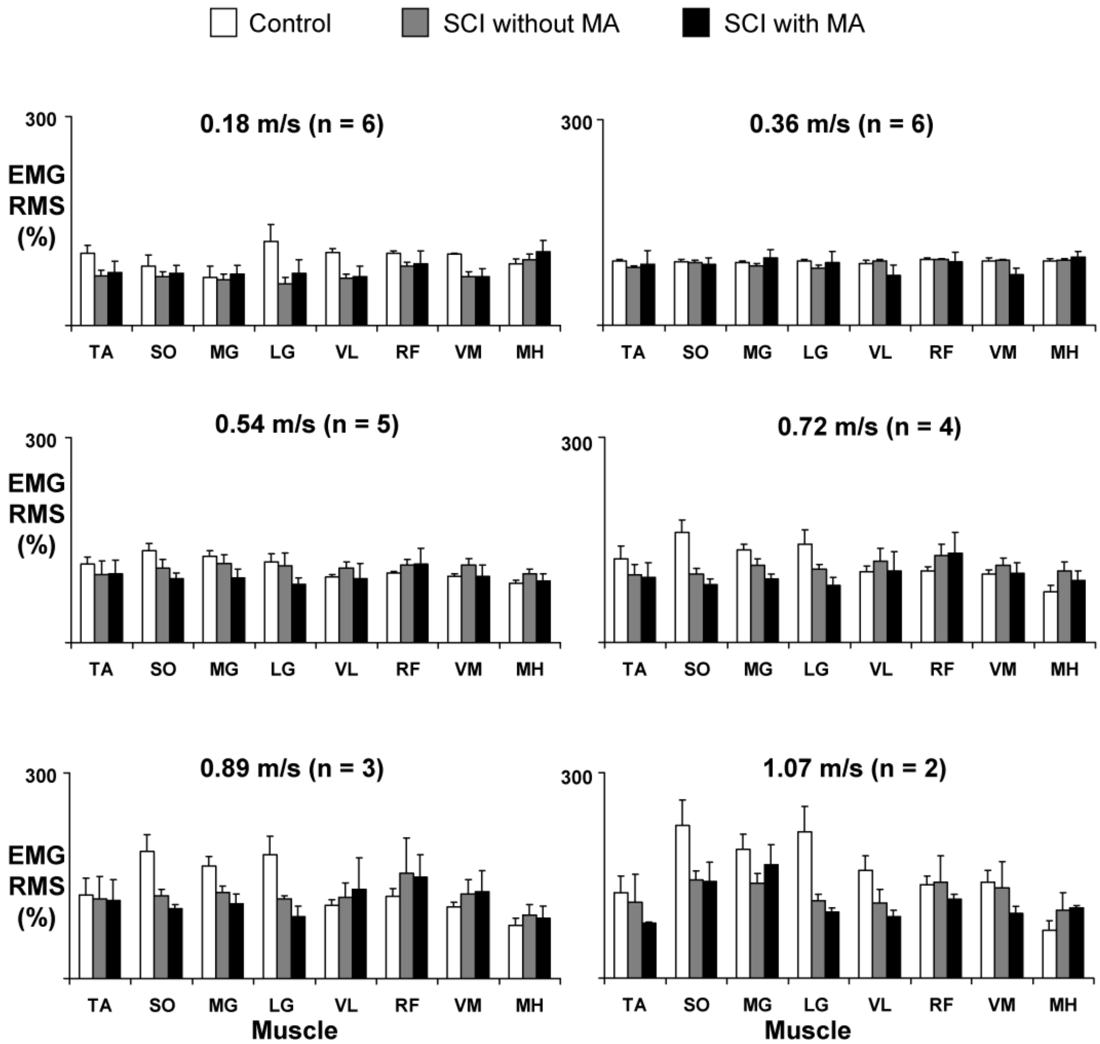

Figure 2.4 Stance phase EMG RMS for subjects with spinal cord injury walking with and without manual assistance and control subjects at six different speeds

Averaged normalized muscle activation amplitudes for tibialis anterior (TA), soleus (SO), medial gastrocnemius (MG), lateral gastrocnemius (LG), vastus medialis (VM), vastus lateralis (VL), rectus femoris (RF), and medial hamstring (MH) for the specified number of subjects with spinal cord injury and six control subjects. RMS data for each muscle were first normalized to the highest average RMS value that occurred among two trials at 0.36 m/s. These normalized values from each muscle were then averaged over two trials of the same condition and speed within a subject, and finally averaged across subjects for the same condition and speed. Bars indicate mean ± standard error. There were no significant differences in muscle activation amplitudes when walking with or without manual assistance (ANOVA, $p > 0.05$).

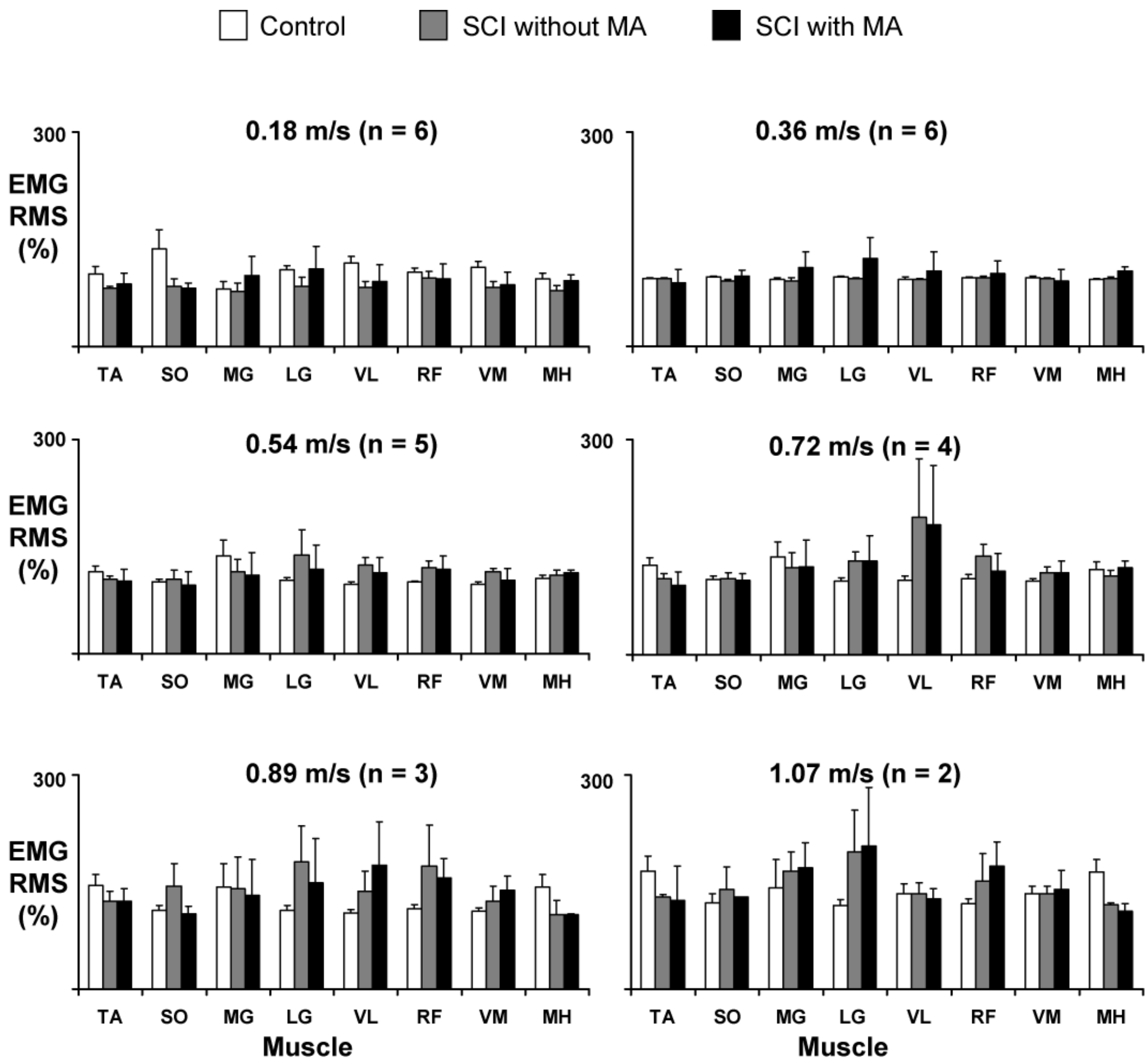

Figure 2.5 Swing phase EMG RMS for subjects with spinal cord injury walking with and without manual assistance and control subjects at six different speeds

Averaged normalized muscle activation amplitudes for tibialis anterior (TA), soleus (SO), medial gastrocnemius (MG), lateral gastrocnemius (LG), vastus medialis (VM), vastus lateralis (VL), rectus femoris (RF), and medial hamstring (MH) for the specified number of subjects with spinal cord injury and 6 control subjects. Bars indicate mean ± standard error. There were no significant differences in muscle activation amplitudes when walking with or without manual assistance (ANOVA, $p > 0.05$).

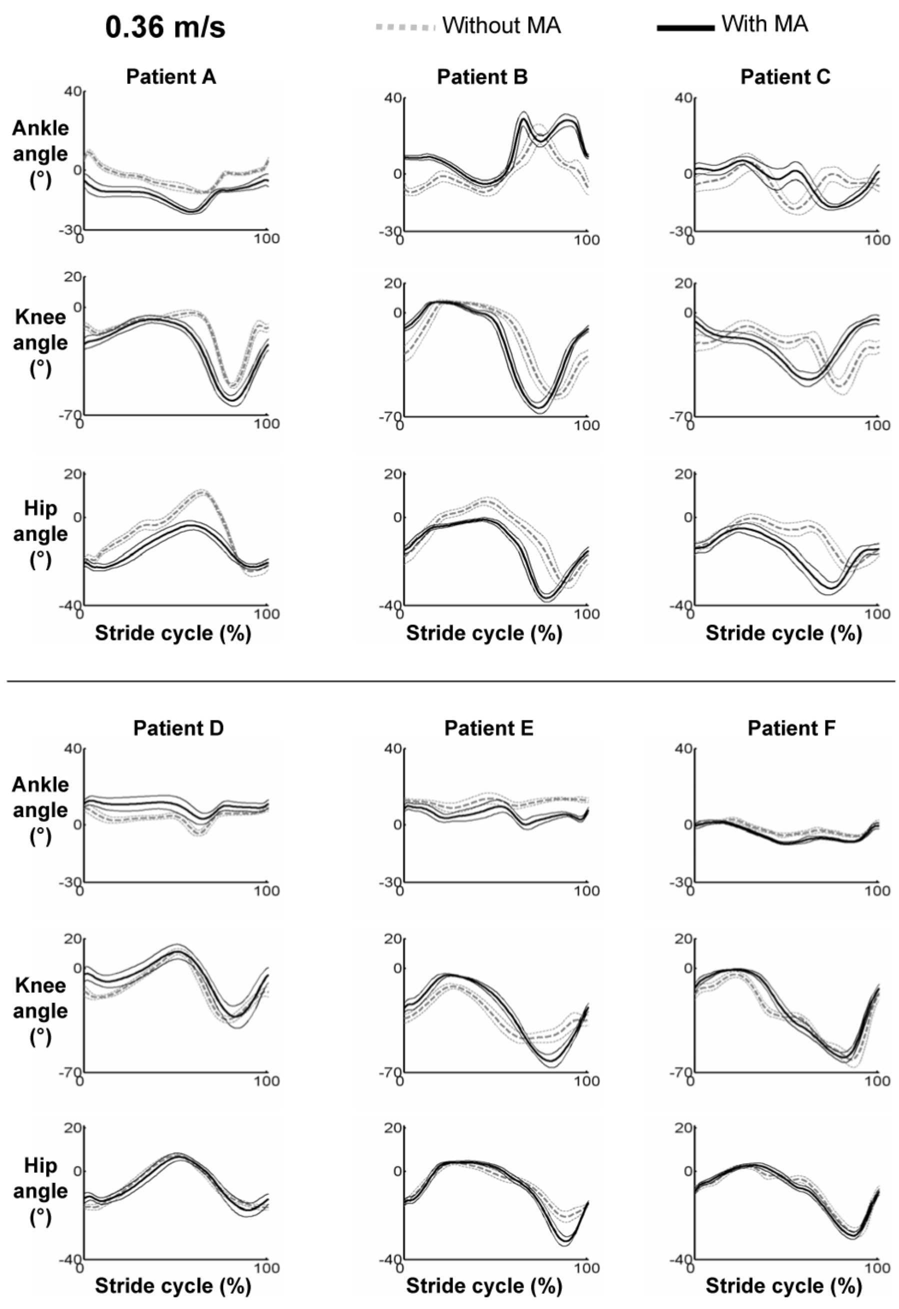

Figure 2.6 Kinematic variability in subjects with spinal cord injury
Figures show joint angle data (heavy line) ± 1 standard deviation (thin lines) for the six different subjects with spinal cord injury walking at 0.36 m/s. Variability increases in some subjects and decreases in others when given manual assistance. Only the ankle joint showed significantly lower variability with subjects were walking with manual assistance.

Tables

Table 2.1. Subject Information.

Data for each subject showing age, body size, injury level, walking ability, body weight support level and walking speeds completed during the study.

Subject	Age (yrs.)	Sex Height Weight	Injury Etiology	Injury Level	ASIA* Level	Post Injury (mos.)	Walking Aids	Overground Walking Speed (m/s)	BWS Level (%) Speeds w/o MA (m/s) Speeds w/ MA (m/s)
A	54	F 165.1 cm 73.7 kg	Dermoid Tumor	T11/T12	C	64	Cane (L,R) Ankle-foot orthosis (L)	0.41	30% 0.18-0.89 0.18-0.89
B	52	F 156.2 cm 58.1 kg	Myxopapillary Ependymoma	T8/L2	D	93	Quad Cane (R)	0.61	30% 0.18-0.36 0.18-0.72
C	38	F 175.3 cm 115.3 kg	Transverse Myelitis	T5	D	77	Cane (R) Ankle-foot orthosis (L)	0.37	50% 0.18-1.07 0.18-1.07
D	24	M 185.4 cm 101.5 kg	Trauma	T10/T11	D	111	–	0.95	30% 0.18-1.07 0.18-1.07
E	55	M 171.5 cm 83.0 kg	Sarcoidosis	C5/C6	C	144	Cane (R)	0.48	60% 0.18-0.54 0.18-0.89
F	50	M 193.0 cm 95.3 kg	Trauma	C4/C5	C	83	Wheelchair Soft ankle brace (L,R)	–	50% 0.18-0.72 0.18-1.07

* ASIA = American Spinal Injury Association Impairment Scale.

A = Complete, E = Normal.

Table 2.2. Cross-correlation analyses of EMG and kinematic profiles.
Values shown are the results of cross correlation analyses comparing data for all speeds and conditions between: spinal cord injury subjects walking without manual assistance and control subject data (WO-Control), spinal cord injury subjects walking with manual assistance and control subject data (MA-control), and spinal cord injury subjects walking without manual assistance and with manual assistance (WO-MA). Waveforms and profiles were normalized to the percentage of the gait cycle and therefore the resulting shifts from the analyses are given in percentages. Statistical analyses were then performed (repeated measure ANOVAs) to find significant differences between R-values and time shifts.

		R-value	shift (%)			R-value	shift (%)
TA EMG	WO-Control	0.81	7	RF EMG	WO-Control	0.93	0
	MA-Control	0.82	5		MA-Control	0.93	0
	WO-MA	0.91*†	0*†		WO-MA	0.94	0
SO EMG	WO-Control	0.82	5	MH EMG	WO-Control	0.87	0
	MA-Control	0.84	2		MA-Control	0.86	0
	WO-MA	0.89*†	1		WO-MA	0.95*†	0
MG EMG	WO-Control	0.80	3	Ankle angle	WO-Control	0.47	-16
	MA-Control	0.80	2		MA-Control	0.37	-8
	WO-MA	0.90*†	0		WO-MA	0.77*†	2
LG EMG	WO-Control	0.83	3	Knee angle	WO-Control	0.87	-8
	MA-Control	0.85	-3		MA-Control	0.91*	-5*
	WO-MA	0.89*†	0		WO-MA	0.96*†	2*†
VM EMG	WO-Control	0.91	0	Hip angle	WO-Control	0.77	3
	MA-Control	0.92	0		MA-Control	0.78	4
	WO-MA	0.93*	0		WO-MA	0.92*†	1
VL EMG	WO-Control	0.91	0				
	MA-Control	0.93*	0				
	WO-MA	0.93	0				

*Indicates significantly different from WO-Control ($p<0.05$)

†Indicates significantly different from MA-Control ($p<0.05$)

Table 2.3. Joint excursions in subjects with spinal cord injury.
Average joint excursion for all subjects with spinal cord injury at all possible speeds while walking with or without manual assistance. Data were averaged separately for the stance and swing phase.

Joint	Without Manual Assistance (º)	With Manual Assistance (º)
Ankle		
Stance	18.8	15.8*
Swing	13.5	14.7
Knee		
Stance	27.9	28.9
Swing	36.1	41.4*
Hip		
Stance	22.3	19.3*
Swing	23.7	22.5

*Indicates significantly different than without manual assistance condition ($p<0.05$)

Chapter 3. Effects of physical guidance on short term learning of walking on a narrow beam

Abstract

Physical guidance is often used in rehabilitation when teaching patients to re-learn movements. However, the effects of guidance on motor learning of complex skills, such as walking balance, are not clear. We tested four groups of healthy subjects that practiced walking on a narrow (1.27 cm) or wide (2.5 cm) treadmill-mounted balance beam, with (Assisted) or without (Unassisted) physical guidance. Assistance was provided by springs attached to a hip belt that applied restoring forces towards beam center. All subjects were evaluated while walking *unassisted* before and after training by calculating the number of times subjects stepped off of the beam per minute of successful walking on the beam (Failures per Minute). Subjects in Unassisted groups had 49.0±4.6% (mean±SEM) less Failures per Minute after training compared to before training, while those in Assisted groups had 2.88±11.6% more after training (ANOVA: P=0.0002). In contrast, during the training period, Unassisted groups had more Failures per Minute (16.3±1.91) than Assisted groups (6.35±1.56; ANOVA: P<0.0001). Task difficulty affected the relationship between physical guidance and learning the

task. Normalized performance gains were relatively smaller in Narrow Beam groups than in Wide Beam groups but the interaction effect was not significant (P=0.071). The Unassisted-Wide and Assisted-Narrow groups had similar Failures per Minute during training (12.26±1.30 and 12.06±2.59, respectively, t-test: P=0.9158), but the Unassisted-Wide group had much greater performance gains after training (-61.2±6.02% and -7.66±7.3%, respectively, P<0.0001). These results imply that task specificity during practice can have substantial effects on short-term motor learning independent of error experience.

Introduction

Physical guidance, or force intended to reduce movement error, is often used during the rehabilitation of walking. Physical guidance may be given to a patient for a variety of reasons: to increase safety, to reduce fear, or to help complete a task that a patient may not otherwise be able to perform on their own. However, little is known about how using assistance affects motor learning of complex tasks such as walking balance. In the elderly, balance is commonly compromised, and most falls occur during walking, not standing (Blake, Morgan et al. 1988; Niino, Tsuzuku et al. 2000). For this reason, it is important to understand how assistance affects learning of walking balance. With this understanding, more effective treatments can be designed for gait rehabilitation.

Studies on the effects of physical guidance on motor learning have varied results. Physical guidance is not helpful for learning simple movements in the upper extremity (Armstrong 1970). Guidance improved performance during practice trials but performance improvements were not present when the guidance was removed. One possible explanation is that physical guidance did not allow for error detection and correction. Error is a critical stimulus for driving motor learning (Rumelhart, Hinton et al. 1986; Lisberger 1988; Dancause, Ptito et al. 2002). Another recent study examined a more complex movement and found slightly different results (Liu, Cramer et al. 2006). Subjects learned to trace a complex three-dimensional trajectory with the upper extremity using either robotic assistance or visual demonstration. The group that practiced with robotic assistance improved in performance but not any better than the group that used visual guidance alone (Liu, Cramer et al. 2006). In a task where subjects learned to bear weight on their legs asymmetrically, manual guidance provided no help (Sidaway, Ahn et al. 2008).

In a more complex whole-body task (learning to use a ski simulator), subjects performed movements better when they practiced with ski poles for stabilizing guidance than without them (Wulf, Shea et al. 1998). The ski poles allowed the subjects to select the magnitude and timing of the assistive forces while maintaining focus on the task dynamics. Body-weight supported treadmill training, where patients are given manual assistance to move the lower extremities through the motions of walking, has been effective in helping subjects with neurological injury to re-learn how to walk (Dietz, Colombo et al. 1995; Hesse, Bertelt et al. 1995; Wernig, Nanassy et al. 1998; Behrman and Harkema

2000; Dobkin, Apple et al. 2006; Hornby, Campbell et al. 2008). However, none of these studies had control groups where subjects practiced without assistance. These studies suggest that assistance is detrimental to learning easier tasks but may be helpful for more difficult tasks. Subjects performing very difficult tasks may benefit from using assistance because too many errors would not give the subject an appropriate example of the actual task (Sanger 2004).

The purpose of this study was to provide insight on the effects of physical guidance on short-term learning of walking balance and to explore if task difficulty alters those effects. We chose to study healthy subjects learning to walk on a narrow balance beam. Beam walking is similar to over ground walking, but is more challenging to dynamic balance because it exploits the lateral instability of walking (Donelan, Shipman et al. 2004; Schrager, Kelly et al. 2008). We tested two groups of subjects that practiced walking on a 2.5 cm-wide treadmill-mounted balance beam for thirty minutes, with or without lateral physical assistance at the hips. All subjects were evaluated on *unassisted* beam walking pre- and post-training. We hypothesized that subjects that received no assistance during training would have greater performance gains than subjects that received assistance. We based this hypothesis on the rationale that error drives motor learning (Rumelhart, Hinton et al. 1986; Lisberger 1988; Dancause, Ptito et al. 2002) and assistance tends to reduce errors. To explore the confounding effects of task difficulty on the relationship between physical assistance and learning balance, we tested two more groups of subjects (with and without assistance) on a narrower balance beam (1.27 cm-wide). We hypothesized that the difference between performance gains in the unassisted and assisted group would be smaller for the more difficult task (narrow beam) than the easier task (wide beam). This was based on the idea that if a task is too difficult, assistance would be helpful in producing examples of the desired task.

Methods

Subjects

We tested 40 neurologically intact subjects (see Table 3.1 for subject characteristics). Subjects were medically stable and had no history of major leg injury. The University of Michigan Institutional Review Board approved this study (IRB#HUM00008186). All subjects gave informed consent according to the Declaration of Helsinki prior to participating.

Equipment

The equipment for this experiment consisted of a treadmill-mounted balance beam (beam-mill), a lateral assist device, force plates and a motion capture system. The beam-mill was made of interchangeable small wooden blocks attached to the treadmill belt that lined up to make a continuous balance beam (Figure 3.1). One beam was 2.5 cm wide by 2.5 cm tall (Wide Beam) and the other was 1.27 cm wide by 2.5 cm tall (Narrow Beam). Smaller wooden blocks were added to either side of the bases of both beams to make them more stable in the frontal plane.

Table 3.1. Subject characteristics.

Group	Gender		Body mass (kg±SD)	Leg length (m±SD)
	M	F		
Unassisted-Narrow	4	6	63.7±9.7	0.90±0.056
Assisted-Narrow	6	4	64.7±8.9	0.91±0.053
Unassisted-Wide	6	4	64.6±14.7	0.89±0.063
Assisted-Wide	5	5	64.1±13.2	0.89±0.045

The lateral assist device was made up of latex tubing and cables that attached to the subject via a padded hip belt. We chose this form of assistance because we could control the amount of assistance that provided lateral stabilization. We provided stabilization in the frontal plane because walking is passively stable in the anterior-posterior direction but unstable in the medio-lateral direction (Bauby and Kuo 2000). This form of stabilization has been used in other studies (Donelan, Shipman et al. 2004; Chang and Ulrich 2008) during treadmill walking. A similar device has also been used in clinical settings to stabilize the torso during bodyweight supported treadmill training (Behrman and Harkema 2000). The springs were stretched and placed laterally so that they provided a restoring force towards the center of the beam. When the subject's pelvis was centered over the beam, zero net force was applied to the subject. We had 4 springs of different stiffnesses. For each subject, we chose the spring that would provide the stiffness closest to the non-dimensionalized spring stiffness of 0.228. To determine the desired spring stiffness, we used the following equation:

$$k = \bar{k} \cdot \frac{l}{mg}$$

where $k =$ dimensionalized stiffness, $\bar{k} =$ non-dimensionalized stiffness, $l =$ leg length and $mg =$ bodyweight. The non-dimensionalized spring stiffness of 0.228 was based on springs used during pilot testing. These springs gave subjects feedback about their position relative to the beam but did not give them so much assistance that it completely prevented them from stepping off the beam. The average stiffness of the springs used was 160.96 N/m. We placed single-axis

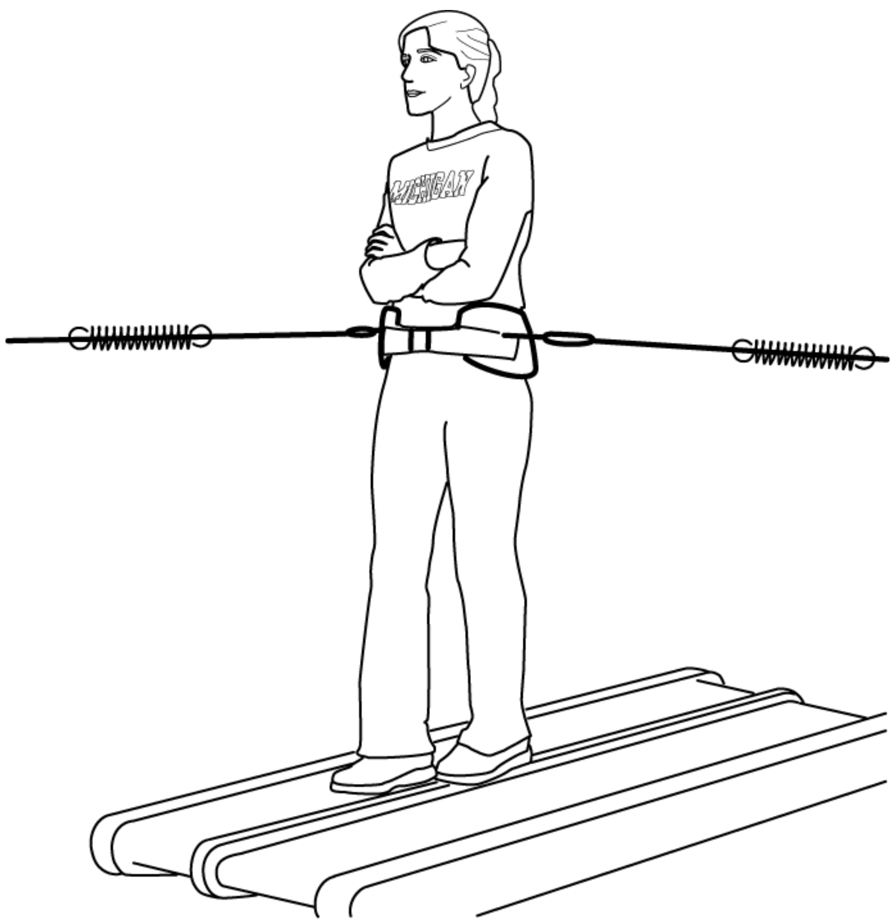

Figure 3.1. Experimental Setup.
A subject walking on the treadmill-mounted balance beam with the lateral assist device. The lateral assist device provided a restoring force towards the center of the beam.

tension/compression load cells (1200 Hz; Omega Engineering, Stamford, CT, USA) in series with the springs on both sides of the subject to measure the force produced by the springs during walking. The lateral assist device provided an average net force of <3.0% of body weight onto the subject during the training period while walking on the beam.

The treadmill was placed above two force plates (sampling rate 1200 Hz; Advanced Mechanical Technology Inc., Watertown, MA, USA) so that we could calculate center of pressure from the forces and moments produced by the subject while walking (Collins, Adamczyk et al. 2009).

We used a 4-camera video system (frame rate 120 Hz; Motion Analysis Corporation, Santa Rosa, CA, USA) to record the positions of 4 reflective markers placed on the subject's pelvis, neck and shoulders during walking.

Procedures

Four groups of 10 subjects walked on the beam-mill for a 3-minute pre-training evaluation, a 30-minute training period, and a 3-minute post-training evaluation. Two groups walked on the Wide Beam and two other groups walked on the Narrow Beam. Treadmill speed was set at 0.22 m/s. This speed was chosen based on pilot experiments. Subjects were instructed to walk on the beam for as long as possible without stepping off. Instructions were given to all subjects by the same experimenter. They had to walk heel-to-toe with arms crossed over their torso. They were instructed not to lean forward, twist their trunk, angle their feet away from the longitudinal direction of the beam, or look down. Subjects wore goggles that obscured view of the walking surface. Subjects were allowed to move their pelvis and trunk laterally to help maintain balance. All subjects wore standardized orthopedic shoes. Subjects had to wait five seconds after stepping off the beam before attempting to walk on it again.

During the training periods, one of the Narrow Beam groups and one of the Wide Beam groups were given assistance via the lateral assist device (Assisted-Narrow, Assisted-Wide), and the other two groups were not given any assistance (Unassisted-Narrow, Unassisted-Wide). The training duration was 30 minutes with rest breaks every 10 minutes. During the pre- and post-evaluation periods, all subjects walked without assistance and were made aware of this at the beginning of the experiment.

We recorded the number of times the subject stepped off the beam per minute. We then divided this quantity by the fraction of time the subject was on the beam (not touching the treadmill surface with either foot). This quotient, Failures per Minute, was our primary performance metric because it took into account the number of errors with respect to the amount of time the subject was successfully able to walk on the beam, both indicators of learning and performance. We also calculated the standard deviation (SD) of the medio-lateral movement of markers placed at the sacrum and the neck (Motion Analysis Corporation, Santa Rosa, CA; 120 Hz) as a measure of movement variability at the upper trunk and pelvis.

We calculated percent change of the performance variables by subtracting the pre-training value from the post-training value and dividing by the pre-training value for each subject to normalize to pre-training performance.

For the pre- and post-training periods, we recorded data for the duration of the 3-minute trial. For the 30-minute training period, we collected only 20 seconds of data per each minute of training. We used a 4th order, zero-lag low pass Butterworth filter with a cutoff frequency of 6 Hz to smooth center of pressure data. Values for SD of markers were calculated only using the data from when subjects were on the beam.

Statistical Analysis

To evaluate whether Narrow Beam walking was more difficult than Wide Beam walking, we performed a 2x2 Analysis of Variance (ANOVA) (assist, beam) to compare results for percentage time on the beam and number of failures between the Narrow Beam groups and the Wide beam groups during pre-training (JMP IN software, SAS Institute, Inc., Cary, NC). We also used this information to determine if both assisted groups and both unassisted groups had similar pre-training scores to each other.

We performed a 2x2 ANOVA (assist, beam, assist*beam) to test for differences between the groups and any interaction effect for each of the following dependent variables: percent change for Failures per Minute, sacral marker SD and neck marker SD. For post-hoc analysis, we performed t-tests to compare

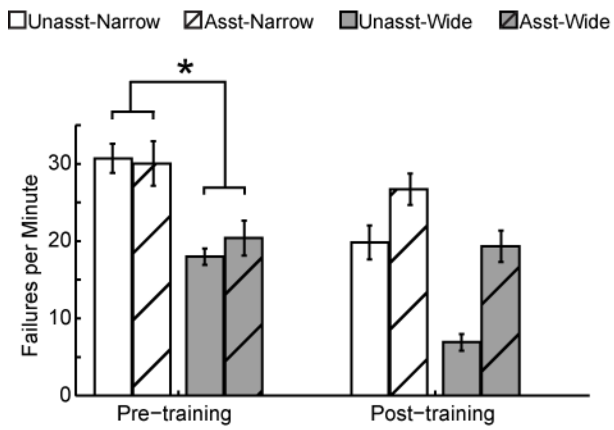

Figure 3.2. Averaged pre- and post-training values for Failures per Minute across subjects in all groups.
Error bars are ±1 standard error of the mean (SEM). Significant differences are indicated by * (ANOVA, $P < 0.05$) Statistical analyses of post-training data not presented here.

results within each beam group as needed to delineate the differences between assist groups, and adjusted the alpha level for multiple comparisons using the Bonferroni correction (α=0.05/number of tests).

We used generalized estimating equations (GEE) to test for differences in the time series data between each group during training (SPSS software, SPSS, Inc., Chicago, IL). We also performed contrast tests using pairwise comparisons to delineate which groups were different from each other.

Results

Pre-training results showed that walking on the Narrow Beam was more difficult than walking on the Wide Beam. The Narrow Beam groups had significantly more Failures per Minute (30.4±1.7, mean±SEM) than the Wide Beam groups

(19.2±1.2) in pre-training (ANOVA, beam: P<0.0001, power>0.99) (Figure 3.2). Both Wide Beam groups had similar pre-training scores, as did both the Narrow Beam groups (ANOVA, assist: P=0.6871).

The assistance used during training greatly hindered learning of the unassisted task compared to those that did not use assistance (Figure 3.3A). The results

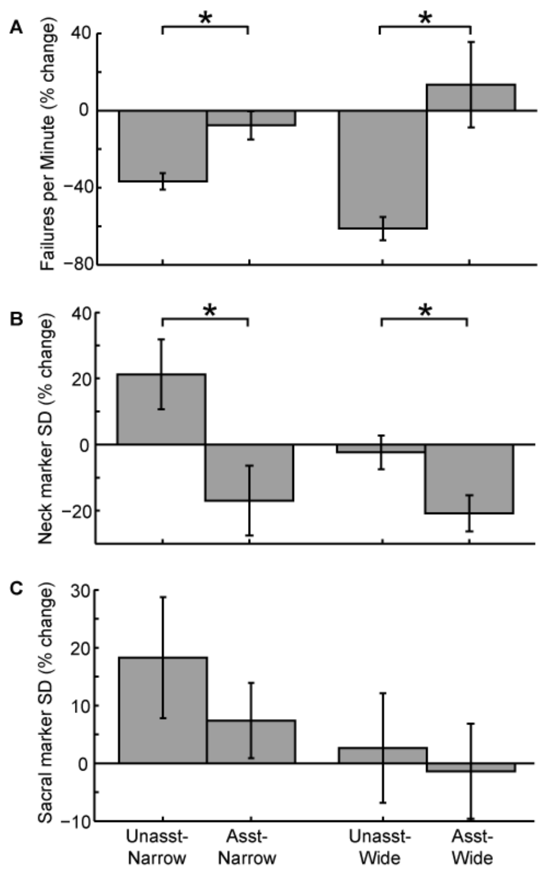

Figure 3.3. Percent change in Failures per Minute, Neck Marker SD, and Sacral Marker SD Error bars are ±1 SEM. Significant differences are indicated by * (t-test, P < 0.05). **A.** Bars represent averaged percent change across subjects in all groups for Failures per Minute. Normalized performance gains were relatively smaller in Narrow Beam groups than in Wide Beam groups but the interaction effect was not significant. **B.** Bars represent the averaged percent change across all subjects in all groups for the standard deviation of the neck marker in the frontal plane. **C.** Bars represent the averaged percent change across all subjects in all groups for the standard deviation of the sacral marker in the frontal plane.

showed that the Unassisted groups had 49.0±4.6% less Failures per Minute after training, and the Assisted groups had 2.88±11.6% more Failure per Minute after training (ANOVA, assist: P=0.0002). Post-hoc tests showed that the Unassisted-Wide group was different than the Assisted-Wide group (t-test: P=0.0045), and that the Unassisted-Narrow group was different than the Assisted-Narrow group (t-test: P=0.0030). Power for post-hoc tests were greater than 0.85. The interaction effect (assist*beam) approached significance (ANOVA: P=0.0712). The Assisted-Wide group had more failures after training (13.4% more failures from pre- to post-training).

Most subjects decreased frontal plane movement variability in the upper body and increased movement variability at the pelvis during post-training compared to pre-training. The percent change in standard deviation of neck marker movement in the medio-lateral direction was significantly different between groups (ANOVA, assist: P=0.0017) (Figure 3.3B). Post-hoc tests showed that the percent change in movement variability at the neck marker was significantly different between the Assisted-Wide and Unassisted-Wide groups (t-test: P=0.0235). The narrow beam groups were also significantly different from each other (t-test: P=0.0200). Because subjects in the assisted groups had little or no improvements after training, the decrease in neck marker movement variability after training in the assisted groups suggests that movement at the upper trunk was correlated with the ability to maintain balance during beam walking. There were no significant differences between groups for sacral marker movement (ANOVA: P=0.4355) (Figure 3.3C).

There were significant differences in the Failures per Minute during training between groups (ANOVA: P<0.0001), but post-hoc tests showed that that the Unassisted-Wide and Assisted Narrow were not significantly different than each other (t-test: P=0.9158) (Figure 3.4A). All other comparisons were significant (P<0.0083) except Assisted-Narrow compared to Assisted-Wide (P=0.0373). All significant findings for these comparisons had a power greater than 0.9. GEE analysis showed similar results when comparing the time series data during training (GEE: P<0.001) (Figure 3.4B). Pairwise comparisons showed that there were differences between all groups in Failures per Minute (P<0.05) except for the Unassisted-Wide and Assisted-Narrow groups (P=0.943).

Movement variability at the sacral marker in frontal plane during training for the different groups (Figure 3.4C) paralleled their respective improvements in performance (Figure 3.3A).

Discussion

Our main result was that practice with assistance hindered short-term learning of a walking balance task compared to unassisted practice. We also found that the effects of physical guidance on motor learning may depend on task difficulty. Using assistance during practice while walking on the narrow beam did not

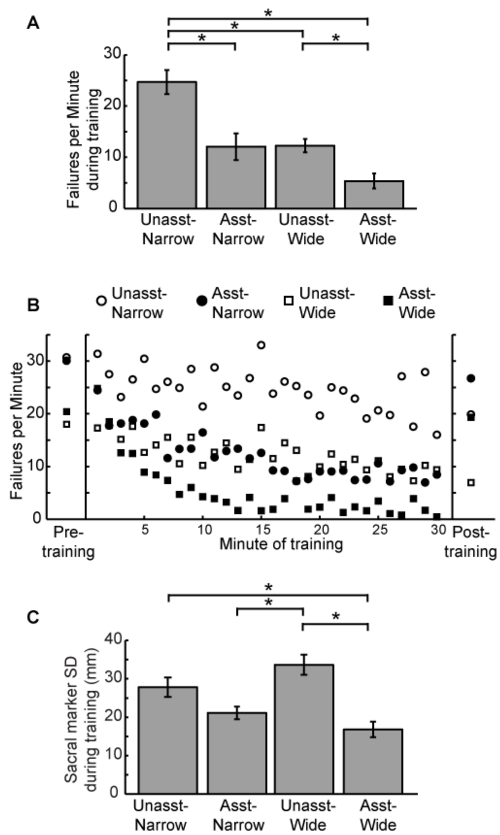

Figure 3.4. Failures per Minute and sacral marker SD during training
Error bars are ±1 SEM. Significant differences are indicated by * (ANOVA, P < 0.05) **A.** Bars represent averaged Failures per Minute of walking on the beam during training for each group. **B.** Time series data represent Failures per Minute during each minute of training. Pairwise comparisons showed that the Unassisted-Wide and Assisted-Narrow groups were the only groups that were not significantly different from each other. **C.** Bars represent averaged standard deviations of sacral marker movement in the frontal plane during training for each group.

hinder learning as much as while walking on the wide beam. Thus, assistance appears less detrimental when used during more difficult motor tasks. This is consistent with what is considered best practices in clinical rehabilitation: that assistance should only be given as much as is needed to complete the task (Ryerson and Levitt 1997). It is also consistent with the challenge point framework for motor learning that states that task difficulty should be adjusted to the learner’s skill level (Guadagnoli and Lee 2004).

Physical guidance was clearly detrimental to learning a relatively easy balancing task (walking on the wide beam). The Unassisted-Wide group had the largest percentage decrease in Failures per Minute for the post-test compared to the pre-test (Figure 3.3A). The performances in the post-test were in direct contrast to the performance during training. The greater amount of learning by the unassisted group could be attributed to experiencing more errors (Failures per Minute) during the training period (Figure 3.4A-B).

The results were different in groups that learned the more difficult task (walking on the narrow beam). There was a smaller difference between the performance gains after training for the assisted and unassisted groups for the narrow beam (Figure 3.3A), despite having relatively similar error experience during training as the wide beam groups (Figure 3.4A-B). The interaction effect of assist and beam approached significance. Thus, for more difficult tasks, physical assistance seems to be less detrimental to motor learning. If the task was even more difficult, it could be that physical assistance might actually be beneficial compared to no assistance.

It is important to dissociate the effects of the mechanical interactions of physical guidance and the error experienced during practice on motor learning. Error experience is proportional to motor learning (Thoroughman and Shadmehr 2000; Scheidt, Dingwell et al. 2001). Because physical guidance reduces errors, it would follow that physical guidance would hinder motor learning. However, there may be some effect of the mechanical interaction itself that may positively or negatively affect learning. To make this distinction, we examined the performance of two groups of subjects that experienced similar amounts or error during training, but one group had assistance (Assisted-Narrow) and the other did not (Unassisted-Wide).

Both the Unassisted-Wide and Assisted-Narrow groups experienced similar amounts of error during training (Figure 3.4A) but had different performance gains after training. The Unassisted-Wide group had a larger percent change in time on the beam and number of failures than the Assisted-Narrow group (Figure 3.3A). This suggests that another factor is important to motor learning of this task other than the amount of errors.

Another possible explanation for the motor learning gains between assisted and unassisted groups is that the training for the unassisted group had greater task specificity. When the subjects were provided assistance, subjects could have learned to rely on the restoring force as an inherent part of the task. According to the specificity of practice hypothesis, motor learning is specific to the available afferent feedback during practice (Proteau, Marteniuk et al. 1992; Proteau, Tremblay et al. 1998). Having task dynamics more similar to the desired task would allow the subjects to explore the state-space of position and velocity parameters and develop the ability to better control balance. Additionally, groups that had greater sacral marker movement variability during training (Figure 3.4C) had greater performance improvements after training (Figure 3.3A). This suggests that when subjects explored their limits of stability, they were better able to learn how to balance while beam walking. This possibility is in agreement with a recent theoretical construct for detecting loss of balance (Ahmed and Ashton-Miller 2004).

Previous studies show little correlation between static balance and dynamic balance or standing and walking balance (Owings, Pavol et al. 2000; Shimada, Obuchi et al. 2003). It is imperative that we devise assessment tools and rehabilitation strategies that specifically target balance during walking. We developed the beam-mill that can specifically assess walking balance and potentially could be used as a means to improve balance during walking.

This study showed that 1) physical assistance hindered short-term learning of this walking task, 2) assistance may be less detrimental in more difficult tasks and 3) task specificity is important to learning, independent of error experience. Future studies should test long-term retention, include wider ranges of difficulty levels and amounts of assistance, and test patient populations to see if these principles still hold.

Acknowledgments

The authors would like to thank Kurt Sieloff for assistance with device construction and design, data collection and processing, and the other members of the Human Neuromechanics Laboratory for their assistance with data collection and processing. This work was supported by the Rackham Graduate Student Research Grant, the Foundation for Physical Therapy PODS II Scholarship, and NIH F31 HD056588-01. A similar version is in press as: Domingo A, Ferris DP. (in press) Effects of physical guidance on short-term learning of walking on a narrow beam. *Gait & Posture*.

Chapter 4. Effects of error augmentation on learning walking balance

Abstract

We studied whether error augmentation during practice would lead to greater performance gains in learning a novel walking balance task compared to practicing the task without error augmentation. We tested four groups of able-bodied subjects that walked on a treadmill-mounted balance beam (2.5 cm wide) before and after 30 minutes of training. Two of these groups practiced walking on the beam during training with a destabilization device that augmented error (Medium Destabilization and High Destabilization groups). A third group of subjects walked on a narrower beam (1.27 cm wide) during training to augment error (Narrow group). The fourth group of subjects practiced walking on the wide balance beam during training (Wide group). To measure performance, we calculated the number of times subjects stepped off of the beam per minute of successful walking on the beam (Failures per Minute). Subjects in the Wide group had 61.2 ± 6.0% (mean ± SEM) less Failures per Minute after training compared to before training. This was significantly better than the improvements after training in the other three groups (Medium Destabilization 23.6 ± 6.2%; High Destabilization 8.1 ± 5.3%; and Narrow 34.6 ± 7.9%; ANOVA, $P < 0.0001$;

THSD $P < 0.05$). The High Destabilization and Narrow groups had significant differences in motor learning ($P < 0.05$) in spite of similar errors during training. These results indicate that increasing errors during motor practice does not always improve motor learning and supports that task specific dynamics are an important consideration during gait rehabilitation for improving walking balance control.

Introduction

Physical guidance is often given in rehabilitation settings via the hands of a therapist. More recently, robotic devices have been developed to provide physical guidance in rehabilitation settings. The use of robotics has much potential in rehabilitation because of their ease of use, reliable measurement of performance and their capability to deliver a high intensity and dosage of therapy (Reinkensmeyer, Emken et al. 2004; Huang and Krakauer 2009). However, to maximize rehabilitation outcomes, we must first understand how best to use physical guidance, robotic or otherwise (Marchal-Crespo and Reinkensmeyer 2009; Reinkensmeyer and Patton 2009).

Several studies have shown that physical guidance during practice hinders motor learning. For upper limb movements, guidance given frequently during practice improved performance, but once the guidance was removed, the improvements were not present (Armstrong 1970; Winstein, Pohl et al. 1994). Similarly, we showed in a recent study that error-reducing physical assistance given during practice was detrimental to learning unassisted walking on a narrow balance beam (Domingo and Ferris in press). These findings are consistent with the theory that error detection and correction allow for forming and updating internal models during motor learning (Wolpert and Ghahramani 2000). Internal models, or neural representations, are used to compare the expected movement to the actual movement produced (Kawato 1999). When errors occur (differences between the expected and actual movement), the internal model is updated, and motor output is modified to produce the correct movement. Over time, these errors drive learning of a new internal model for new limb dynamics or environment. Previous studies have shown that motor learning is proportional to motor errors experienced in upper limb tasks (Thoroughman and Shadmehr 2000; Scheidt, Dingwell et al. 2001). From this evidence, it could be inferred that magnifying errors, rather than reducing errors, that the subject experiences may increase the rate of motor learning.

Some studies have already shown that amplifying errors improves motor learning. Error augmentation can enhance learning of visuo-motor rotations in the upper extremities of healthy subjects (Wei, Bajaj et al. 2005). In another study, robot-generated forces were applied to the arm of individuals with stroke while the moving their arm through a plane. After training, the individuals had improved movement trajectories in directions where error was amplified more than when error was reduced or was zero (Patton, Stoykov et al. 2006). For motor learning in the lower limb, Emken & Reinkensmeyer (2005) showed that error amplification lead to faster formation of the internal model in a novel walking task. No study as of yet has tested whether error augmentation could be used to improve motor learning of walking balance. This is an important question because dynamic balance is a critical component of gait control necessary for patients to safely practice walking.

The purpose of this study was to determine if augmenting error during training affects short-term learning of walking balance. We studied able-bodied subjects learning to walk on a treadmill-mounted balance beam (beam-mill). Beam walking is similar to over ground walking, but is more challenging to dynamic balance because it exploits the lateral instability of walking (Donelan, Shipman et al. 2004; Schrager, Kelly et al. 2008; Domingo and Ferris in press). We hypothesized that using error augmentation during training would improve motor learning of walking on the beam-mill more than practice without error augmentation.

Two groups of subjects practiced walking on the wide beam (2.5 cm) with a destabilization device applied at the hips (Figure 1). The destabilization device had the properties of a spring with negative stiffness and was used as a form of error augmentation. There were two levels of spring stiffness used (Medium Destabilization and High Destabilization groups). We then compared these results to a group that practiced Wide beam walking without the destabilization device (Wide group). All subjects from the three groups were evaluated on the

Wide beam without the device pre- and post-training. We hypothesized that subjects using error augmentation during practice would have greater performance gains than subjects that did not use error augmentation. We based this hypothesis on the rationale that error drives motor learning (Rumelhart, Hinton et al. 1986; Lisberger 1988; Dancause, Ptito et al. 2002), and therefore augmenting error would lead to a faster rate of learning.

We also tested a group of subjects that walked on the narrow beam during the training period (Narrow group) and was evaluated pre- and post-training on the wide beam. This group had increased task difficulty during training (i.e. error augmentation) but was more similar to the evaluation task than using the destabilization device. We hypothesized that subjects walking on the narrow beam during practice would have greater performance gains than those that practiced with the destabilization device on the wide beam. We based this hypothesis on the principle that practicing on the narrow beam would have task specific dynamics more similar to testing on the wide beam, compared to practicing with the destabilization device and testing on the wide beam. Using the destabilization device during the training period introduces an additional set of external forces applied at the pelvis. Walking on the narrow beam has more similar task dynamics to walking on the wide beam because moments are still generated at the foot to help maintain balance and no additional external forces are introduced anywhere else in the body. We specifically wanted to include this comparison because task difficulty would be similar between the Destabilization groups and the Narrow group. This removed the effect of training difficulty and would delineate the effects of practicing with more similar task specific dynamics.

Methods

Subjects

We tested 40 able-bodied subjects (see Table 4.1 for subject characteristics). Subjects were medically stable and had no history of major leg injuries. The University of Michigan Institutional Review Board approved this study. All subjects gave informed consent according to the Declaration of Helsinki prior to participating.

Procedures

Four groups of 10 subjects walked on the beam-mill for a 3-minute pre-training evaluation, a 30-minute training period (with rest breaks every 10 minutes), and a 3-minute post-training evaluation. During the pre- and post-evaluation periods, all subjects walked on the wide beam (2.5 cm wide) to test for performance gains and were made aware of this at the beginning of the experiment. The first two groups walked with the destabilization device with medium spring stiffness or

Table 4.1. Subject characteristics.

Group	Gender		Body mass (kg)	Leg length (m)
	M	F		
Narrow	2	8	60.3±10.5	0.88±0.032
Medium Destabilization	3	7	59.1±8.3	0.88±0.054
High Destabilization	2	8	60.7±8.9	0.87±0.048
Wide	4	6	64.6±14.7	0.89±0.063

high spring stiffness during the training period (Medium Destabilization or High Destabilization groups, respectively). A third group walked on the narrow beam (1.27 cm wide) during the training period (Narrow group). A fourth group walked on the wide beam during the training period (Wide group). Data presented in this paper from the Wide group were collected and published in a previous study (Domingo and Ferris in press) but are used here to compare to the data from the other three groups.

Treadmill speed was set at 0.22 m/s. This speed was determined during pilot testing. Subjects were instructed to walk on the beam for as long as possible without stepping off. Instructions were given to all subjects by the same experimenter. They had to walk heel-to-toe with arms crossed over their torso. They were also instructed not to lean forward, twist their trunk, angle their feet away from the longitudinal direction of the beam, or look down at their feet. View of the walking surface was obscured by using dribble goggles. Subjects were allowed to move their pelvis and hips laterally to help maintain balance. All subjects wore standardized orthopedic shoes. Subjects had to wait five seconds after stepping off the beam before attempting to walk on it again.

For the Medium Destabilization, High Destabilization, and Narrow groups, we also had a second day of testing. This test occurred the day immediately following the initial day of testing. Subjects walked for 3-minutes on the wide beam without the device to test for delayed retention. We also assessed subjects’ intrinsic motivation after each testing session in these three groups. We wanted to ensure that the error augmentation was not so difficult that subjects’ motivation to do well and effort would diminish. We also recorded the number of hours of sleep subjects had between the two days of testing to take into account for any differences in consolidation. The motivation questionnaire and its results are presented in Appendix 4.1 and 4.2.

Equipment

The equipment for this experiment consisted of a treadmill-mounted balance beam (beam-mill), a destabilization device, force plates and a motion capture system. The beam-mill was made of interchangeable small wooden blocks attached to the treadmill belt that lined up to make a continuous balance beam (Figure 4.1). One beam was 2.5 cm wide by 2.5 cm tall (Wide) and the other was 1.27 cm wide by 2.5 cm tall (Narrow). Smaller wooden blocks were added to either side of the bases of both beams to make them more stable in the frontal plane.

The destabilization device was made up of latex tubing springs, an over-center linkage and cables that attached to the subject via a padded hip belt (Figure 4.1). This device applied forces onto the subject with springs with an effective negative stiffness. The negative spring stiffness was achieved by placing an over-center linkage between the subject and the spring. This linkage changed the moment arms of the spring and the subject as the subject’s position changed. As the person's hips moved away from the center of the beam, a proportional force was applied to the subject in the same direction. The device made it difficult to stay on the beam if the hips moved away from the center of the beam. The device also gave subjects feedback about their position relative to the beam. The subjects were made aware of the function of the device and were encouraged not to translate anteriorly or posteriorly on the treadmill.

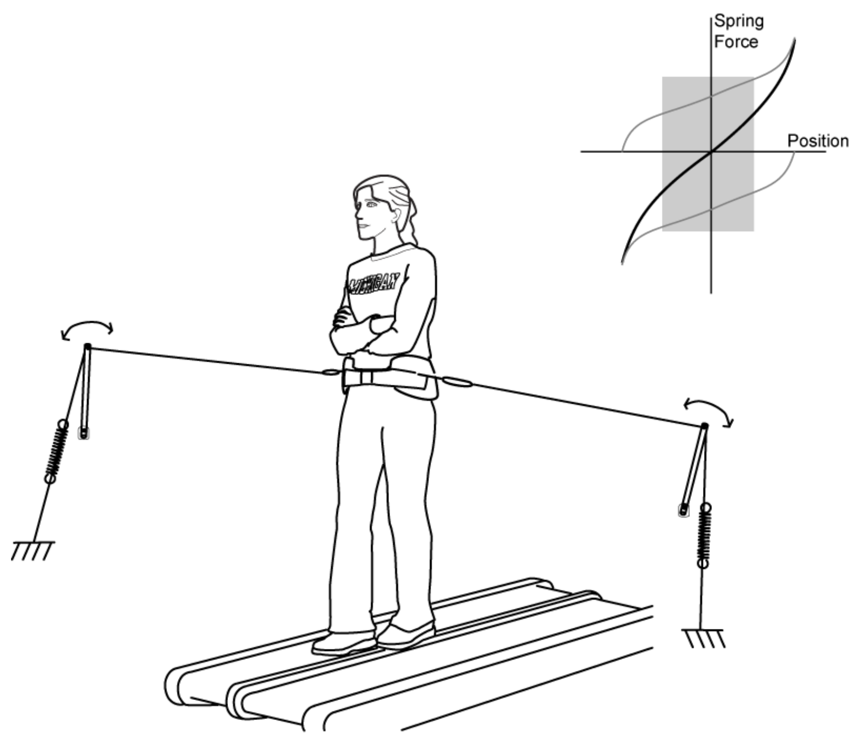

Figure 4.1. A subject walking on the beam-mill with the destabilization device used to apply forces on the subject with springs that appeared to have negative stiffness. This was accomplished by varying the moment arms via an over-center linkage placed between the springs and the subject. When the subject's pelvis moved away from the center of the beam, the device applied a proportional force onto the subject in the same direction that the subject was moving. The inset graph shows a simplified representation of the properties of the device. The thin gray lines represent the forces due to each spring, where the heavy black line shows the net force due to both springs as a function of the subject's pelvis. The shaded area represents the operating range of the device. Physical blocks were set so that the device was would stop applying additional force soon after the subject stepped off of the beam.

When the subject’s pelvis was centered over the beam, there was approximately zero net force applied to the subject. We had 8 springs of different stiffnesses. For each subject, we chose the spring that would provide the stiffness closest to the non-dimensionalized spring stiffness of 0.2978 for the Medium Destabilization group and 0.4404 for the High Destabilization group. To determine the desired spring stiffness, we used the following equation:

$$k = \bar{k} \cdot \frac{l}{mg}$$

where $k =$ dimensionalized stiffness, $\bar{k} =$ non-dimensionalized stiffness, $l =$ leg length and $mg =$ bodyweight. The non-dimensionalized spring stiffnesses of 0.2978 and 0.4404 were based on springs used during pilot testing. The average total stiffness of the device was 192.5 N/m for the Medium Destabilization group and 298.4 N/m for the High Destabilization group.

The treadmill was placed above two force plates (sampling rate 1200 Hz; Advanced Mechanical Technology Inc., Watertown, MA, USA) so that we could calculate center of pressure from the forces and moments produced by the subject while walking. The center of pressure helped us determine when the subject was on or off the beam.

We used an 8-camera video system (frame rate 120 Hz; Motion Analysis Corporation, Santa Rosa, CA, USA) to record the positions of 4 reflective markers placed on the subject's pelvis, neck and shoulders during walking. We calculated the standard deviation of the medio-lateral movement of the marker placed at the sacrum and neck to determine movement variability.

Performance measures

We recorded the number of times the subject stepped off the beam per minute. We then divided this quantity by the fraction of time the subject was on the beam (not touching the treadmill surface with either foot). This quotient, Failures per Minute, was our primary performance metric because it took into account the number of errors with respect to the amount of time the subject successfully walked on the beam. We also calculated the standard deviation (SD) of the medio-lateral movement of markers placed at the sacrum and the neck (Motion Analysis Corporation, Santa Rosa, CA; 120 Hz). We calculated percent change of the performance variables by subtracting the pre-training value from the post-training value and dividing by the pre-training value for each subject to normalize to pre-training performance. For the groups that were tested over two days, we also calculated percent change for the performance variables between the delayed retention test and the pre-training values.

For the pre- and post-training periods and delayed retention tests, we recorded data for the duration of the 3-minute trial. For the 30-minute training period, we collected only 20 seconds of data per each minute of training. We used a 4th order low-pass zero-lag Butterworth filter with a cutoff frequency of 6 Hz to smooth raw marker data. Values for SD of markers were calculated only using the data from when subjects were on the beam. We used a 4th order low-pass zero-lag Butterworth filter with a cutoff frequency of 25 Hz to smooth raw force data, then a 4th order low-pass zero-lag Butterworth filter with a cut off frequency of 6 Hz to smooth center of pressure data. Data were processed using custom software written in MATLAB (The MathWorks, Inc., Natick, MA).

Statistical Analysis

We first performed an Analysis of Variance (ANOVA) to determine if groups evaluated on the same beams had similar Failures per Minute during pre-training (JMP IN software, SAS Institute, Inc., Cary, NC).

We then performed an ANOVA to test for differences between the groups for each of the following dependent variables: percent change for Failures per Minute, Failures per Minute during training, and sacral marker SD. For post-hoc analysis, we performed Tukey's Honestly Significant Difference (THSD) test to compare results between groups as needed to delineate the differences between groups.

To test for differences between post-training and delayed retention, we performed a repeated measures ANOVA (day, group, group*day, subject(random)) for the percent change in Failures per Minute compared to the pre-training values.

We also calculated the correlation coefficient between sacral marker standard deviation and percent change in Failures per Minute. This would help to determine if there was a relationship between movement variability while walking on the beam and the performance gains.

Results

The groups that practiced with error augmentation experienced more Failures per Minute (Medium Destabilization: 27.3±2.0, High Destabilization: 29.6±1.4, Narrow: 26.5±2.8, mean±SEM) during the training period than the Wide group (12.6±1.3) (Figure 4.2) (ANOVA, $P < 0.0001$, power = 0.99, THSD, $P < 0.05$). All three error augmentation groups had similar amounts of Failures per Minute during training (THSD, $P > 0.05$).

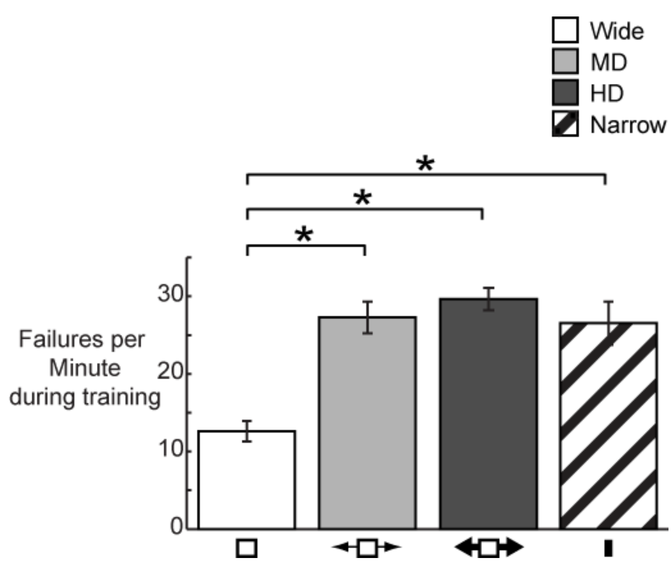

Figure 4.2. Averaged Failures per Minute during training across subjects for each group. Error bars are ±1 Standard Error of the Mean (SEM). The Medium Destabilization (MD), High Destabilization (HD) and Narrow groups had significantly greater Failures per Minute during training than the Wide group (ANOVA: $P < 0.0001$, THSD *: $P < 0.05$). There was no statistical difference between the error augmentation groups (MD, HD, Narrow) (THSD: $P > 0.05$).

Although more error was experienced during practice in the error augmentation groups than in the Wide group, the Wide group had significantly greater performance gains than all other groups (-61.2±6.0%) (ANOVA, $P < 0.0001$, power = 0.99, THSD, $P < 0.05$) (Figure 4.3). The High Destabilization group had a smaller percent change in Failures per Minute (-8.1±5.3%) than the Medium Destabilization group (-23.6±6.2%), but the difference was not significant (THSD, $P > 0.05$). The performance gains were significantly higher in the Narrow group than in the High Destabilization group (THSD, $P < 0.05$) (Figure 4.3). The Narrow group had a -34.6±7.9% change in Failures per Minute.

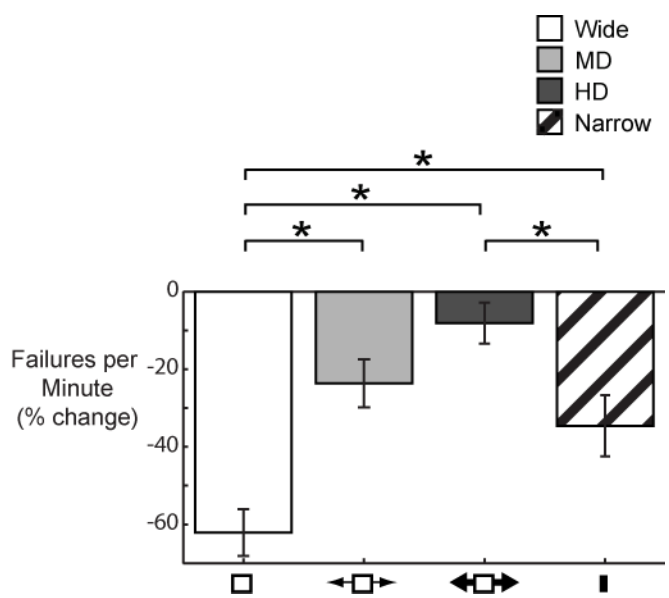

Figure 4.3. Averaged percent change ((post-training - pre-training)/pre-training values) for Failures per Minute across subjects for each group.
Error bars are ±1 SEM. The Wide group had greater performance gains after training than both Destabilization groups and Narrow group (ANOVA: $P < 0.0001$, THSD *: $P < 0.05$). The Narrow group had significantly greater performance gains than the High Destabilization group (THSD: $P > 0.05$).

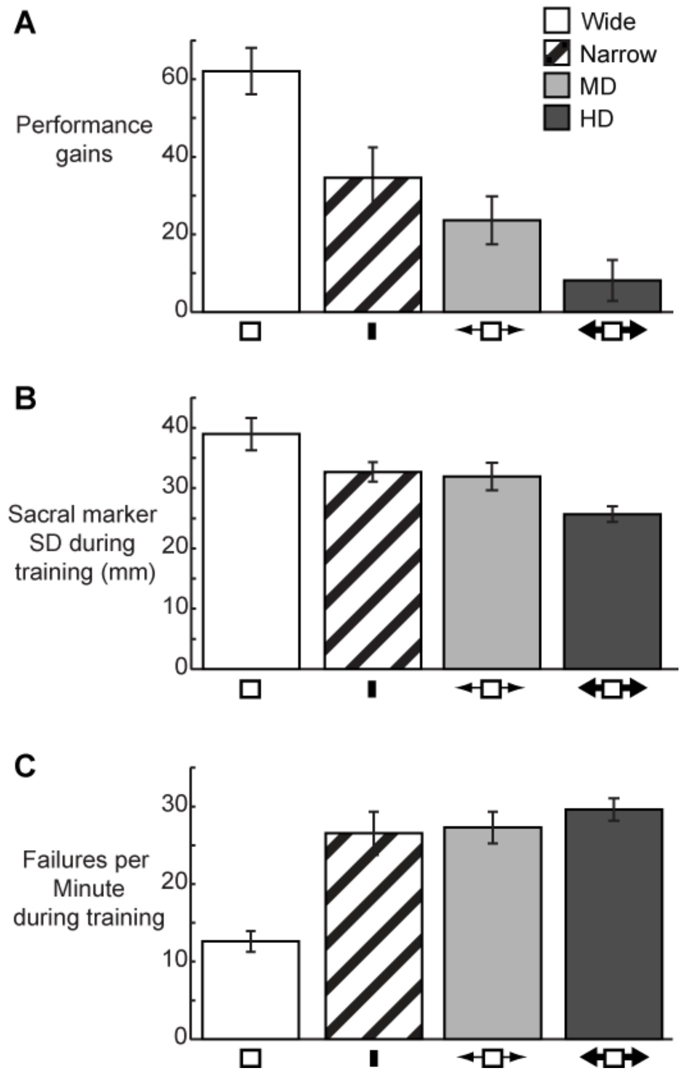

Figure 4.4. Performance gains vs. sacral marker movement variability and failures per minute during training.
Error bars are ±1 SEM. **A.** Performance gains are the absolute values of the percent change in Failures per Minute for each group. The relative performance gains between groups were similar to the **B.** relative sacral marker movement variability during training between groups, and had an inverse relationship with **C.** Failures per Minute during training.

Sacral marker movement variability vs. performance gains: The relative trend in performance gains for all groups was similar to that in the movement variability of the sacral marker (Figure 4.4A & B). The correlation coefficient, ρ, between these variables was 0.4281 ($p = 0.0059$), and $R^2 = 0.1833$. The relative trend in Failures per Minute was opposite that of the performance gains (Figure 4.4A & C).

Post training vs. delayed retention: The repeated measures ANOVA analysis showed that there was no significant difference between post-training and delayed retention values for percent change in failures per minute (ANOVA, day: $P = 0.6773$). There also was no interaction effect (ANOVA, group*day: $P = 0.2334$).

Discussion

The main result of this study showed that augmented error training with either the destabilizing device (Medium Destabilization and High Destabilization) or with a narrower balance beam was actually worse for learning walking balance than unaltered practice. This was contrary to our hypothesis based on the theory that motor learning occurs due to movement errors. We also found that when the error augmentation has more similar task dynamics to the desired task (narrow beam training), it led to greater performance gains compared to error augmentation with less similar task dynamics compared to the desired task (destabilization device training).

One explanation for why practicing with the destabilization device led to poorer performance gains compared to unaltered practice is the role of internal models in motor learning. Considerable research supports the theory that the nervous system forms internal models of movement dynamics during motor learning (Kawato 1999; Wolpert, Ghahramani et al. 2001). Recent studies have provided specific evidence that humans use internal models during walking (Emken and Reinkensmeyer 2005; Lam, Anderschitz et al. 2006) and stationary balance (Ahmed and Ashton-Miller 2007). When using the destabilization device, the dynamics of the task were changed. As a result, the learner may have formed an internal model for walking on the beam that includes the device dynamics. Once the device was removed, the subjects had an inappropriate internal model for beam walking and exhibited minimal learning during the post-training period. Detecting and correcting errors are important for motor learning, but the errors must be specific to the dynamics of the desired task.

The importance of task dynamics on internal models could also explain why subjects in the Narrow group had greater performance gains than the High Destabilization group (Figure 4.3). Walking on the narrow beam during practice likely has more similar task dynamics than walking with the destabilization device because using the destabilization device applies additional external forces to the pelvis and walking on the narrow beam does not. As a result, the internal model formed during narrow beam walking was more transferable to wide beam walking than the internal model formed during walking with the destabilizing device.

Another possible reason why the Wide beam group may have had the greatest performance gains is that practicing on the wide beam unassisted may have provided optimal level of error experience (i.e., stepping off the beam) during practice (i.e. at the “optimal challenge point” (Guadagnoli and Lee 2004)). Too many errors experienced during practice may not allow for an appropriate example of the task (Sanger 2004), and may lead to decreased motivation because of frustration. In contrast, too few errors experienced during practice may not provide enough feedback to refine the internal model of task dynamics (Scheidt, Reinkensmeyer et al. 2000; Patton, Stoykov et al. 2006). The error augmentation groups in our study had experienced more errors during practice than the Wide group. The task difficulty may have been too high to stimulate motor learning.

Our findings were different than previous studies that found error augmentation to be beneficial for motor learning. There are several reasons why this may be the case. We specifically tested learning of walking balance, while others examined learning of discrete arm movements in a plane (Patton and Mussa-Ivaldi 2004; Patton, Stoykov et al. 2006) or learning to step through a viscous force field (Emken and Reinkensmeyer 2005). These types of movements may be less complex than the task of maintaining walking balance, which involves multiple sensory inputs (visual, vestibular, and proprioceptive) and a high degree of coordination among multiple body segments in the upper and lower body. Perhaps the complexity and higher degree of difficulty of our task would not be aided by error augmentation, especially in the earlier stages of learning for our naïve subjects.

There may be some instances when error augmentation for walking balance may be useful. A common issue in rehabilitation is preparing patients for the “real world.” Walking does not always occur in a straight line and over smooth surfaces. Practicing with error augmentation may help patients respond to perturbations or changes in the environment. If the unaltered task can be performed proficiently, augmenting error with different task dynamics may be beneficial. By having diverse practice conditions, individuals can generalize learning so that learning of a new task happens at a faster rate (Seidler 2004).

Sacral marker movement variability in the frontal plane correlated well with performance gains (the absolute value of the percent change in Failures per Minute) (Figure 4.4A). The destabilization device in this study increased catastrophic error (i.e., stepping off the beam) based on the subject’s movements, but it also limited the amount of movement variability that the subject was able to experience while walking on the beam (Figure 4.4B). Movement variability at the pelvis may reflect the number of smaller errors in control that are made, allowing for updates to the internal model. This may be an alternative indicator of learning compared to catastrophic errors experienced during practice. The destabilization device may have increased catastrophic errors, but also decreased the smaller errors experienced while walking on the beam that are evidenced by movement variability. There was a significant correlation between movement variability and performance gains ($\rho = 0.4281$, $p = 0.0059$), but only 18% of the variance was explained by this relationship due to high inter-subject variability.

These ideas are consistent with the concept that humans detect a loss of balance via a "control error signal anomaly" (Ahmed and Ashton-Miller 2004). This theory suggests that humans compare incoming sensory feedback with predicted sensory feedback during movement using a forward internal model of movement dynamics. When the internal model dynamics deviate enough from the ongoing sensory feedback, then a failure is detected. Our results suggest that the subjects were improving their balance by actively exploring the movement space and learning from smaller movement errors.

This study showed that: 1) error augmentation is not always better than practicing the task unaltered, 2) task specific dynamics are important considerations for practice, and 3) movement variability of the pelvis correlates well with performance gains for beam-walking. Future studies should more specifically examine the role of movement variability and smaller control errors to delineate its effects on learning relative to catastrophic error experience. For example, learning to ride a bike with training wheels that do not touch the ground while the bicycle is vertical should provide a means for riders to explore the task space of balancing without falling over. A similar type of stabilization device for walking could provide a channel of very low forces on the torso during task space exploration while providing high forces if the torso moves too far to one side or the other to prevent failure. It could be hypothesized that practice with this type of intervention would allow for similar motor learning as unassisted practice with reduced catastrophic failures (i.e. falling down). This could be seen as similar to the type of kinematic channel in hindlimb movement used during robotic locomotor training in spinalized mice (Cai, Fong et al. 2006).

Appendix 4.1

Subject code: _____

Circle one: Day 1 Day 2

For Day 2: How many hours of sleep did you get last night? _____hours

For each of the following statements, please indicate how true it is for you, using the following scale:

1	2	3	4	5	6	7
not at all true			somewhat true			very true

_____ 1. While I was doing this activity, I was thinking about how much I enjoyed it.

_____ 2. I tried very hard on this activity.

_____ 3. I put a lot of effort into this.

_____ 4. I thought this was a boring activity.

_____ 5. I didn't try very hard to do well at this activity.

_____ 6. It was important to me to do well at this task.

_____ 7. I would describe this activity as very interesting.

_____ 8. This activity did not hold my attention at all.

_____ 9. This activity was fun to do.

_____10. I didn't put much energy into this.

_____11. I enjoyed doing this activity very much.

_____12. I thought this activity was quite enjoyable.

Appendix 4.1 Motivation questionnaire adapted from the Intrinsic Motivation Inventory. This inventory has been shown to have strong validity (McAuley, E., Duncan, T., & Tammen, V. V. (1987). Psychometric properties of the Intrinsic Motivation Inventory in a competitive sport setting: A confirmatory factor analysis. *Research Quarterly for Exercise and Sport*, 60, 48-58).

Appendix 4.2

	MD				HD				Narrow			
	Day 1		Day 2		Day 1		Day 2		Day 1		Day 2	
Quest.	mean	SEM	mean	SEM	mean	SEM	mean	SEM	mean	SEM	mean	SEM
1	4.30	0.40	4.50	0.50	3.10	0.53	4.50	0.50	3.40	0.34	4.70	0.30
2	5.60	0.45	5.70	0.54	6.30	0.26	6.20	0.24	5.70	0.26	6.10	0.23
3	5.60	0.40	5.70	0.50	6.00	0.39	6.10	0.34	6.00	0.30	6.20	0.20
4	2.40	0.37	2.50	0.40	2.50	0.27	2.10	0.35	3.40	0.31	2.40	0.27
5	1.80	0.29	1.60	0.27	1.40	0.16	1.70	0.26	1.80	0.20	1.50	0.17
6	4.80	0.55	4.90	0.48	5.50	0.34	5.20	0.39	4.80	0.49	5.10	0.50
7	5.50	0.34	5.40	0.43	4.90	0.35	5.30	0.34	4.50	0.37	5.20	0.42
8	1.40	0.16	1.70	0.26	1.90	0.28	1.70	0.21	2.20	0.25	1.80	0.25
9	5.40	0.50	5.20	0.47	4.80	0.44	5.40	0.34	4.80	0.39	5.60	0.34
10	1.80	0.25	2.10	0.35	1.70	0.15	1.90	0.23	2.00	0.26	2.00	0.30
11	5.10	0.46	5.00	0.59	4.70	0.37	5.00	0.47	4.50	0.34	5.30	0.45
12	5.20	0.36	4.90	0.47	4.60	0.45	4.90	0.43	4.60	0.37	5.60	0.45
Sleep (hr)	6.30	0.52			5.90	0.32			6.80	0.40		

Appendix 4.2 Questionnaire results for the three error augmentation groups.
MD: Medium Destabilization group, HD: High Destabilization group, Narrow: Narrow group. No significant differences were found in comparisons between groups (ANOVA, $P > 0.05$).

Chapter 5. Effects of using “assistance as needed” for learning to walk on a narrow beam

Abstract

It is common for therapists to give patients “assistance as needed” in rehabilitation settings. This paradigm has also been suggested as a means for controlling robotic devices used for neurological rehabilitation. One specific way to implement assistance as needed is to provide assistance only when the learner goes outside of a pre-determined kinematic channel, allowing unassisted movement variability within the channel. Assistance such as this greatly decreases large movement errors that may result in falling down while walking. We wanted to test if making catastrophic errors was important for learning to maintain balance during walking. We used a novel treadmill mounted balance beam to study learning walking balance in able-bodied human subjects. In this case, we define catastrophic error as losing balance so that beam-walking is no longer possible. One group practiced walking on the beam with an assist device (Assisted group) that allowed some movement of the pelvis in the frontal plane but restricted lateral movement outside a limited channel. This setup was similar to training wheels placed on a bicycle some distance from the ground, reducing

catastrophic errors but allowing movement variability. The second group practiced without the device (Unassisted group). All subjects were evaluated while walking *unassisted* before and after 30 minutes of training by calculating the number of times subjects stepped off of the beam per minute of successful walking on the beam (Failures per Minute). The two groups experienced similar amounts of movement variability during the training period (ANOVA, $P = 0.2626$). The Assisted group had significantly less Failures per Minute during training (1.7 ± 0.44) than the Unassisted group (12.6 ± 1.3) (ANOVA, $P < 0.0001$). Performance gains were significantly greater in the Unassisted group ($61.2 \pm 6.0\%$ change) than the Assisted group ($1.7 \pm 11.7\%$ change) (ANOVA, $P < 0.0001$). These results indicate that making catastrophic errors are important for learning walking balance and should not be restricted if they can be made without risk of injury.

Introduction

In rehabilitation settings, physical assistance is often given to patients to increase safety, reduce fear, or help with task completion. There are many ways to give physical assistance, but best practices indicate that assistance should only be given “as needed” (Ryerson and Levitt 1997). The amount of assistance given should only be enough to help the patient complete the task. This strategy is not only used by therapists, but has also been used in designing the control robotic devices used for rehabilitation (Marchal-Crespo and Reinkensmeyer 2009).

One example for providing assistance as needed is allowing movement to occur with some variability but providing assistance when the learner goes outside of a pre-determined kinematic “channel.” This idea is akin to using training wheels when learning to ride a bicycle. If the training wheels are placed some distance off the ground, the bicycle rider could still experience some movement variability without catastrophic failure (losing balance so forward movement is no longer possible). Remarkably, in spite of the ubiquitous use of training wheels for bicycles, there is no published scientific data indicating their effects aiding or hindering learning of balance during bicycling. For learning to walk on a narrow balance beam, a similar type of stabilization device could provide a channel of very low forces on the pelvis to allow for exploration of the task space. If the subject’s pelvis moves so far away from the center of the beam that stepping off the beam is inevitable, the device would apply higher forces to the pelvis to prevent failure. A comparable paradigm has been used to control robotic assistance that moved the legs through the motions of walking in spinalized mice (Cai, Fong et al. 2006).

We showed in a previous study that subjects that had greater performance gains in learning to walk on a narrow beam also had relatively more movement variability at the pelvis during practice (Domingo and Ferris in press). Greater movement variability (as measured by the standard deviation of the position of the sacral marker) could be an indication of the “exploration” of the subject’s limits of stability. Subjects that can explore the state-space of position and velocity parameters needed for successful beam-walking may become more aware of their limits of stability and develop the ability to better control balance.

Allowing for movement variability may be important for learning, but it has also been shown that making errors are essential for motor learning (Rumelhart, Hinton et al. 1986; Lisberger 1988; Dancause, Ptito et al. 2002). Internal models used for motor control are updated based on movement errors (Kawato 1999; Wolpert and Ghahramani 2000). Previous studies have also shown that a proportionality exists between motor errors experienced and motor learning (Thoroughman and Shadmehr 2000; Scheidt, Dingwell et al. 2001).

When patients are given assistance as needed, it often means that movement errors are limited (to ensure safety) while a normal amount of movement variability is maintained. Therefore, it is important to know the relative importance of each of these parameters of practice (making catastrophic errors and exploration of the task space) on motor learning.

In this study, subjects learned to walk on a treadmill-mounted balance beam (beam-mill) with or without a stabilizing device. We wanted to specifically examine the relative roles of "exploration" of the task space (movement variability of the pelvis) and catastrophic error (stepping off the beam) in learning to walk on the beam-mill. We tested two groups of subjects that had the experienced the same amount of exploration of the task space during beam walking but different amounts of catastrophic error during practice. We hypothesized that subjects would have greater performance gains when they experienced more catastrophic errors during practice. We based this on the idea that making errors are a critical component of learning as well as exploration of the task space.

Methods

Subjects

We tested 20 neurologically intact subjects (see Table 5.1 for subject characteristics). Subjects were medically stable and had no history of major leg injury. The University of Michigan Institutional Review Board approved this study. All subjects gave informed consent according to the Declaration of Helsinki prior to participating. Data presented in this paper from the Unassisted group were collected and published in a previous study (Domingo and Ferris, 2009).

Equipment

The equipment for this experiment consisted of a treadmill-mounted balance beam (beam-mill), an assist device, force plates and a motion capture system. The beam-mill was composed of small interchangeable wooden blocks (2.5 cm

Table 5.1. Subject characteristics.

Group	Gender		Body mass (kg)	Leg length (m)
	M	F		
Assisted	4	6	66.5±7.2	0.91±0.019
Unassisted	4	6	64.6±14.7	0.89±0.063

wide) attached to the treadmill belt that lined up into a continuous balance beam (Figure 5.1). Smaller wooden blocks were added to either side of the base of each main wooden block to make them more stable in the frontal plane.

The training device was made up of lightweight cables and adjustable straps that attached to the subject via a padded hip belt (Figure 5.1). The straps were set so that each subject would have maximal space to move in the frontal plane but not so much space that the subjects would be unable to right themselves as they were beam walking so that their pelvis was over the center of the beam. We placed single-axis tension/compression load cells (1200 Hz; Omega Engineering, Stamford, CT, USA) in series with the cables on both sides of the subject to measure the tension in the cables produced by the subjects during walking.

The treadmill was placed above two force plates (sampling rate 1200 Hz; Advanced Mechanical Technology Inc., Watertown, MA, USA) so that we could calculate center of pressure from the forces and moments produced by the subject while walking. The center of pressure helped us determine when the subject was on or off the beam.

We used an 8-camera video system (frame rate 120 Hz; Motion Analysis Corporation, Santa Rosa, CA, USA) to record the positions of 4 reflective markers placed on the subject's pelvis, neck and shoulders during walking. We calculated the standard deviation of the medio-lateral movement of the markers placed at the sacrum and neck to determine movement variability.

Procedures

Subjects walked on the beam-mill for a 3-minute pre-training evaluation, a 30-minute training period, and a 3-minute post-training evaluation. The Unassisted group walked on the beam-mill without the device during the training period. The Assisted group walked on the beam-mill with the assist device attached to a padded hip belt. The training duration was 30 minutes with rest breaks every 10 minutes. During the pre- and post-evaluation periods, all subjects walked without assistance and were made aware of this at the beginning of the experiment. It was emphasized to the subjects in the Assist group to use the device only “as needed” and not to become dependent on it, because they would not be able to use it during the post-training evaluation period.

Treadmill speed was set at 0.22 m/s. Subjects were instructed to walk on the beam for as long as possible without stepping off. Instructions were given to all subjects by the same experimenter. They had to walk heel-to-toe with arms crossed over their torso. They were also instructed not to lean forward, twist their trunk, angle their feet away from the longitudinal direction of the beam, or look down at their feet. View of the walking surface was obscured by using dribble goggles. Subjects were allowed to move their pelvis and hips laterally to help maintain balance. All subjects wore standardized orthopedic shoes. Subjects had to wait five seconds after stepping off the beam before attempting to walk on it again.

We recorded the number of times the subject stepped off the beam per minute. We then divided this quantity by the fraction of time the subject was on the beam (not touching the treadmill surface with either foot). This quotient, Failures per Minute, was our primary performance metric because it took into account the number of errors made with respect to the amount of time the subject successfully walked on the beam. We also calculated the standard deviation

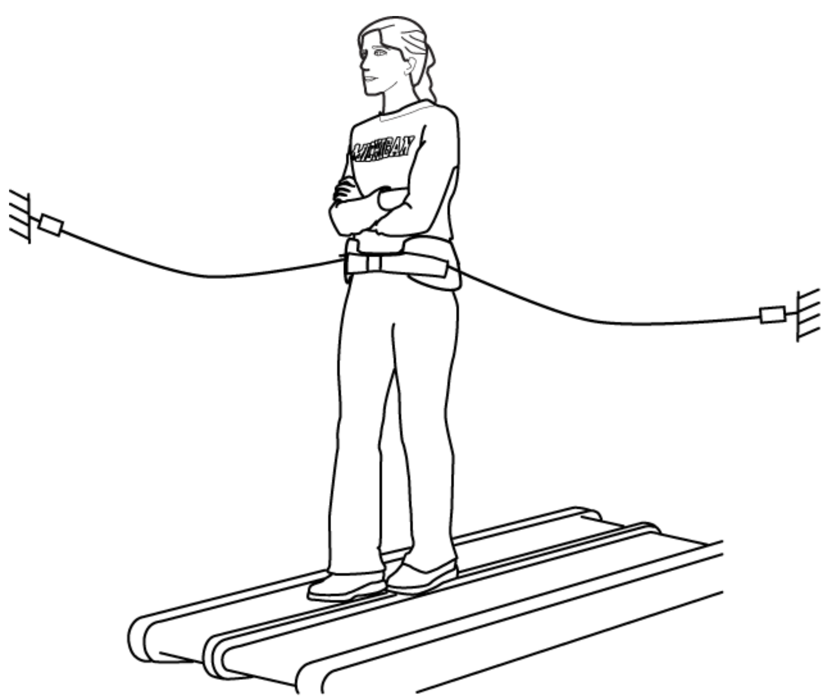

Figure 5.1. Experimental Setup.
A subject walking on the treadmill-mounted balance beam with the assist device. The assist device had straps that were set so that each subject would have maximal space to move in the frontal plane but not so much space that the subjects would be unable to right themselves as they were beam walking.

(SD) of the medio-lateral movement of markers placed at the sacrum and the neck (Motion Analysis Corporation, Santa Rosa, CA; 120 Hz) from the data when the subject was walking on the beam. As a measure of performance gains, we calculated percent change of the performance variables by subtracting the pre-training value from the post-training value and dividing by the pre-training value for each subject to normalize to pre-training performance.

For the pre- and post-training periods, we recorded data for the duration of the 3-minute trial. For the 30-minute training period, we collected only 20 seconds of data per each minute of training. We used a 4th order low-pass zero-lag Butterworth filter with a cutoff frequency of 6 Hz to smooth raw marker data. Values for SD of markers were calculated only using the data from when subjects were on the beam. We used a 4th order low-pass zero-lag Butterworth filter with a cutoff frequency of 25 Hz to smooth raw force data, then a 4th order low-pass zero-lag Butterworth filter with a cut off frequency of 6 Hz to smooth center of pressure data. Data was processed using custom software written in MATLAB (The MathWorks, Inc., Natick, MA).

We approximated the net force of the assist device on the subject by taking the difference between the tension in each cable as measured by the load cells. We then normalized the force data by dividing by bodyweight for each subject. We calculated the root-mean-square (RMS) of the normalized net force data from when the subject was on the beam as a measure of how much the subjects used the assist device for 20 seconds during each minute of training.

Statistical Analysis

We performed an Analysis of variance (ANOVA) to test for differences between the groups for each of the following dependent variables: percent change for Failures per Minute, Failures per Minute during training, and sacral marker SD during training.

To analyze the force data from the assist device, we averaged the RMS data of the training period into six 5-minute blocks. We then performed a repeated measures ANOVA as an omnibus test to find differences in force RMS between the 5-minute blocks. We performed a paired t-test to find statistical difference between the first and last 5-minute block of force RMS data.

We also compared the sacral marker SD data during training between groups to further examine if movement variability was similar between groups throughout the training period. We performed t-tests to compare data between groups during the first 5-minutes of training and the last 5-minutes of training.

Results

Figure 5.2 shows the force profiles of the assist device for 3 typical subjects for 20 seconds during Minutes 1, 5 and 30 of the training period. Figure 5.3 shows

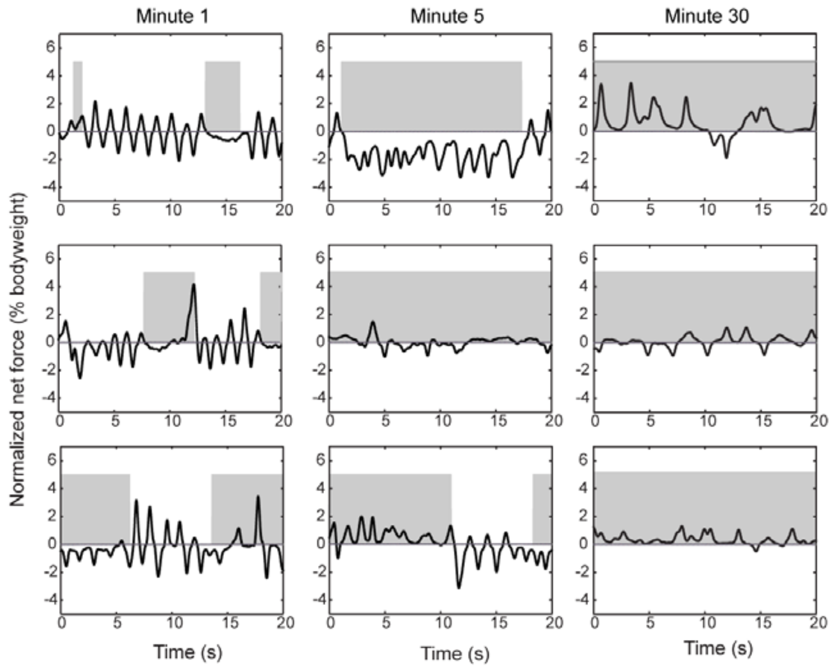

Figure 5.2. Representative force profile data of assist device.
Data are from three typical subjects during Minutes 1, 5, and 30 of the training period. Shaded gray areas represent the times the subject was on the beam.

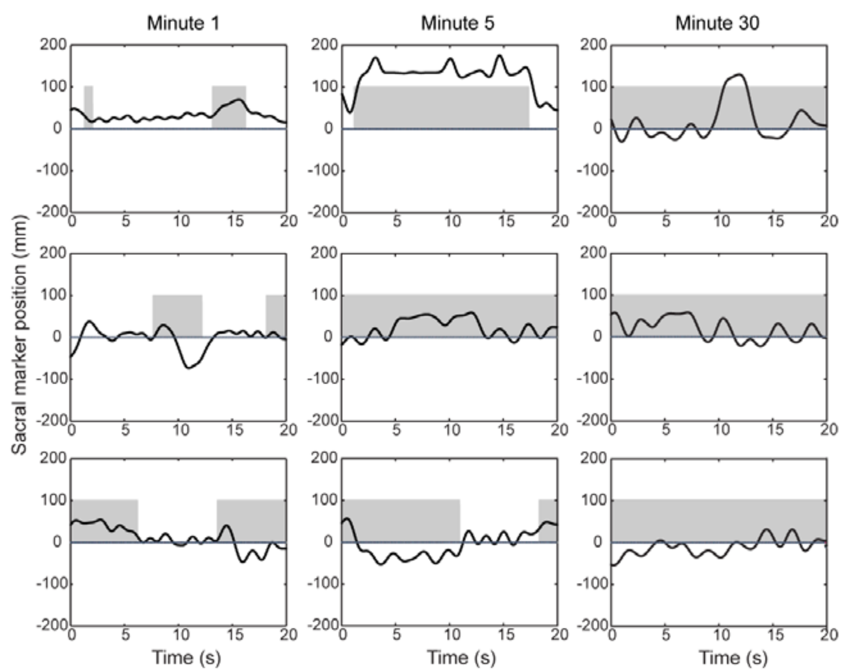

Figure 5.3. Representative sacral marker position data in the frontal plane.
Data are from three typical subjects during Minutes 1, 5, and 30 of the training period. Shaded gray areas represent the times the subject was on the beam.

the position of the sacral marker in the frontal plane for same three subjects for during Minutes 1, 5 and 30 of the training period.

Subjects in the Assisted group decreased use of the device as the training period progressed. Root-mean-square (RMS) of the net force (normalized to bodyweight) per minute was calculated for each subject. Force data was only included in calculations from when the subject was on the beam. Figure 5.4A shows the averaged force RMS for each minute of data across subjects that used assistance during the training period. We averaged the RMS across 5-minute intervals, and then performed a repeated measures ANOVA to find if there were differences in force RMS across the 30-minute training period. The analysis showed that there was a statistically significant difference between the different 5-minute blocks (ANOVA, $P < 0.0001$). Post hoc analysis showed that there the force RMS for the first 5 minutes (1.2 ± 0.24 % bodyweight) was significantly greater than for the last 5 minutes of the training period (0.72 ± 0.15% bodyweight) (paired t-test, $P = 0.0265$).

Using the assist device greatly reduced the number of failures during training, but during post-training, the number of errors returned to pre-training values. Figure 5.4B shows the averaged Failures per Minute for both groups during pre- and post-training and during each minute of training. Figure 5.4C shows the averaged sacral marker SD for both groups during pre-and post-training and during each minute of training.

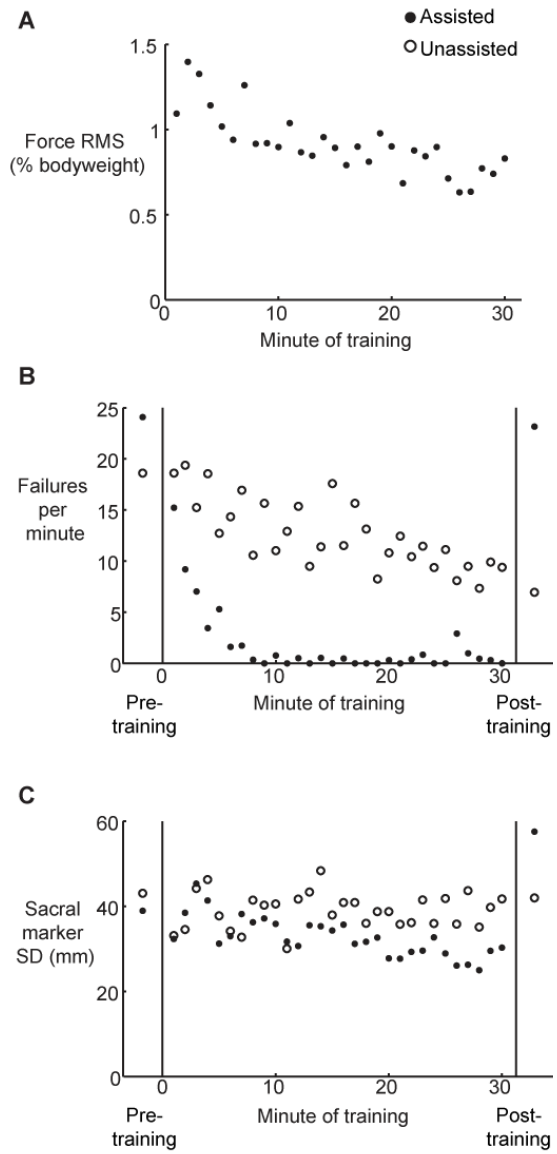

Figure 5.4. Averaged time series data from the training period.
A. Averaged root-mean-square (RMS) of net force from the assist device as a percent of bodyweight. Data are taken only from when subjects were walking on the beam. **B.** Averaged number of Failures per Minute for each minute across subjects for each group. The Assisted group had very few Failures per Minute after 10 minutes of the training period. **C.** Averaged standard deviations (SD) for the sacral marker in the frontal plane as a measure of movement variability across subjects for each group. Data included in the calculation was only from when subjects were on the beam. Averaged data from the first 5 minutes of training showed that there were no differences in movement variability (SD) between groups. Averaged data from the last 5 minutes of training showed that movement variability was higher in the Unassisted group (ANOVA, P = 0.0143).

We wanted to verify that both groups were had similar amounts of movement variability over the whole training period. Sacral marker movement variability was slightly greater in the Unassisted group (39.0 ± 2.7 mm, mean ± SEM) than the Assisted group (32.7 ± 4.7 mm), but the difference was not significant (ANOVA, P = 0.2626) (Figure 5.5A). When comparing 5-minute blocks of data during the training period, we found that there were no differences in sacral marker SD (t-test, P = 0.8760) between groups during the first 5 minutes of training. During the last 5 minutes of training, movement variability was greater in the Unassisted group (t-test, P = 0.0143) by 30%.

We also wanted to ensure that the assist device was effective at preventing subjects from stepping off the beam. We compared the number of Failures per Minute during training for both groups and found that they were significantly different (ANOVA, P < 0.0001) (Figure 5.5B). The Assisted group had an average of 1.7 ± 0.44 Failures per Minute during the training period, while the Unassisted group had an average of 12.6 ± 1.3 Failures per Minute during training.

Practicing with the assist device clearly hindered learning (Figure 5.5C). The Assisted group had -1.7 ± 11.7% change in Failures per Minute, while the Unassisted group had -61.2 ± 6.0% change in Failures per Minute. There were much greater performance gains in the Unassisted group (ANOVA, P = 0.0003).

Discussion

The main result of this study showed that experiencing catastrophic errors (stepping off the beam) during practice is important for learning this beam walking task. Subjects that received "assistance as needed" experienced

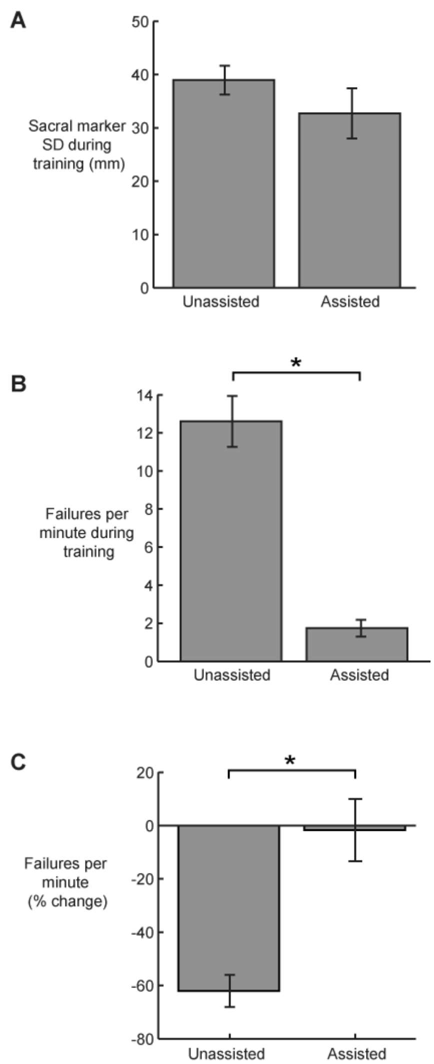

Figure 5.5. Averaged sacral marker SD during training, Failures per Minute during training, and percent change in Failures per minute across subjects for each group.
A. Sacral marker SD calculated from marker position in the medio-lateral direction when subjects were on the beam. Data is averaged over the entire training period across all subjects. The difference in movement variability between groups was not significant (ANOVA, P = 0.2626) **B.** Averaged Failures per Minute over the entire training period across all subjects. The Assisted group had significantly less Failures per Minute during training (ANOVA *: P < 0.0001). **C.** Averaged percent change in Failures per Minute from pre- to post-training. Using the assist device during practice clearly hindered learning, as there were significantly greater performance gains in the Unassisted group than the Unassisted group (ANOVA *: P = 0.0003).

movement variability similar to unassisted subjects but had a reduced number of failures during practice. This suggests that giving assistance in this manner, akin to using static training wheels when learning to ride a bicycle, may increase safety, but is not helpful for motor learning of a walking balance task.

There are several reasons why this type of assistance may have hindered learning of narrow beam walking. A learner's ability to recognize and correct their errors increases as movement skill improves (Liu and Wrisberg 1997). Although using the assist device allowed for a similar amount of movement variability as the unassisted group, it also greatly reduced opportunities for error detection and correction. After about 10 minutes of training, the Assisted group rarely stepped off the beam, while the Unassisted group continued to step off the beam throughout the training period. The lack of error experience in the Assisted group may have hindered learning. This is in agreement with the idea that internal models are formed and updated based on movement errors (Kawato 1999; Wolpert and Ghahramani 2000). If no movement error is sensed, then the internal model cannot be updated and refined for the desired task.

The assist device also changed task dynamics by applying forces to stop lateral translation of the pelvis once the subject reached a pre-determined distance away from center. The presence of these forces could greatly affect how subjects learn to maintain balance on the beam-mill. Strategies formed to balance while using the assist device are likely very different than those used without the device.

Subjects in the Assisted group had about the same amount of pelvis movement variability as the Unassisted group throughout most of the training period (Figure 5.4C). This shows that the assist device did allow enough space for normal movement variability. However, the subjects in the Assisted group were less variable with their movements by the end of training. These subjects were told at the beginning of the experiment not to become dependent on the device because they would be evaluated on Unassisted beam-walking. They may have concentrated too much on avoiding using the device assistance and as a result, ended up with reduced movement variability. Alternatively, subjects may have been able to use very low forces from the device towards the end of training for feedback to limit their movement variability.

Although the Assisted group used the device minimally during training, especially towards the end of training (Figures 5.2 & 5.4A), even very small forces may have helped to maintain balance. Several studies have shown that light touch (less than 1 Newton of force) at the fingertip can greatly reduce postural sway during standing with eyes closed due to the augmented sensory feedback rather than physical stabilization (Holden, Ventura et al. 1994; Jeka and Lackner 1994; Kouzaki and Masani 2008). In our study, most subjects in the Assisted group reported that they felt they had greatly decreased the use of the device at the end of the training period. However, it is possible that subjects unknowingly became dependent on the very low forces during practice. These forces may have been able to give some cues to their position in space. Perhaps these low forces from the device within the kinematic channel could be eliminated by placing physical blocks a small distance away from each side of the pelvis. Even so, the restriction in movement provided by these blocks would likely change task dynamics enough to hinder learning.

A recent study comparing the effectiveness of locomotor training using the Lokomat (a robotic exoskeleton used for automated treadmill stepping) versus conventional gait training in patients with subacute stroke (Hidler, Nichols et al. 2009) supported the results of our study. They found that subjects that received conventional gait training had greater improvements in gait speed and walking distance than those that trained in the Lokomat. They attributed these results in part to how the Lokomat provides guidance of the lower extremities and greatly restricts motion at the trunk and pelvis. If motion is limited at the trunk and pelvis, the patients are unable to sense and correct for movement errors during walking and would greatly limit learning of balance.

There is another possible reason why the Assisted group had lower performance gains. There are two separate parts of this task that require different dynamics: getting on the beam initially and then taking steps to stay on the beam. Because this group spent most of their time walking on the beam, they had fewer opportunities to learn the act of successfully getting back on the beam after a failure. Without this skill, subjects were more likely to step off the beam soon after stepping on, greatly increasing the number of Failures per Minute.

Task space exploration and making errors are closely related. This is supported by the observation that humans seem to detect a loss of balance with a “control error signal anomaly” (CEA) during standing balance (Ahmed and Ashton-Miller 2004; Ahmed and Ashton-Miller 2007). To determine motor output for a desired movement, the central nervous system creates an internal model of limb dynamics based on previous sensorimotor experiences. The expected sensory feedback is then compared to the actual sensory feedback. If a sufficiently large error (CEA) is detected, then a compensatory response will occur. Subjects that successfully learned to walk on the narrow balance beam may have been actively exploring the task space, detecting errors, and using the movement errors to update the internal model.

This study showed that making catastrophic errors was important for learning to walk on the beam-mill. Rehabilitation strategies should be devised so that assistance allows patients to make catastrophic errors (so that the goal movement is no longer possible) during practice but still maintain safety and prevent falls. Our results also showed that maintaining task dynamics during practice is also of utmost importance.

Chapter 6. General Discussion

The overall goal of this dissertation was to examine the effects of physical guidance on motor control and learning of walking. In my first experiment, I found that using manual assistance to help move the legs through the motions of walking did not substantially change muscle activation amplitudes and also helped to keep muscle activation patterns more similar to those in neurologically intact subjects while stepping at faster speeds. The results from this study suggest that physical guidance can be helpful for gait rehabilitation.

In the next three experiments, I investigated the effects of different types of physical assistance used during practice on learning unassisted beam-walking. Subjects that used the spring-based lateral stabilization device saw little to no improvement in unassisted beam-walking. Subjects that practiced with error augmentation had improved performance after training. The magnitude of their improvements depended on how similar the task dynamics of practice were compared to those of the desired task (walking unassisted). The subjects that practiced with the destabilization device had smaller performance gains than the subjects that practiced on a narrower beam as a form of error augmentation. In the last experiment, subjects that practiced with a device that allowed normal movement variability and smaller control errors but limited catastrophic error (stepping off the beam) had little to no performance gains. In each of these studies, the group that had the greatest performance gains was always the group that practiced without assistance.

There is a lack of controlled studies on how physical guidance affects control and learning of a whole body task. This is surprising considering that physical guidance is used so frequently in rehabilitation settings. The need to understand how best to use physical guidance grows with the advent of robotics being used for gait rehabilitation. The studies in this dissertation were able to provide some insight as to why current robotic devices have not been as effective in locomotor rehabilitation.

The results of my studies emphasized the importance of maintaining task specificity during practice and that they should be prioritized when determining treatment protocols. Physical guidance may alter task dynamics to varying degrees, thereby affecting learning. My experimental outcomes were grounded in the theory of the internal model of motor control. The internal model controls movement by comparing expected sensory feedback with ongoing sensory feedback. If a movement error is sensed, the internal model is re-calibrated so that upcoming motor commands will help make the correct movements. These errors are usually specific to the task and environment. If the practice environment and desired task environment are too different, performance of the desired task will not improve.

Although several studies have shown that error is proportional to learning (Thoroughman and Shadmehr 2000; Scheidt, Dingwell et al. 2001), there is evidence to the contrary for some movement tasks. Fine and Thoroughman (2006) found that motor adaptation was insensitive to the amplitude of error but was sensitive to the direction of error. These results might be a result of the constraints of the specific task and the magnitude of perturbations used.

My results are in accord with studies that show gait training with a robotic exoskeleton results in limited improvements in gait in subjects with incomplete spinal cord injury (Wirz, Zemon et al. 2005) and stroke (Hidler, Nichols et al. 2009) compared to body-weight supported treadmill training or conventional gait training. There are several reasons why this may have happened. First, subjects walking in the exoskeleton were not permitted to experience errors in movement. In addition, the movement of the pelvis and trunk is greatly restricted in the exoskeleton. These restrictions in movement do not allow the subject to learn walking balance because the device obviates the need to control balance. Robotic devices such as the Lokomat may be useful for the earliest stages of mobilization, but likely should not be used for patients that are able to produce their own steps because of the manner in which it provides guidance.

Overall, the studies in this dissertation support the use of physical guidance during gait rehabilitation but emphasize that task specificity must be maintained as much as possible during practice. This can be problematic as most forms of physical guidance used during balance re-training substantially alter task dynamics. These findings should provide important insight for designing robotic devices for gait rehabilitation. Usually errors are prevented during practice to maintain patient safety, but it may be more beneficial to find methods that allow patients to make movement errors without the risk of injury, so they can learn from them.

Strengths of approach

I built a treadmill-mounted balance beam that provided a means to specifically assess walking balance in able-bodied subjects. A task such as this is distinctive because most quantitative measures of balance are measures of static or standing balance. Since most falls occur during walking and not standing (Blake, Morgan et al. 1988), a tool such as the beam-mill could help to gain insight on the principles of learning walking balance. Understanding how humans learn walking balance would be very important for designing treatment protocols for gait rehabilitation. The beam-mill also allowed me to vary task difficulty because the blocks that formed the balance beam were interchangeable.

The task of walking on the beam-mill can provide direct insight into how people control and learn to balance during walking. There are established methods to quantify standing balance (computerized dynamic posturography), but there is no equivalent test for walking balance. The beam-mill provided a method to discretely quantify walking balance.

Study limitations

For the walking balance studies, the level of assistance given to subjects did not change throughout training. This was one way of controlling the manner in which assistance was given between groups. However, recent studies suggest that task difficulty should be dynamically adjusted to the skill level of the learner (Guadagnoli and Lee 2004; Choi, Qi et al. 2008). Changing task difficulty so that some level of error experienced is maintained helps to ensure that performance gains continue to occur. Subjects that received assistance that reduced error (Chapters 3 and 5) essentially stopped making errors (stepping off the beam) after about 10 minutes of training. We could have attempted to maintain the “optimal challenge point” by reducing spring stiffness in the lateral stabilization device once the subjects went below this point. However it may take extensive pilot testing to determine what the “optimal challenge point” is for this task.

Another limitation to the walking balance studies was the limited amount of practice the subjects had on the beam-mill. Although I did observe differences in learning between groups, the results may have been different if I had multiple days of training rather than just one. In groups where I tested delayed retention, there were no differences between performance immediately after training compared to performance during the second day of testing. However, it is possible that there would be differences in delayed or long-term retention if subjects practiced walking on the beam over multiple days.

It is possible that the lack of performance gains observed when using the assist devices may have been due to the specific devices themselves and not the use of physical guidance as a whole. There are several parameters of these devices that could be modified. For example, if the springs used in the stabilization device used in Chapter 3 were less stiff, then greater performance gains may have been observed. In Chapter 5, it is possible that if physical blocks or bumpers were used to limit catastrophic error rather than cables connected to the hip belt, the low forces within the kinematic channel would have been eliminated. This may have led to greater performance gains.

It could also be argued that the unassisted groups had the greatest performance gains because the control subjects had training most similar to the evaluation test. The groups that used the lateral stabilization device or had augmented error during practice could be considered as having performed transfer tasks during the post-test. This is an important consideration but in rehabilitation settings the practice environment is almost always different from the desired task. This is why we tested different forms of error augmentation in Chapter 4.

In Chapters 3-5, we compared different groups of subjects that practiced under different conditions. Alternatively, the study could have been a repeated-measures or Latin Squares design. Subjects would have practiced with each of the different devices and then immediately tested on unassisted beam-walking to eliminate any between-subject differences. However, this design also has limitations because of the practice effects. Every time the subjects practiced walking on the beam, it would affect subsequent performance. For this reason, I chose to test each subject with only one practice condition.

One specific limitation of using the lateral destabilization device is that the subjects' goals may have changed while wearing the device. During the pre- and post-training trials, the subjects' instructions were to walk on the beam for as long as possible without stepping off. However, when the subject's wore the lateral destabilization device, their goal may have changed to keep their hips as still as possible so the device would not pull them off the beam. This strategy is useful for staying on the beam without the device, but subtly changed the goals of the subject between the training trial and the evaluation trials.

When assistance was used in the walking balance studies, the level of assistance was not changed during practice. This may have limited performance gains because studies using knowledge of results to augment feedback during practice have shown it to be more effective in enhancing learning when it is tapered rather than presented with every trial (Salmoni, Schmidt et al. 1984). Perhaps if the physical guidance was tapered during practice based on time or performance (Choi, Qi et al. 2008), subjects using assistance during practice would have experienced greater performance gains. This could be accomplished by decreasing spring stiffness or decreasing beam width.

Another weakness to the experimental design was the small number of subjects in each group (N = 10). Because of this, small differences in the baseline performance between groups may have affected the experimental outcomes. Collecting more subjects would add more statistical power. However, I am confident the results would hold for a larger sample size given the quality of the results and their match with underlying motor learning theory.

Chapter 7. Conclusions

Major findings

Chapter 2:

Hypothesis: For this study, there were two competing hypotheses. EMG activity in individuals with incomplete spinal cord injury (SCI) could decrease because the manual assistance given to the subjects would decrease effort. EMG activity could also increase because the manual assistance would help to provide more normative joint kinematics and proprioceptive input.

Findings: EMG amplitudes in individuals with incomplete SCI did not change with manual assistance. EMG profiles stayed more similar to those of able bodied subjects at higher speeds when they walked with manual assistance.

Chapter 3:

Hypothesis: Able-bodied subjects that practiced without physical guidance would have greater performance gains in unassisted beam walking than those that did not because error drives motor learning and assistance tends to reduce errors. I also hypothesized the difference in performance gains would be less for those learning the more difficult task (walking on the narrower beam).

Findings: 1) Using physical guidance that reduced errors during practice hindered short-term learning of narrow beam walking, 2) assistance may be less detrimental in more difficult tasks and 3) task specificity is important to learning, independent of error experience.

Chapter 4:

Hypothesis: Subjects that used the error augmentation device during practice will have greater performance gains in unassisted walking.

Findings: 1) Error augmentation is not always better than practicing the task unaltered, 2) task specificity is an important considerations for practice and 3) movement variability of the pelvis correlates well with performance gains for beam-walking.

.

Chapter 5:

Hypothesis: Holding movement variability equal, subjects that experience catastrophic error will have greater performance gains than those that do not experience catastrophic error during training.

Findings: Subjects that experienced catastrophic error during practice had much greater performance gains than those that did not. Although the exploration of task space is important, experiencing catastrophic error is essential for motor learning of a narrow beam walking task.

Recommendations for future work

My dissertation studies have revealed the relative importance of task specificity during practice. Although physical assistance has proven to be overall detrimental to learning narrow beam walking, it will continue to be a mainstay in rehabilitation settings because safety cannot be compromised. In addition, some patients may have decreased strength or dyscoordination and would need physical guidance to complete movements.

Results from my dissertation studies suggest that another next step for this line of research should examine the use of performance-based adaptive practice schedules (Choi, Qi et al. 2008). In each of my studies, I kept the level of assistance constant throughout the practice period. The Challenge Point Framework for motor learning suggests the difficulty of the task should be dynamically adjusted to the skill level of the learner (Guadagnoli and Lee 2004). Therapists constantly re-assess their patients' abilities and adapt their treatment activities to maintain a moderately high level of difficulty so that performance will improve at a steady rate. However, it is difficult to define and articulate how and when to make these changes to the treatment program. Extensive pilot testing (or clinical experience when dealing with patients) would be required to know how and when to change task difficulty to maximize learning.

It is possible that physical guidance could be helpful in the earlier stages of learning very difficult tasks. Physical guidance could help provide successful examples of the task. In the studies described in this dissertation, subjects were able to take at least a couple of steps in their first attempts of walking on the beam-mill. If able-bodied subjects were tested on an even more difficult task, such as walking on a tightrope instead of a balance beam (or a patient with stroke learning to walk again), it is possible that physical guidance could increase the rate of learning in the beginning stages. Otherwise, these tasks would be almost impossible to perform initially. Future studies should also investigate how physical guidance affects motor learning in clinical populations to see if these principles still hold.

The beam-mill also has potential to be used as a balance assessment and/or treatment tool because it specifically challenges walking balance. Walking balance deficits could be quantified with the beam-mill by testing subjects on different width beams. Studies would need to show the reliability and validity of the beam-mill as a balance assessment tool. Patients that need to improve their balance could also walk on the beam-mill to challenge their dynamic balance and lessen any existing balance deficits.

References

Ahmed, A. A. and J. A. Ashton-Miller (2004). "Is a "loss of balance" a control error signal anomaly? Evidence for three-sigma failure detection in young adults." Gait Posture **19**(3): 252-62.

Ahmed, A. A. and J. A. Ashton-Miller (2007). "On use of a nominal internal model to detect a loss of balance in a maximal forward reach." J Neurophysiol **97**(3): 2439-47.

Allum, J. H. and N. T. Shepard (1999). "An overview of the clinical use of dynamic posturography in the differential diagnosis of balance disorders." J Vestib Res **9**(4): 223-52.

Armstrong, T. R. (1970). Training for the production of memorized movement patterns. Psychology. Ann Arbor, University of Michigan.

Arsenault, A. B., D. A. Winter, et al. (1986). "How many strides are required for the analysis of electromyographic data in gait?" Scandinavian Journal of Rehabilitation Medicine **18**(3): 133-135.

Barbeau, H., M. Ladouceur, et al. (2002). "The effect of locomotor training combined with functional electrical stimulation in chronic spinal cord injured subjects: walking and reflex studies." Brain Research Reviews **40**(1-3): 274-91.

Bauby, C. E. and A. D. Kuo (2000). "Active control of lateral balance in human walking." Journal of Biomechanics **33**(11): 1433-40.

Behrman, A. L. and S. J. Harkema (2000). "Locomotor training after human spinal cord injury: a series of case studies." Physical Therapy **80**(7): 688-700.

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

Blake, A. J., K. Morgan, et al. (1988). "Falls by elderly people at home: prevalence and associated factors." Age Ageing **17**(6): 365-72.

Cai, L. L., A. J. Fong, et al. (2006). "Implications of assist-as-needed robotic step training after a complete spinal cord injury on intrinsic strategies of motor learning." Journal of Neuroscience **26**(41): 10564-8.

Chang, C. L. and B. D. Ulrich (2008). "Lateral stabilization improves walking in people with myelomeningocele." J Biomech **41**(6): 1317-23.

Choi, Y., F. Qi, et al. (2008). "Performance-based adaptive schedules enhance motor learning." J Mot Behav **40**(4): 273-80.

Collins, S. H., P. G. Adamczyk, et al. (2009). "A simple method for calibrating force plates and force treadmills using an instrumented pole." Gait Posture **29**(1): 59-64.

Colombo, G., M. Wirz, et al. (2001). "Driven gait orthosis for improvement of locomotor training in paraplegic patients." Spinal Cord **39**(5): 252-255.

Dancause, N., A. Ptito, et al. (2002). "Error correction strategies for motor behavior after unilateral brain damage: short-term motor learning processes." Neuropsychologia **40**(8): 1313-23.

Dietz, V., G. Colombo, et al. (1995). "Locomotor capacity of spinal cord in paraplegic patients." Annals of Neurology **37**(5): 574-582.

Dietz, V., M. Wirz, et al. (1998). "Locomotor pattern in paraplegic patients: Training effects and recovery of spinal cord function." Spinal Cord **36**(6): 380-390.

Dobkin, B. (1999). "An overview of treadmill locomotor training with partial weight support: A neorophysiologically sound approach whose time has come for randomized clinical trials." Neurorehabilitation and Neural Repair **13**: 157-165.

Dobkin, B., D. Apple, et al. (2006). "Weight-supported treadmill vs over-ground training for walking after acute incomplete SCI." Neurology **66**(4): 484-93.

Dobkin, B. H., D. Apple, et al. (2003). "Methods for a randomized trial of weight-supported treadmill training versus conventional training for walking during inpatient rehabilitaiton after incomplete traumatic spinal cord injury." Neurorehabilitation and Neural Repair **17**(3): 153-167.

Dobkin, B. H., S. Harkema, et al. (1995). "Modulation of locomotor-like EMG activity in subjects with complete and incomplete spinal cord injury." Journal of Neurologic Rehabilitation **9**: 183-190.

Domingo, A. and D. P. Ferris (in press). "Effects of physical guidance on short-term learning of walking on a narrow beam." Gait & Posture.

Donelan, J. M., D. W. Shipman, et al. (2004). "Mechanical and metabolic requirements for active lateral stabilization in human walking." J Biomech **37**(6): 827-35.

Emken, J. L. and D. J. Reinkensmeyer (2005). "Robot-enhanced motor learning: accelerating internal model formation during locomotion by transient dynamic amplification." IEEE Transactions on Neural Systems and Rehabilitation Engineering **13**(1): 33-9.

Ferris, D. P., K. E. Gordon, et al. (2004). "Muscle activation during unilateral stepping occurs in the nonstepping limb of humans with clinically complete spinal cord injury." Spinal Cord **42**(1): 14-23.

Field-Fote, E. C. (2001). "Combined use of body weight support, functional electric stimulation, and treadmill training to improve walking ability in individuals with chronic incomplete spinal cord injury." Archives of Physical Medicine and Rehabilitation **82**(6): 818-824.

Field-Fote, E. C. and D. Tepavac (2002). "Improved intralimb coordination in people with incomplete spinal cord injury following training with body weight support and electrical stimulation." Physical Therapy **82**(7): 707-15.

Fine, M. S. and K. A. Thoroughman (2006). "Motor adaptation to single force pulses: sensitive to direction but insensitive to within-movement pulse placement and magnitude." J Neurophysiol **96**(2): 710-20.

Grasso, R., Y. P. Ivanenko, et al. (2004). "Distributed plasticity of locomotor pattern generators in spinal cord injured patients." Brain **127**(5): 1019-34.

Guadagnoli, M. A. and T. D. Lee (2004). "Challenge point: a framework for conceptualizing the effects of various practice conditions in motor learning." J Mot Behav **36**(2): 212-24.

Harkema, S. J., S. L. Hurley, et al. (1997). "Human lumbosacral spinal cord interprets loading during stepping." Journal of Neurophysiology **77**(2): 797-811.

Hesse, S., C. Bertelt, et al. (1995). "Treadmill training with partial body weight support compared with physiotherapy in nonambulatory hemiparetic patients." Stroke **26**(6): 976-981.

Hesse, S., D. Uhlenbrock, et al. (2000). "A mechanized gait trainer for restoring gait in nonambulatory subjects." Archives of Physical Medicine and Rehabilitation **81**(9): 1158-1161.

Hicks, A. L., M. M. Adams, et al. (2005). "Long-term body-weight-supported treadmill training and subsequent follow-up in persons with chronic SCI: effects on functional walking ability and measures of subjective well-being." Spinal Cord **43**(5): 291-8.

Hidler, J., D. Nichols, et al. (2009). "Multicenter randomized clinical trial evaluating the effectiveness of the Lokomat in subacute stroke." Neurorehabil Neural Repair **23**(1): 5-13.

Hidler, J. M. (2005). "Guest Editorial: What is next for locomotor-based studies?" Journal of Rehabilitation Research and Development **42**(1): xi-xiv.

Hogrel, J. Y. (2005). "Clinical applications of surface electromyography in neuromuscular disorders." Neurophysiol Clin **35**(2-3): 59-71.

Holden, M., J. Ventura, et al. (1994). "Stabilization of posture by precision contact of the index finger." J Vestib Res **4**(4): 285-301.

Hornby, T. G., D. D. Campbell, et al. (2008). "Enhanced gait-related improvements after therapist- versus robotic-assisted locomotor training in subjects with chronic stroke: a randomized controlled study." Stroke **39**(6): 1786-92.

Huang, H. J. and D. P. Ferris (2004). "Neural coupling between upper and lower limbs during recumbent stepping." Journal of Applied Physiology **97**(4): 1299-308.

Huang, V. S. and J. W. Krakauer (2009). "Robotic neurorehabilitation: a computational motor learning perspective." J Neuroeng Rehabil **6**: 5.

Israel, J. F., D. D. Campbell, et al. (2006). "Metabolic costs and muscle activity patterns during robotic- and therapist-assisted treadmill walking in individuals with incomplete spinal cord injury." Physical Therapy **86**(11): 1466-78.

Jeka, J. J. and J. R. Lackner (1994). "Fingertip contact influences human postural control." Exp Brain Res **100**(3): 495-502.

Kaelin-Lang, A., L. Sawaki, et al. (2005). "Role of voluntary drive in encoding an elementary motor memory." Journal of Neurophysiology **93**(2): 1099-103.

Kao, P. C. and D. P. Ferris (2005). "The effect of movement frequency on interlimb coupling during recumbent stepping." Motor Control **9**(2): 144-163.

Kawashima, N., D. Nozaki, et al. (2005). "Alternate leg movement amplifies locomotor-like muscle activity in spinal cord injured persons." Journal of Neurophysiology **93**(2): 777-785.

Kawato, M. (1999). "Internal models for motor control and trajectory planning." Current Opinion in Neurobiology **9**(6): 718-727.

Kouzaki, M. and K. Masani (2008). "Reduced postural sway during quiet standing by light touch is due to finger tactile feedback but not mechanical support." Exp Brain Res **188**(1): 153-8.
Kuo, A. D. (1999). "Stabilization of lateral motion in passive dynamic walking." International Journal of Robotics Research **18**: 917–930.
Lam, T., M. Anderschitz, et al. (2006). "Contribution of feedback and feedforward strategies to locomotor adaptations." Journal of Neurophysiology **95**(2): 766-73.
Lam, T., C. Wolstenholme, et al. (2003). "How do infants adapt to loading of the limb during the swing phase of stepping?" Journal of Neurophysiology **89**(4): 1920-8.
Lisberger, S. G. (1988). "The neural basis for learning of simple motor skills." Science **242**(4879): 728-35.
Liu, J., S. Cramer, et al. (2006). "Learning to perform a new movement with robotic assistance: comparison of haptic guidance and visual demonstration." Journal of Neuroengineering and Rehabilitation **3**: 20.
Liu, J. and C. A. Wrisberg (1997). "The effect of knowledge of results delay and the subjective estimation of movement form on the acquisition and retention of a motor skill." Res Q Exerc Sport **68**(2): 145-51.
Llewellyn, M., J. F. Yang, et al. (1990). "Human H-reflexes are smaller in difficult beam walking than in normal treadmill walking." Experimental Brain Research **83**(1).
Lotze, M., C. Braun, et al. (2003). "Motor learning elicited by voluntary drive." Brain **126**: 866-72.
Maegele, M., S. Muller, et al. (2002). "Recruitment of spinal motor pools during voluntary movements versus stepping after human spinal cord injury." Journal of Neurotrauma **19**(10): 1217-29.
Marchal-Crespo, L. and D. J. Reinkensmeyer (2009). "Review of control strategies for robotic movement training after neurologic injury." J Neuroeng Rehabil **6**(1): 20.
McCain, K. J., F. E. Pollo, et al. (2008). "Locomotor treadmill training with partial body-weight support before overground gait in adults with acute stroke: a pilot study." Arch Phys Med Rehabil **89**(4): 684-91.
Meinders, M., A. Gitter, et al. (1998). "The role of ankle plantar flexor muscle work during walking." Scandinavian Journal of Rehabilitation Medicine **30**(1): 39-46.
Menz, H. B., S. R. Lord, et al. (2003). "Age-related differences in walking stability." Age Ageing **32**(2): 137-42.
Monsell, E. M., J. M. Furman, et al. (1997). "Computerized dynamic platform posturography." Otolaryngol Head Neck Surg **117**(4): 394-8.
Niino, N., S. Tsuzuku, et al. (2000). "Frequencies and circumstances of falls in the National Institute for Longevity Sciences, Longitudinal Study of Aging (NILS-LSA)." Journal of Epidemiology **10**(1 Suppl): S90-4.
Niino, N., S. Tsuzuku, et al. (2000). "Frequencies and circumstances of falls in the National Institute for Longevity Sciences, Longitudinal Study of Aging (NILS-LSA)." J Epidemiol **10**(1 Suppl): S90-4.

Owings, T. M., M. J. Pavol, et al. (2000). "Measures of postural stability are not predictors of recovery from large postural disturbances in healthy older adults." J Am Geriatr Soc **48**(1): 42-50.
Pang, M. Y., T. Lam, et al. (2003). "Infants adapt their stepping to repeated trip-inducing stimuli." Journal of Neurophysiology **90**(4): 2731-40.
Patton, J. L. and F. A. Mussa-Ivaldi (2004). "Robot-assisted adaptive training: custom force fields for teaching movement patterns." IEEE Trans Biomed Eng **51**(4): 636-46.
Patton, J. L., M. E. Stoykov, et al. (2006). "Evaluation of robotic training forces that either enhance or reduce error in chronic hemiparetic stroke survivors." Experimental Brain Research **168**(3): 368-83.
Pearson, K. G. (2000). "Neural adaptation in the generation of rhythmic behavior." Annual Review of Physiology **62**: 723-753.
Pepin, A., K. E. Norman, et al. (2003). "Treadmill walking in incomplete spinal-cord-injured subjects: 1. Adaptation to changes in speed." Spinal Cord **41**(5): 257-70.
Pohl, M., J. Mehrholz, et al. (2002). "Speed-dependent treadmill training in ambulatory hemiparetic stroke patients: a randomized controlled trial." Stroke **33**(2): 553-8.
Prochazka, A., M. Hulliger, et al. (1987). Dynamic and static fusimotor set in various behavioural contexts. Mechanoreceptors: Development, Structure, and Function. P. Hnik, T. Soukup, R. Vejsada and J. Zelena. New York, Plenum Press: 417-430.
Proteau, L., R. G. Marteniuk, et al. (1992). "A sensorimotor basis for motor learning: evidence indicating specificity of practice." Q J Exp Psychol A **44**(3): 557-75.
Proteau, L., L. Tremblay, et al. (1998). "Practice does not diminish the role of visual information in on-line control fo a precision walking task: support for the specificity of practice hypothesis." Journal of Motor Behavior **30**(2): 143-150.
Proteau, L., L. Tremblay, et al. (1998). "Practice does not diminish the role of visual information in on-line control of a precision walking task: support for the specificity of practice hypothesis." Journal of Motor Behavior **30**(2): 143-150.
Pryse-Phillips, W. E. M. and T. J. Murray (1992). Essential Neurology: A Concise Textbook. New York, Elsevier.
Reinkensmeyer, D. J., J. L. Emken, et al. (2004). "Robotics, motor learning, and neurologic recovery." Annual Review of Biomedical Engineering **6**: 497-525.
Reinkensmeyer, D. J. and J. L. Patton (2009). "Can robots help the learning of skilled actions?" Exerc Sport Sci Rev **37**(1): 43-51.
Ringsberg, K., P. Gerdhem, et al. (1999). "Is there a relationship between balance, gait performance and muscular strength in 75-year-old women?" Age Ageing **28**(3): 289-93.
Rumelhart, D. E., G. E. Hinton, et al. (1986). "Learning representations by back-propagating errors." Nature **323**(6088): 533-536.

Ryerson, S. and K. Levitt (1997). Functional Movement Reeducation. New York, Churchill Livingstone.
Ryerson, S. and K. Levitt (1997). Functional Movement Reeducation. New York, Churchill-Livingstone.
Salmoni, A. W., R. A. Schmidt, et al. (1984). "Knowledge of results and motor learning: a review and critical reappraisal." Psychol Bull **95**(3): 355-86.
Sanger, T. D. (2004). "Failure of motor learning for large initial errors." Neural Comput **16**(9): 1873-86.
Sawicki, G. S., A. Domingo, et al. (2006). "The effects of powered ankle-foot orthoses on joint kinematics and muscle activation during walking in individuals with incomplete spinal cord injury." Journal of Neuroengineering and Rehabilitation **3**: 3.
Scheidt, R. A., J. B. Dingwell, et al. (2001). "Learning to move amid uncertainty." J Neurophysiol **86**(2): 971-85.
Scheidt, R. A., D. J. Reinkensmeyer, et al. (2000). "Persistence of motor adaptation during constrained, multi-joint, arm movements." Journal of Neurophysiology **84**(2): 853-862.
Schmidt, R. A. and R. A. Bjork (1992). "New conceptualizations of practice: common principles in three paradigms suggest new concepts for training." Psychological Science **3**: 207-217.
Schmidt, R. A. and T. D. Lee (1999). Motor Control and Learning: A Behavioral Emphasis. Champaign, IL, Human Kinetics.
Schrager, M. A., V. E. Kelly, et al. (2008). "The effects of age on medio-lateral stability during normal and narrow base walking." Gait Posture **28**(3): 466-71.
Seidler, R. D. (2004). "Multiple motor learning experiences enhance motor adaptability." J Cogn Neurosci **16**(1): 65-73.
Shadmehr, R. and F. A. Mussa-Ivaldi (1994). "Adaptive representation of dynamics during learning of a motor task." Journal of Neuroscience **14**(5): 3208-3224.
Shimada, H., S. Obuchi, et al. (2003). "Relationship with dynamic balance function during standing and walking." Am J Phys Med Rehabil **82**(7): 511-6.
Shumway-Cook, A., D. Anson, et al. (1988). "Postural sway biofeedback: its effect on reestablishing stance stability in hemiplegic patients." Arch Phys Med Rehabil **69**(6): 395-400.
Sidaway, B., S. Ahn, et al. (2008). "A comparison of manual guidance and knowledge of results in the learning of a weight-bearing skill." J Neurol Phys Ther **32**(1): 32-8.
Sidaway, B., B. Moore, et al. (1991). "Summary and frequency of KR presentation effects on retention of a motor skill." Res Q Exerc Sport **62**(1): 27-32.
Singer, R. N. and D. Pease (1976). "A comparison of discovery learning and guided instructional strategies on motor skill learning, retention, and transfer." Res Q **47**(4): 788-96.

Sullivan, K. J., B. J. Knowlton, et al. (2002). "Step training with body weight support: effect of treadmill speed and practice paradigms on poststroke locomotor recovery." Archives of Physical Medicine and Rehabilitation **83**(5): 683-91.

Thoroughman, K. A. and R. Shadmehr (2000). "Learning of action through adaptive combination of motor primitives." Nature **407**(6805): 742-7.

Thoroughman, K. A. and R. Shadmehr (2000). "Learning of action through adaptive combination of motor primitives." Nature **407**(6805): 742-747.

Visintin, M., H. Barbeau, et al. (1998). "A new approach to retrain gait in stroke patients through body weight support and treadmill stimulation." Stroke **29**(6): 1122-8.

Visser, J. E., M. G. Carpenter, et al. (2008). "The clinical utility of posturography." Clin Neurophysiol **119**(11): 2424-36.

Waters, R. L. and B. R. Lunsford (1985). "Energy cost of paraplegic locomotion." Journal of Bone and Joint Surgery [American] **67**(8): 1245-50.

Wei, Y., P. Bajaj, et al. (2005). Visual error augmentation for enhancing motor learning and rehabilitative relearning. International Conference on Rehabilitation Robotics, Chicago, IL, IEEE.

Wernig, A. (2005). ""Ineffectiveness" of automated locomotor training." Archives of Physical Medicine and Rehabilitation **86**(12): 2385-6; author reply 2386-7.

Wernig, A. and S. Muller (1992). "Laufband locomotion with body weight support improved walking in persons with severe spinal cord injuries." Paraplegia **30**(4): 229-238.

Wernig, A., S. Muller, et al. (1995). "Laufband therapy based on 'rules of spinal locomotion' is effective in spinal cord injured persons." European Journal of Neuroscience **7**(4): 823-829.

Wernig, A., A. Nanassy, et al. (1998). "Maintenance of locomotor abilities following Laufband (treadmill) therapy in para- and tetraplegic persons: follow-up studies." Spinal Cord **36**(11): 744-749.

Winstein, C. J., P. S. Pohl, et al. (1994). "Effects of physical guidance and knowledge of results on motor learning: support for the guidance hypothesis." Res Q Exerc Sport **65**(4): 316-23.

Winter, D. A. (1990). Biomechanics and Motor Control of Human Movement. New York, John Wiley & Sons.

Winter, D. A. (1991). The biomechanics and motor control of human gait: normal, elderly and pathological. Waterloo, Ontario, Waterloo Biomechanics.

Winter, D. A., A. J. Fuglevand, et al. (1994). "Crosstalk in surface electromyography: theoretical and practical estimates." Journal of Electromyography and Kinesiology **4**(1): 15-26.

Wirz, M., D. H. Zemon, et al. (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): 672-80.

Wolpert, D. M. and Z. Ghahramani (2000). "Computational principles of movement neuroscience." Nat Neurosci **3**(Suppl): 1212-7.

Wolpert, D. M., Z. Ghahramani, et al. (2001). "Perspectives and problems in motor learning." Trends Cogn Sci **5**(11): 487-494.

Woollacott, M. H. and P. F. Tang (1997). "Balance control during walking in the older adult: research and its implications." Phys Ther **77**(6): 646-60.

Wren, T. A., K. P. Do, et al. (2006). "Cross-correlation as a method for comparing dynamic electromyography signals during gait." J Biomech **39**(14): 2714-8.

Wulf, G. and C. H. Shea (2002). "Principles derived from the study of simple skills do not generalize to complex skill learning." Psychon Bull Rev **9**(2): 185-211.

Wulf, G., C. H. Shea, et al. (1998). "Physical guidance benefits in learning a complex motor skill." Journal of Motor Behavior **30**: 367-380.