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Is gait training with the elliptically based robotic gait trainer (EBRGT) feasible in ambulatory patients after stroke?

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Is gait training with the elliptically based robotic gait trainer (EBRGT) feasible in ambulatory patients after stroke?

A dissertation submitted in partial fulfillment of the requirements for the degree of
Doctor of Philosophy at Virginia Commonwealth University

By
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May, 2011
I must start by thanking my generous and loving parents. Without them I wouldn’t have had all of the opportunities that have gotten me to this point. Thank you, mom and dad.

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Is gait training with the elliptically based robotic gait trainer (EBRGT) feasible in ambulatory patients after stroke?

By J. Cortney Bradford, Ph.D.

A dissertation submitted in partial fulfillment of the requirements for the degree of Doctor of Philosophy at Virginia Commonwealth University.

Virginia Commonwealth University, 2011.

Major Directory: Peter E. Pidcoe, PT, DPT, PhD, Associate Professor, Department of Physical Therapy

In response to the potential benefits of task specific training in rehabilitation of gait after stroke and the need for affordable, simple ways to implement it, our group designed the elliptically based robotic gait trainer (EBRGT). A design review of the EBRGT, covering the design goals, an overview of the mechanical and electrical design, and a discussion of the novelty of the device and why it may be beneficial for individuals with hemiparesis secondary to stroke is discussed (Chapter 2).
To characterize the new device, a study was performed to determine if the EBRGT produced a gait pattern that mimicked level surface walking in healthy adults (chapter 3). Sagittal plane kinematic analysis suggested the EBRGT produced joint movement patterns that are similar to level surface walking at the hip and knee with less similarity between activities at the ankle. Electromyography (EMG) revealed that the EBRGT induced a cyclic muscle firing pattern that had some similarities when compared to level surface walking.

We also examined the feasibility of ambulatory individuals after stroke to use the EBRGT and if their movement patterns were similar to healthy adults walking on the same device (Chapter 4). All six participants were able to walk on the device with minimal assistance. These participants had joint kinematics and EMG similar to healthy adults, suggesting that individuals with hemiparesis perform a gait like movement when using the EBRGT.

Lastly, a study was performed to determine if the EBRGT could improve gait parameters and function in ambulatory individuals with hemiparesis after stroke (chapter 5). Four participants walked on the EBRGT 3x/week for 4 or 8 weeks. After the intervention, all 4 participants increased their preferred gait speed. One participant had an improvement in gait speed that indicated functional gains.

The results of this research suggest that the EBRGT can produce a gait pattern that has some similarities to level surface walking and that it is feasible for ambulatory individuals with hemiparesis to use the device. The device may also improve gait
parameters in ambulatory individuals after stroke, but future studies with a control group need to be performed.
Epidemiology

Stroke is one of the leading causes of disability in the United States with 750,000 individuals affected each year (Williams et al., 1999). Each year the cost of stroke is nearly $30 billion in direct medical costs and nearly $20 billion in lost productivity (Mayo, 1993). Many of the people that survive stroke are left with severe disabilities (Roseamond et al., 2007). These deficits include hemiparesis, a weakness on one side of the body, which can impair ambulation. While the majority of stroke survivors will regain some ability to walk, 40% will require assistance with walking and of those who are independent, 60% will still achieve only limited community ambulation (Jorgensen et al., 1995). Fewer than 20% will achieve unlimited community ambulation (Perry et al., 1995). In a study by Lord et al. (2004), over 90% of the subjects who had suffered stroke considered the ability to get out and about in the community to be important and it was
considered essential by 40% of the participants. Restoration of gait is a major goal of rehabilitation for persons with stroke.

Gait description

Walking or gait is a person’s natural way of moving from one location to another. It is also the most convenient and efficient means of traveling short distances. In a normal functioning person, the lower limbs have the ability to adapt to different surfaces and obstacles such as stairs and uneven ground. Because of these advantages of walking, individuals strive to retain the ability to walk even in the presence of severe impairment.

Gait is often defined as a cycle that can be broken up into two major divisions: stance and swing. Stance constitutes 60% of the total gait cycle and is the period when the foot is in contact with the ground. Swing constitutes 40% of the gait cycle and this period begins when the toe is lifted off of the floor. Stance can be divided into three subintervals: Initial double stance, single limb support, and terminal double stance. Initial double support begins with initial contact (IC), where the leg in swing contacts the floor and both feet are in contact with the ground. The next interval, single limb support, begins when the opposite or trailing limb is lifted off of the floor. During this period, the entire body weight is supported by one limb. The stance period ends with terminal double stance. This interval begins with the initial contact of the opposite limb (the limb opposite to the stance limb). The stance period ends and swing begins when the trailing limb is lifted off of the floor. The major divisions of the gait cycle, as well as the
subintervals are depicted in Figure 1. Each limb undergoes its own gait cycle and is
temporally 180° out of phase with the contralateral side. Human gait is quite efficient
and results in minimal off-axis COM deviations (2-3cm vertically and 3-4cm
horizontally) (Perry, 1992). Deviations from this pattern as a result of stroke increase the
energy requirements (Perry, 1992).

Stroke and effects on gait

The pathophysiological basis of stroke is damage to the central nervous system.
This damage is caused by hemorrhage or thrombus affecting the arterial supply of the
brain, creating an ischemic condition that results in tissue death. Damage typically only
occurs in one hemisphere of the brain, affecting motor and sensory functions of the
contralateral side of the body. Patients who suffer a stroke have immediate impairments.
These can impact both motor and sensory pathways and produce symptoms that include:
(1) decreased muscle force generation (the inability to generate voluntary muscle
contractions), (2) decreased motor control, including the inability to appropriately
coordinate muscle firing patterns and (3) decreased proprioception. Stroke can also lead
to impairments that begin days to weeks after the initial injury. These include muscle
spasticity and changes in the mechanical properties of the muscles (Olney and Richards,
1996). All of these impairments can affect gait to varying degrees and contribute to a
loss of functional mobility.
Although deviations from normal gait can vary from patient to patient, evidence of motor impairments is often easily recognized following stroke. Studies have reported the following differences for the paretic limb when compared to the uninvolved side: decreased hip flexion at initial contact, decreased hip extension at toe off, and decreased hip flexion during mid-swing, more knee flexion at initial contact and less knee flexion at toe off and mid swing, and increased ankle plantarflexion at initial contact and mid swing and less plantarflexion at toe off, pelvic hiking and circumduction of the paretic limb (Burdett et al., 1998, Chen et al., 2005). These kinematic changes can also result in toe-drag; a common problem for patients who have suffered stroke (Perry, 1992).

Temporal gait deviations are also often observed in subjects who have suffered a stroke. On the involved (or paretic) side, the duration of the stance phase is longer and thus accounts for a larger percentage of the total gait cycle. On average, only 33% of the cycle is spent in swing (compared to the normal 40%). The duration of stance on the uninvolved side is also typically increased (Olney and Richards, 1996). It has been reported that 80% of the cycle of the unaffected leg is spent during stance (compared to the normal 60%) (Perry, 1969) Also there is an increase in the time spent in double support (Olney and Richards, 1996). The temporal gait deviations are not equal for the involved and uninvolved sides, thus leading to asymmetry in the gait pattern. Temporal asymmetry has been shown to correlate with motor impairment and to a lesser extent to gait velocity, where greater asymmetry correlates with greater motor impairment and a slower velocity in individuals with hemiparesis (Patterson et al., 2008).

Gait in individuals with hemiparesis secondary to stroke is characterized by reduced speed, cadence, and stride length (Olney and Richards, 1996). Decreased
cadence and stride length are consistent with slower walking speeds (Nakamura et al., 1988). Normal healthy adults, 20-65 years old, typically ambulate at a speed of about 1.5 m/s (Sutherland et al., 1994). Average walking speeds for individuals post stroke have been found to range from 0.23 m/s (Burdett et al., 1988) to 0.73 m/s (von Schroeder et al., 1995). It has been demonstrated that walking speed is a good predictor of community ambulation (Perry et al., 1995). Perry et al. (1995) found that a walking speed over 0.8 m/s correlated with unlimited community ambulation, while speeds between 0.4 m/s and 0.8 m/s were correlated with limited community ambulation. Walking speeds less than 0.4 m/s correlated with household ambulation only. Thus, individuals often walk at speeds which limit their community ambulation after stroke.

Individuals with gait deviations secondary to stroke often have decreased endurance in addition to gait disturbances. Only 50% of the higher functioning stroke survivors are typically able to complete a 6-minute walk test (Mayo et al., 1999). A six minute walk test is often employed as a tool to functionally test a patient’s endurance and ambulation speed. The decreased endurance in this population may be due to the mechanical inefficiency associated with hemiparetic gait (Chen et al., 2005). The increase in mechanical energy requirements may be expressed by the patient as reduced endurance or a feeling of early fatigue during ambulation. Low endurance can lead to a very restricted scope of community mobility, reducing the normal activities of daily living (ADL) for these individuals (Lerner et al., 1986). Another possible explanation for reduced performance in the 6-minute walk test is compromised balance. Balance deficits have been associated with low ambulatory activity after stroke (Michael et al., 2005). In fact, stroke survivors may become more sedentary because of balance deficits and an
associated fear of falling. This can result in becoming even further deconditioned (Ryan et al., 2000).

Endurance changes, mechanical inefficiency, and balance are inter-related problems that impact mobility in patients who have suffered stroke. Regardless of the causal relationships, it is important to minimize these effects. Patients with stroke are limited in their ability to function normally, in part due to impaired gait. The rehabilitation of a patient following stroke attempts to return these movements to a more normal gait pattern in an effort to restore function.

Theoretical framework

The current focus for gait restoration after stroke is on task specific, repetitive rehabilitation techniques. The theory assumes that task specific training provides sensory stimulation which promotes neural plasticity that in turn produces improved motor output (Nudo and Friel 1999, 1999). It has been well established that task-specific practice is required for motor learning to occur (Nudo and Friel, 1999, Bayona et al., 2005, Marshall, 1985, Nudo, 2006, Wielock and Nikolich, 2006, Krackauer, 2006, Adkins, 2006). Yen et al. (2008) have shown that task specific training, in comparison to traditional stroke rehabilitation may yield cortical reorganization specific to the corresponding areas being used. It has also been reported that training intensity alone does not account for the differences between traditional stroke rehabilitation and task-specific training (Page, 2003). Repetition of task specific exercises plays an important
role in inducing and maintaining changes in the brain (Nudo and Friel, 1999, Yen et al., 2008). Another theory is that central pattern generators within the spinal cord may be stimulated during task specific training for gait and may help make residual ability more functional (Kautz et al., 2006). Rehabilitation of subjects with neurological impairments has focused on simulations of normal gait kinematics to provide appropriate sensory stimulus for improving motor patterns required for ambulation (Bayona et al., 2005).

Because individuals experience some level of spontaneous recovery after stroke, it is important to note the time since stroke when beginning an intervention (Jorgensen et al., 1995b). Individuals post stroke are often grouped into three groups depending on the time since stroke: acute (≤2 weeks), sub-acute (2-12 weeks), chronic (≥12 weeks).

Task specific training for walking involves the subject, with assistance, moving their lower-limbs in a pattern similar to normal gait while bearing part or all of their body weight. Over-ground gait training, body weight supported treadmill training (BWSTT), and more recently, robotic gait training have been used to administer task specific gait training in patients with stroke.

In body weight supported treadmill training, the patient ambulates on a treadmill while part of their body weight is supported by a harness. In many cases, two or more therapists are needed to help assist the patient weight shift and advance the paretic limb. Sullivan et al. (2007) compared BWSTT to cycling and lower extremity exercise. They found that BWSTT was superior to cycling and lower extremity exercise for improving gait parameters in individuals with stroke. Several other studies have found task specific training, including BWSTT, to be superior to conventional physical therapy (Visintin et al., 1998, Ada et al., 2003, Salbach et al., 2004, Dean et al., 2000, Richards et al., 1993).
In a recent study by Moore et al. (2010), the authors found that additional intensive BWSTT in individuals with chronic stroke improved their daily stepping. However, a systematic review of BWSTT by Moseley et al. (2005) concluded that there was conflicting evidence as to whether or not BWSTT was superior to conventional therapy for improving gait after stroke.

Despite the possible benefits to using BWSTT it has not been widely accepted in the clinic. While the technique employs a harness to off-load body weight, there are still significant physical demands placed on the therapists (Winchester and Querry, 2006). As a solution, robotic devices have been developed to assist stepping. These devices were designed to reduce the physical demands on the therapist during gait training and to make it possible to practice at more intense levels safely (Winchester and Querry, 2006). Most robotic intervention studies to date have employed either the Lokomat® (Dias et al., 2007) or the GaitTrainer® (Hesse and Uhlenbrack, 2000). The Lokomat® consists of two robotic arms that apply torque at the hip and knee bilaterally, while the subject is suspended over a treadmill. The limbs are driven reciprocally to produce a gait like pattern. The GaitTrainer® consists of two footplates and a body weight unloading harness. The footplates guide the patient’s feet in a reciprocal pattern that simulates normal gait. Both devices are successful at reducing the number of therapists required to administer the therapy and reducing the physical demands on the therapist.

Each device has been tested in individuals post stroke with varying degrees of gait impairments and at various time points post stroke. One of the first controlled trials using the GaitTrainer® as a gait training intervention post stroke tested the device in non-ambulatory individuals who had endured a stroke (Werner et al., 2002). The authors
found that in this population of individuals with sub-acute stroke, using the GaitTrainer® improved both gait and function as compared to conventional physical therapy. A larger multi-center trial using the GaitTrainer® was performed by Pohl et al. (2007). This study was also performed in non-ambulatory subjects with sub-acute stroke. After four weeks of using the GaitTrainer®, the authors reported that a significantly higher percent of patients in the experimental group could walk independently relative to the control group. A third trial by Tong et al. (2006), also included only individuals with sub-acute stroke. This trial found that the GaitTrainer® produced larger increases in gait speed and function compared to conventional physical therapy. Only one controlled trial to date has tested the GaitTrainer® in a population of subjects with chronic stroke (Dias et al. 2007). This trial compared conventional therapy with gait training using the GaitTrainer. The authors found that, while both groups saw improvements in gait, there were no significant differences between groups.

The Lokomat® gait trainer has also been tested in patients with both sub-acute and chronic stroke. In a study by Peurala et al. (2005), chronic stroke patients practiced walking in the Lokomat® while a control group practiced walking over ground. The authors found that in both groups, gait speed, dynamic stability, and motor task performance improved significantly, however there was no difference between groups. In another study, individuals with chronic stroke trained on the Lokomat and were compared to a similar group of patients trained using BWSTT (Hornby et al., 2008). The patients in the BWSTT group had a significantly greater increase in gait speed than the group training on the Lokomat®. The authors proposed that the differences between the groups may be because of the increased cardiovascular demands associated with
BWSTT. In a pilot study of a randomized controlled trial, Westlake and Patten (2009) also compared the Lokomat® with BWSTT in 16 subjects with chronic stroke and found no significant difference between groups in terms of gait speed.

The results from studies examining the use of the Lokomat® to restore gait in a group of individuals with sub-acute stroke are just as mixed as the results for chronic stroke survivors (Schwartz et al., 2009, Husemann et al., 2007, Hidler et al., 2009). Schwartz et al (2009) found that the use of the Lokomat restored independent gait in a significantly higher percentage of patients than conventional physical therapy. While, Husemann et al. (2007) showed no difference between Lokomat and conventional therapy, Hidler et al. (2009) found conventional therapy to be superior in improving gait speed. Note that the ambulatory level of patients in the Hidler et al. (2009) study was slightly higher at study onset.

Ada et al. (2010) and Merholz et al. (2007) both performed meta-analyses on the effectiveness of robotic gait trainers in individuals with stroke. These reviews combined data from trials on both the Lokomat® and the GaitTrainer®. When the data are compiled, robotic gait trainers have better outcomes compared to other interventions in terms of restoring gait in non-ambulatory individuals that have had a stroke. However, they are no better than any other treatment at improving gait speed in individuals that are ambulatory after stroke.

In response to the potential benefits of task specific training in rehabilitation of gait after stroke interventions our group designed a new semi-robotic gait trainer. The
elliptically based robotic gait trainer (EBRGT) is both affordable and simple to use and has features that encompass characteristics of both BWSTT and robotic gait training.

Purpose of research

While the newly developed gait trainer is similar to a gait trainer that already exists (Hesse and Uhlenbrock, 2000), the device has yet to be tested in a patient population. A previous version of the device has tested the concept in a normal, healthy adult population which showed movement patterns that somewhat mimicked walking (Bradford et al., 2007). Our device is very similar to commercially available elliptical. One study has tested the feasibility of using elliptical trainers in individuals with hemiparesis and they found the activity safe and feasible (Jackson et al., 2010). There is promising evidence that our device will be easily implemented in a hemiparetic population, however this has yet to be tested. The purpose of this research is to determine the feasibility of using the newly developed gait training device, EBRGT, in individuals post stroke.
Specific aims and hypotheses

Specific aims:

o To determine if the elliptically based robotic gait trainer (EBRGT) produces sagittal plane kinematics that mimic level surface walking in normal healthy adult subjects.

o To determine if the EBRGT produces sagittal plane kinematics that mimic level surface walking in patients with gait deficits secondary to stroke.

o To determine the feasibility of a gait training intervention using the EBRGT in patients with gait deficits secondary to stroke.

o To determine if a 4-week gait training intervention using the EBRGT can improve temporal-spatial gait parameters and Functional Independence Measure (FIM) score in patients with gait deficits secondary to stroke.

o To determine the acceptability of the device by individuals post stroke.
Hypotheses:

a. The EBRGT will produce sagittal plane gait kinematics that mimics level surface walking.

b. The EBRGT will produce sagittal plate gait kinematics that mimic level surface walking in individuals with gait deficits secondary to stroke.

c. Individuals with gait deficits secondary to stroke will see improvements in FIM score, gait speed, stride length, and single limb support time for the affected limb after a 4 week gait training intervention using the EBRGT.
**Figure 1.** Diagram of the gait cycle for walking. One complete gait cycle from initial contact to initial contact of the right lower limb.
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Introduction

This chapter will cover the design goals, an overview of the mechanical and electrical design of the new device, rationale for design choices, and a discussion of the novelty of the device and how it may be beneficial for patients with hemiparesis secondary to stroke.

The design goals for the device were chosen based on previous research and the characteristics of the target population. Previous research has demonstrated that while BWSTT and robotic or mechanically assisted gait training have shown positive results in terms of improving walking ability in patients with gait deficits secondary to stroke (Sullivan et al., 2007, Visintin et al., 1998, Ada et al., 2003, Salbach et al., 2004, Dean et al., 2000, Richards et al., 1993, Ada et al., 2010, Merholz et al., 2007), the methods have not been widely accepted due to cost (Morrison and Backus, 207). The first design goals are to develop a device that simulates level surface walking yet is low-cost relative to what already exists on the market.

While the theory behind task-specific training for gait is that patients practice a gait pattern that mimics ‘normal’ level surface walking to improve walking ability, there
may be a case where not all users of the device can follow a single prescribed pattern. For example, joint range of motion (ROM) can often be limited in this population (Perry, 1992), yet it may still be beneficial to practice walking. For this scenario, another design goal is that the device has an easily adjusted gait pattern. To include a wide range of patients, the final design goal is that the device be able to support up to a 250lb patient.

To suit these design goals, an elliptical trainer was chosen as the base for the design. Elliptical trainers already guide the user through a movement pattern that has some characteristics of gait and it is an affordable design solution to make modifications to an existing device rather than building from the ground up. During use of the elliptical trainer the hip and knee kinematics mimic level surface while the ankle kinematics have a poor relationship with level surface walking (Bradford et al., 2007). To improve the ankle kinematics, which are important for effective walking (Perry, 1992), modifications were made to the footplates of an elliptical trainer.

Features of the new gait training device are novel and set it apart from other devices that exist to help patients with hemiparesis practice walking: (1) it is the first known device that specifically controls ankle articulation while also providing mechanical guidance for the entire lower extremity during walking, (2) the device is powered by the user and only provides guidance of the most distal portion of the lower extremity. The rational for how these features may be beneficial during gait training in a population of individuals with hemiparesis secondary to stroke are discussed later in this chapter.
Design Goals

- The device needs to simulate level surface walking
- The device needs to be easily adjustable to accommodate a variety of gait patterns
- The device needs to be low cost relative to devices that already exist
- The device needs to be able to support at least a 250lbs patient

Device Design

Choosing the elliptical trainer

Prior to the current design, two prototypes addressing this problem were built and tested. An elliptical trainer was chosen as the base for the gait trainer device since it already produced a gait like movement. This was also a cost saving measure since the entire device did not have to be built from the ground up.

Elliptical trainers guide the user’s feet in a reciprocal, elliptical pattern while the user bears full body weight on the lower extremities (Figure 1). However, kinematic analysis (Bradford et al., 2007, Burnfield et al., 2010) has shown that while the hip and knee motions are similar to level surface walking, the ankle joint angles in the sagittal
plane do not mimic level surface walking. Based on this observation, in the first iteration of design, the footplates were articulated to provide the sagittal plane ankle movements called dorsiflexion and plantarflexion. This motion was controlled by a cam-pushrod linkage mechanism. The size, shape, and orientation of the cams were designed so that the footplates mimicked a normal gait (Reese, 2004).

The first design proved to be mechanically deficient in handling the imposed loads of a subject but did provide proof-of-concept (Reese, 2004). More robust materials were used in a second design iteration. This device was tested in a study with 8 normal, healthy subjects (Bradford et al., 2007). Sagittal plane kinematics were captured via video camera and joint angles were plotted against percent of the gait cycle and compared to level surface walking (Figure 2). The foot angles with respect to horizon using the modified elliptical trainer values were not a perfect representation of normal gait, however, they were in phase with normal gait. The non-modified elliptical trainer values were out of phase with normal gait values.

Despite the lack of perfect congruency of the foot and ankle kinematics to normal gait, we decided that it was appropriate to continue using the elliptical trainer as the base for the design because: (1) It has however been shown that doing a weight-bearing gait-like activity is better than a seated cycling or strengthening activities for improving gait parameters in a hemiparetic population (Sullivan et al., 2007). (2) It may be possible to further improve the modifications to the elliptical trainer so that lower extremity joint kinematics better mimic level surface walking. As a result, it seems reasonable to conclude that a weight-bearing gait-like activity is a good modality in the treatment of
gait disorders and that an elliptical trainer is a reasonable choice as the base for this design.

*Modifications to the elliptical trainer*

In the third iteration of the design, the focus of the modifications was again on the footplates and distal control. In order to more accurately mimic level surface walking, the footplate mechanism was redesigned to achieve a greater range-of-motion (ROM). This motion was controlled using a servo-motor based electromechanical system. This design allows the pattern of the footplate articulation to be easily manipulated. There were a number of design constraints driving the modifications. These are numerated below:

1. The device was intended to carry a subject weighing a maximum of 250 lbs. It had to be designed to manage that load in a worst case scenario. One in which the subject placed all of their body weight (BW) on one leg in a location that would maximally stress the mechanics of the design. A previous study on an elliptical trainer with instrumented footplates showed that the maximum force at the footplate never exceeded 1xBW (Lu et al, 2007), therefore the maximum load in our design would be 250lbs. If this load were placed at the maximum leverage point of a single footplate (6in from the axis of rotation for that footplate), the maximum torque would be 125 ft-lbs. This is depicted in the free body diagram in Figure 3. This maximum torque computation assumes that the mass of
the footplate is negligible as its weight is a small fraction of the load created by the subject standing on the footplate.

2. The footplates needed to have at least 30° of total excursion in the sagittal plate to mimic level surface walking. This is the typical range of motion at the ankle during level surface walking (Perry, 1992).

3. Maximum rotational speed of footplate needed to match normal level surface walking. Average walking cadence is 1 step/second = 1 full cycle (stride) every 2 seconds. Based on normal gait joint angle data (Winter DA, 1990), at that gait speed, the maximum angular velocity of the foot during normal gait is 22rpm. Thus the footplate, which will be guiding the foot in a physiologic pattern, needs to be able to rotate at that rate.

_Mechanical modifications_

A new footplate and footplate mounting apparatus were designed to meet the previously defined design specifications. To accomplish functional articulation, the footplate (Figure 4) needed to be elevated above elliptical ski to provide additional clearance for the required sagittal plane range-of-motion (ROM). The ski also needed to be stiffened to provide a more rigid mechanical linkage between the footplate and the
proposed electromechanical drive mechanism. Aluminum was used for most of the new parts as it strong, but light, minimizing addition of mass to the system.

Solid models (Figure 5) of the new parts were created to ensure the parts met our design specifications before fabrication. Table 1 provides the materials list and a legend for the parts labeled in Figure 5. A 60:1 ratio zero backlash worm gear box (item 7) was selected for its high gear ratio and its self-locking abilities. A high gear ratio was needed in order to reduce the size of the motor needed to drive the footplate motion, due to limitations on space and weight. The push-pull rod mechanism (items 19, 20, and 21) was used to transfer the motion of the gearbox to the footplate. This provided the greatest range of motion while allowing the drive components (motor and gear box) to be placed in line with, and posterior to the footplate, keeping the drive components in a place where they would not interfere with use of the device. The rendering below only represents the right side modifications to the elliptical trainer. A mirror image of these parts was created for the left side of the device.

The implemented modifications to the elliptical are shown in Figure 6. Note that the mounting plate is attached to the peg supporting the back of the existing ski. The front of the mounting plate is bolted directly to ski.

Controller design

The motion of each footplate is controlled by a separate single axis controller. To link the motion of the left and right footplate, each controller receives feedback from a
common optical encoder mounted on the axis of the elliptical trainer flywheel (Crank in figure 1). Figure 7 illustrates the control schematic. Recall that the flywheel/crank drives the motion of both skis, with the left and right ski $180^\circ$ out of phase. The motion of the footplate was controlled as follows: The user drives the motion of the elliptical by applying force to the footplates. This causes the flywheel to start turning. The encoder provides feedback about the position of the flywheel to the controllers and the position of the footplate (orientation in the sagittal plane) is adjusted based on the position of the flywheel. We synchronized specific events in a normal gait cycle to the position of the flywheel on the elliptical so that a heel strike position occurs when the ski is in the most forward position and toe off occurs when the ski is in the most rearward position. For a detailed parts list of electrical components see Appendix A. For controller programming, see Appendix B.

Two single axis controllers were chosen for use in this design. These controllers were mounted on the base at the rear of the elliptical trainer. Figure 8 illustrates the mechanical and electrical modifications.

As a safety precaution, an emergency stop button was placed on the console of the elliptical trainer (Figure 9). This would allow the subject or operator to stop the motion of the footplates if necessary. Mechanical, end of travel limit switches were also implemented to limit the range of motion of the footplate. These were wired using an active logic circuit. A broken wire or switch would engage the safety protocol and stop the device. The motor powering the footplates was also torque and error limited to provide another level of safety via the controlling software.
Figure 10 illustrates the new device during use. As the user applies force to the footplates, moving the flywheel of the elliptical around, the footplate is moved to a position that corresponds to the flywheel position. Starting at the far left of figure 10, the first position corresponds with initial contact. Here the flywheel has rotated so that the support peg of the ski is in the most anterior position. The footplate tipped the toe upward so that the heel ‘strikes’ first at initial contact. Next the flywheel has rotated so that the support peg of the ski is at bottom-dead-center and corresponds with mid-stance. The footplate is approximately parallel to floor, since the foot would be flat on the floor at this instance in level surface walking. The third position, where the support peg is in the most posterior position, corresponds to toe off. The front of the footplate is tipping downward here simulating level surface walking where the heel leads the toe as the foot leaves the ground and the limb beings swing phase. The final position, far right of figure 10, corresponds with mid-swing. Here the ankle is in a neutral position so that the toe ‘clears the ground’ as the limb is advanced.

Discussion

Low cost relative to other gait training devices and techniques

Body weight supported treadmill training has been shown to improve gait in hemiparetic stroke subjects (Sullivan et al., 2007, Visintin et al.,1998, Ada et al.,2003), however it has failed to become widely used because of the high cost associated with its
implementation (the number of staff required to perform the treatment) (Morrison and Backus, 2007). Robotic devices have been developed that only require the assistance of one skilled therapist during gait training (Winchester and Querry, 2006). However, a major disadvantage of these robotic gait trainers is again, the high cost of the device. While the number of staff required to perform the therapy is reduced, these devices can still be too expensive to be utilized in a clinical setting. The Lokomat®, for example, starts at a price of $195,000 (From Hocoma pricing brochure).

Our solution to this problem was to create a low-cost robotic device that has the ability to simulate normal gait. The platform for our design was a commercially available elliptical trainer ($600) coupled with a body-weight support system. The modifications to elliptical totaled around $8000, bringing the total to well under $10,000. This is much less than the cost of the Lokomat® gait trainer (Starting at $195,000).

The design also maintains a low cost of implementation by reducing the number of skilled staff needed to administer therapy during training on the device (Morrison and Backus, 2007). BWSTT requires at least two skilled therapists and one non-skilled assistant to operate. Our device is designed so that only one skilled therapist is required. There is one other known device on the market that uses a distal control mechanism and this device only requires one skilled therapist during its operation (Werner et al., 2002). We expect similar results. The therapist’s responsibilities should only be to monitor the subject during training and exercise.
Distal control/Controlled ankle articulation

Another feature that makes our design unique is that it specifically controls ankle articulation. Most of the other robotic gait training devices that have been developed focus on controlling the movement of the hip and knee (Winchester and Querry, 2006). One other device uses a distal control strategy, similar to our device, where the foot is guided and the hip and knee are expected to follow in a normal gait pattern (Hesse and Uhlenbrock, 2000). However, this device does not control ankle articulation in a physiological manner; it only controls the path of the foot though space, not the orientation. There may be some benefit to controlling the articulation of the ankle by exploiting stretch reflexes.

It has been shown that the stretch reflex makes an important mechanical contribution to the joints they influence (Toft et al., 1991, Mrachacz-Kersting and Sinkjaer 2003). It has also been shown that reflexes are modulated depending on the movement/task being performed (Faist et al., 1999, Knikou et al., 2006, Stein and Thompson, 2006) and are also modulated depending on phase during cyclic movements such as walking (Duysens et al., 2000, Larsen et al., 2006, Mrachacz-Kersting et al., 2004, Dietz et al., 1990). The behavior of the stretch reflex of the quadriceps muscle has specifically been addressed in a few studies during locomotion (Larsen et al., 2006, Marchacz-Kersting et al., 2004, Dietz et al., 1990, Garret and Luckwill, 1983). A study by Mrachacz-Kersting et al. (2004) demonstrated that the highest reflex responses in the knee extensors occur during the stance phase of the gait.
cycle when the stability of the knee is of major importance to maintain balance. While it has been shown that the stretch reflex of the quadriceps is modulated during gait (Mrachacz-Kersting et al., 2004), the exact mechanism has not been determined. However, in study by Koceja et al. (1990) it was shown that a conditioning tap to the Achilles tendon had an excitatory effect on the quadriceps (knee extensor) muscles.

In our design, we control the articulation of the ankle thus ensuring that dorsiflexion is achieved just before initial contact in the gait cycle, as would be the case during normal gait (Perry, 1992). By eliciting dorsi-flexion similar to that seen in normal gait just before heel strike where the knee is extended, the ankle plantarflexors are stretched. This mechanical stimulation would be similar to the Achilles tendon tap stimulation used in the experiments by Koceja et al. (1990), where in both cases the Achilles tendon is being stretched. If this results in the response seen in their experiments, there will be an excitatory effect on the quadriceps muscles, facilitating torque production in the knee extensors. This may be beneficial to a stroke patient who may have limited quadriceps muscle control and force production capability resulting in the individual being unsteady during weight acceptance just after initial contact (Olney and Richards, 1996). Weak ankle dorsiflexor muscles may also prevent them from lifting their toes through swing phase and prior to heel strike. This inability to elicit the necessary muscle contraction for gait compromises the efficiency of the movement. Controlling the relative position of the foot, shank, and thigh may allow modulation of the stretch reflex to improve performance. However, this may not be beneficial for an individual with spastic paresis.
Unlike other robotic gait training devices that have been developed (Winchester and Querry, 2006), our robotic gait trainer is designed such that the user must power it themselves. This is achieved in a manner similar to a standard elliptical, where the user exerts a force on the footplates that in turn causes the flywheel to turn. The flywheel is connected to each footplate via a leaf spring ski. Since there is a central flywheel to which both of the footplates are linked, the motion of one footplate is related to the other. One footplate is always 180° out of phase with the other, just as one foot is 180° out of phase with the other during level surface walking. Other robotic gait training devices that have been developed drive the user’s lower extremities in a normal gait pattern (Winchester and Querry, 2006). This means that the subject may not have to contribute much, if at all, to the movement.

One potential benefit of the user having to power the elliptical is cardiovascular conditioning. Patients with gait deficits resulting from stroke have been shown to become deconditioned due to inactivity (Ryan et al., 2000). A standard elliptical trainer has been shown to produce similar physiological improvements when compared to a treadmill and stair climber (Equana and Donne, 2004). Stroke survivors have been shown to tolerate treadmill exercise and benefit from cardiovascular conditioning (Macko et al., 2005). Our device operates very similar to a standard elliptical trainer except that the footplates are articulated in a more physiologic pattern. This suggests that
hemiparetic subjects may see cardiovascular improvements in addition to functional gains as a result of ambulating on the modified elliptical trainer.

Another benefit of the user driving the gait trainer may be that the modified elliptical trainer exploits inter-limb coupling. During human walking, a coordination of muscle activation between the two legs seems to be achieved by neuronal coupling within the spinal cord (Dietz et al., 2002). During gait, a perturbation of one leg evokes a purposeful response of both legs which is thought to be mediated by the spinal cord (Berger et al., 1984). In a more recent study, Ferris et al. (2004) recorded EMG, joint kinematics, and vertical ground reaction forces while subjects with clinically complete SCI stepped on a treadmill with manual assistance and partial body weight support. While they stood on one leg, they stepped with the contralateral leg. The authors found that this produced rhythmic EMG patterns in the non-stepping leg, suggesting that there is a coupling between the lower limbs.

While there is damage to the brain as a result of stroke, the spinal cord remains intact. This means that central pattern generators located in spinal cord remain undamaged. However, the supra-spinal centers that initiate and modulated these central pattern generators may be lost or damaged (Enoka, 2002). In a study by Fujiwara et al. (1999), individuals with post-stroke hemiplegia were asked to perform knee flexions with the un-affected (contralateral) limb while the EMG of the affected (ipsilateral) limb was measured. The results showed an increase in ipsilateral (affected) rectus femoris and tibialis anterior muscle activity relative to a voluntary knee extension of that same limb. Kautz et al. (2006) also found that during unilateral pedaling in individuals with hemiparesis secondary to stroke, a rhythmic alternating muscle activity pattern was
evoked in the non-pedaling leg. Cunningham et al. (2002) found improvement in paretic arm performance with bilateral versus unilateral tasks. These results suggest inter-limb coupling in subjects with stroke. However, other studies have shown that inter-limb coupling after stroke is disrupted (Lewis and Byblow, 2004, Rice and Newell, 2001).

Since the user has to drive the modified elliptical trainer, the effects of inter-limb coupling may augment the muscle activity in the paretic limb. As the user drives the elliptical with their unaffected limb, the muscle activity of the paretic limb may be modulated or enhanced by the activity of the unaffected limb, thus creating a more normal EMG pattern that is thought to be important for task-specific training.
## Table 1. Bill of materials for new design.

<table>
<thead>
<tr>
<th>Item #</th>
<th>Description</th>
<th>Material</th>
<th>Total Count</th>
</tr>
</thead>
<tbody>
<tr>
<td>14</td>
<td>New footplate</td>
<td>3/8&quot; Aluminum plate</td>
<td>2</td>
</tr>
<tr>
<td>12</td>
<td>Footplate cushion block</td>
<td>3/4&quot; Aluminum block</td>
<td>4</td>
</tr>
<tr>
<td>2</td>
<td>Existing ski assembly</td>
<td></td>
<td>2</td>
</tr>
<tr>
<td>19</td>
<td>Arm of push-pull assembly</td>
<td>1/4&quot; Stainless steel plate</td>
<td>2</td>
</tr>
<tr>
<td>22</td>
<td>Fin</td>
<td>1/2&quot; Aluminum plate</td>
<td>2</td>
</tr>
<tr>
<td>21</td>
<td>Push-Pull Rod</td>
<td>1/2&quot; Stainless steel threaded</td>
<td>2</td>
</tr>
<tr>
<td></td>
<td></td>
<td>rod</td>
<td></td>
</tr>
<tr>
<td>20</td>
<td>Clevis</td>
<td>Steel</td>
<td>4</td>
</tr>
<tr>
<td>9</td>
<td>Mounting plate support</td>
<td>3/8&quot; Aluminum plate</td>
<td>4</td>
</tr>
<tr>
<td>5</td>
<td>Mounting plate</td>
<td>3/8&quot; Aluminum Plate</td>
<td>2</td>
</tr>
<tr>
<td>7</td>
<td>60:1 Worm gear box</td>
<td>Steel</td>
<td>2</td>
</tr>
<tr>
<td>8</td>
<td>Servo motor</td>
<td></td>
<td>2</td>
</tr>
<tr>
<td>16</td>
<td>Motor Mount</td>
<td>Aluminum</td>
<td>2</td>
</tr>
<tr>
<td>17</td>
<td>Ballbearing</td>
<td>Steel</td>
<td>4</td>
</tr>
</tbody>
</table>
Figure 1. The relationship between walking on a level surface (top row) and walking on an elliptical trainer (bottom row) during each phase of gait. Reprinted from Burnfield et al. (2010).
Figure 2. Sagittal plane foot angle with respect to horizon. Modified (black line) and non-modified elliptical (dark grey line) values are an average of all 8 subjects (Bradford et al., 2007). Normal gait data (light grey line) is from Winter (1990).
**Figure 3.** Footplate with applied loads and single axis of rotation.
Figure 4. The commercially available elliptical trainer before modifications.
Figure 5. Solid model of new components for modifications to the elliptical trainer.
Item #2 is the existing ski of the commercially available elliptical trainer (See Figure 4).
See Table 1 for a list of parts and materials.
Figure 6. Right side of elliptical trainer with modifications.
Figure 7. Control block diagram for modified elliptical gait trainer. An independent controller provides movement commands to the motor driving either the left or right footplates. Each motor is equipped with an encoder, which sends feedback to the controller about whether the motor reached the commanded position. The controller determines where to position the motor, and thus the footplate, based on input from a central encoder tracking the position of the elliptical trainer’s flywheel.
Figure 8. Elliptical trainer with electromechanical modifications.
Figure 9. Top view of the console of the modified elliptical trainer (User’s view). Note the illuminated buttons in the upper right corner of the figure. The red button is an emergency stop. When depressed, both the left and right footplate motion is stopped.
Figure 10. Snapshots of the right side of the user while the device is in operation. Plate 1 (Upper left) – Right side initial contact. Plate 2 (Upper right) – Right side mid-stance. Plate 3 (lower left) – Right side toe off. Plate 4 (Lower right) – Right side mid-swing. The orange line highlights the footplate/foot motion.
Works Cited


Appendix
## Appendix A

### Electro-mechanical parts list

<table>
<thead>
<tr>
<th>Supplier</th>
<th>Part #</th>
<th>Item Description</th>
<th>Quantity</th>
</tr>
</thead>
<tbody>
<tr>
<td>Parker</td>
<td>AR-04CE</td>
<td>Aries Series Servo Drive/Controller</td>
<td>2</td>
</tr>
<tr>
<td></td>
<td></td>
<td>• 120-240VAC single phase input</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>• 400W output</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>• 7 inputs/ 4 outputs</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>• Ethernet communications</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>• Includes Cam function</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>• Encoder Feedback</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>• CE (EMC &amp;LVD), UL recognized</td>
<td></td>
</tr>
<tr>
<td>Parker</td>
<td>VM25</td>
<td>I/O breakout with terminal block</td>
<td>2</td>
</tr>
<tr>
<td>Parker</td>
<td>MPP0921C3E-NPSN</td>
<td>Servo Motor with Smart Encoder</td>
<td>2</td>
</tr>
<tr>
<td>Parker</td>
<td>P-1A1-10</td>
<td>10 ft Motor Power Cable</td>
<td>2</td>
</tr>
<tr>
<td>Parker</td>
<td>F-1A1-10</td>
<td>10 ft Encoder Feedback Cable</td>
<td>2</td>
</tr>
<tr>
<td>Omron</td>
<td>S8VM-05024CD</td>
<td>24VDC, 2.2 Amp power supply</td>
<td>1</td>
</tr>
<tr>
<td>DynaPar;</td>
<td>HA5252500033M</td>
<td>Flange Mounted Rotary Encoder</td>
<td>1</td>
</tr>
<tr>
<td></td>
<td></td>
<td>• 2500 lines/rev differential encoder</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>• Input voltage: 5-26VDC</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>• Output: 5V differential signal</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>• Includes 10ft cable</td>
<td></td>
</tr>
<tr>
<td>IDEC</td>
<td>FB2W-211Z</td>
<td>Two button Enclosure</td>
<td>1</td>
</tr>
<tr>
<td>IDEC</td>
<td>HW1L-MF2F11QD-G-24V</td>
<td>22mm Illuminated Green Push Button with NO/NC contacts</td>
<td>1</td>
</tr>
<tr>
<td>IDEC</td>
<td>HW1L-MF2F11QD-R-24V</td>
<td>22mm Illuminated Red Push Button with NO/NC contacts</td>
<td>1</td>
</tr>
<tr>
<td>Omron</td>
<td>D2SW-01L1MS</td>
<td>Snap action switch with hinge lever</td>
<td>4</td>
</tr>
</tbody>
</table>
APPENDIX B

Left Footplate Controller Programs

PROGRAM 0 – MAIN MOTION
PROGRAM
PBOOT
DETACH
ATTACH MASTER0
ATTACH SLAVE0 AXIS0 "L"
PPU L 8000.0000
AXIS0 EXC (5,-5) : REM set excess error limits
SET BIT8469 : REM enable EXC response
TLM L7 : REM set torque limit to +- 7V
REM Axis Gains values
AXIS0 PGAIN 0.008
AXIS0 IGAIN 0
AXIS0 ILIMIT 0
AXIS0 IDELAY 0
AXIS0 DGAIN 0.0001
AXIS0 DWIDTH 0
AXIS0 FFVEL 0
AXIS0 FFACC 0
AXIS0 TLM 10
AXIS0 FBVEL 0
REM Axis Limits
AXIS0 HLBIT 1
AXIS0 HLDEC 100
HLIM L3
SET BIT16144
SET BIT16145
CLR BIT16146
SET BIT16148
SET BIT16149
AXIS0 SLM (20,-20)
AXIS0 SLDEC 100
SLIM L3
SET BIT16150
SET BIT16151
REM MOTION PROFILE
REM the desired master acceleration
ACC 100
REM the desired master deceleration ramp
DEC 100
REM the desired master stop ramp (deceleration at end of move)
STP 250
REM the desired master velocity
VEL 10
REM the desired acceleration versus time profile.
JRK 0
JOG VEL L 1
JOG ACC L 25
JOG DEC L 25
REM BEGIN HOMING SEQUENCE
CLR BIT136
PRINT "Press green button To start homing, press red button To stop at any time"

_MAIN1
IF (NOT BIT 3 AND NOT BIT 136) THEN GOTO HOMING : REM IF GREEN BUTTON IS DEPRESSED AND BIT 136 IS NOT YET SET START HOMING
IF (BIT 136 and NOT BIT 137) THEN GOTO CAMMING : REM IF BIT 136 (USER DEFINED = HOMING COMPLETE) IS SET, START CAMMING
IF (BIT 8467) THEN CLR BIT 136 : REM IF A KILL ALL MOTION FLAG IS SET (8467) THEN CLEAR BIT 136 AND TURN THE CAM OFF
IF (NOT BIT 3 AND NOT BIT 0) THEN END : REM IF BOTH THE RED AND GREEN BUTTON ARE DEPRESSED SIMULTANEOUSLY, END THE PROGRAM
GOTO MAIN1

_HOMING
PRINT "BEGIN HOMING"
BIT798=0 : REM CHECK JOG LIMITS WHEN JOGGING FWD/REV
JOG VEL L1 : REM SET JOG VELOCITY TO 1 REV/S
DRIVE ON L
CLR 8467
JOG FWD L
PRINT " JOGGING IN POSITIVE DIRECTION "
INH -792 : REM WAIT UNTIL MOTION HAS STOPPED
PRINT " POSITIVE LIMIT SWITCH FOUND "
CLR 8467 : REM CLEAR KILL ALL MOVES FLAG THAT IS SET WHEN A LIMIT IS REACHED
JOG REV L
PRINT " JOGGING IN NEGATIVE DIRECTION "
INH -792
PRINT " NEGATIVE LIMIT SWITCH FOUND "
PRINT " ZERO POSITION AT NEG SWITCH "
CLR 8467
JOG INC L6.30
PRINT " MOVING TO OFFSET POSITION "
INH -792
PRINT " AT OFFSET POSITION"
JOG RES L0
RES L0
PRINT " ZERO POSITION REGISTER AT HOME POSITION "
'PRINT " MOVING To START POSITION "
'JOG INC L-1.34
'INH -792
SET BIT 136
clr BIT 137
GOTO MAIN1
REM START THE CAMMING
_CAMMING
AXIS0 EXC (5,-5) : REM set excess error limits (0.01 is about 5 deg of motor rotation,
less than .1 for footplate)
'SET BIT8469 : REM enable EXC response
'drive on L
DIM LA(1) : REM Dimension 1 long arrays
DIM LA0(69) : REM LAO has 69 elements
PRINT "SLOWLY MOVE FLYWHEEL FORWARD UNTIL THE FOOTPLATES BEGIN MOVING"

INTCAP AXIS0 10 : REM arms capture of axis0 position when HS inp 4 rises (designated by 10)

INH 777: REM wait for flag 777 to be set (flag 777 is set when inp 4 trips intcap)

ecl res -4134 : REM resets encoder to -3700 so it is zero at BDC on the right.

PRINT "Index detected. Encoder reset."

CAM DIM L1 : REM Define 1 cam segments

CAM SEG L(0,10000,LA0) : REM Define cam segment range and source

CAM SCALE L (1/1000) : REM scales cam output back to revolutions

CAM SRC L1 : REM Define cam source as ENC1

CAM SRC RES : REM resets the cam source to 0

set bit 137
_loop

IF (p6160 = 0) THEN CAM ON L

IF (BIT 790) THEN GOTO MAIN1: REM Start camming

GOTO loop
ENDP

PROGRAM 3 — EMERGENCY STOP

PROGRAM

'Program 3 - emergency stop

"TODO: edit your program here

PBOOT

_stop

If (not bit0) THEN set bit 8467
goto stop

ENDP
Appendix C

Right Footplate Controller Programs

Program 0 – Main Motion

PROGRAM
PBOOT
DETACH
ATTACH MASTER0
ATTACH SLAVE0 AXIS0 "R"
PPU R 8000.0000
AXIS0 EXC (0.1,-0.1) : REM set excess error limits (0.01 is about 5 deg of motor rotation, less than .1 for footplate)
SET BIT8469 : REM enable EXC response
TLM R6 : REM set torque limit to +- 2 V
REM Axis Gains values
AXIS0 PGAIN 0.008
AXIS0 IGAIN 0
AXIS0 ILIMIT 0
AXIS0 IDELAY 0
AXIS0 DGAIN 0.0001
AXIS0 DWIDTH 0
AXIS0 FFVEL 0
AXIS0 FFACC 0
AXIS0 TLM 10
AXIS0 FBVEL 0
REM Axis Limits
AXIS0 HLBIT 1
AXIS0 HLDEC 100
HLIM R3
SET BIT16144
SET BIT16145
CLR BIT16146
SET BIT16148
SET BIT16149
AXIS0 SLM (20, -20)
AXIS0 SLDEC 100
SLIM R3
SET BIT16150
SET BIT16151
REM MOTION PROFILE
REM the desired master acceleration
ACC 100
REM the desired master deceleration ramp
DEC 100
REM the desired master stop ramp (deceleration at end of move)
STP 250
REM the desired master velocity
VEL 10
REM the desired acceleration versus time profile.
JRK 0
JOG VEL R 1
JOG ACC R 25
JOG DEC R 25
REM BEGIN HOMING SEQUENCE
CLR BIT136
PRINT "Press green button To start homing, press red button To stop at any time"
_Main1
IF (NOT BIT 3 AND NOT BIT 136) THEN GOTO HOMING : REM IF GREEN BUTTON IS DEPRESSED AND BIT 136 IS NOT YET SET START HOMING
IF (BIT 136 and NOT BIT 137) THEN GOTO CAMMING : REM IF BIT 136 (USER DEFINED = HOMING COMPLETE) IS SET, START CAMMING
IF (BIT 8467) THEN CLR BIT 136 : REM IF A KILL ALL MOTION FLAG IS SET (8467) THEN CLEAR BIT 136 AND TURN THE CAM OFF
IF (NOT BIT 3 AND NOT BIT 0) THEN END : REM IF BOTH THE RED AND GREEN BUTTON ARE DEPRESSED SIMULTANEOUSLY, END THE PROGRAM
GOTO MAIN1
_HOMING
PRINT "BEGIN HOMING"
BIT798=0 : REM CHECK JOG LIMITS WHEN JOGGING FWD/REV
JOG VEL R1 : REM SET JOG VELOCITY TO 1 REV/S
DRIVE ON R
CLR 8467
JOG FWD R
PRINT " JOGGING IN POSITIVE DIRECTION "
INH -792 : REM WAIT UNTIL MOTION HAS STOPPED
PRINT " POSITIVE LIMIT SWITCH FOUND"
CLR 8467 : REM CLEAR KILL ALL MOVES FLAG THAT IS SET WHEN A LIMIT IS REACHED
JOG REV R  
PRINT " JOGGING IN NEGATIVE DIRECTION "  
INH -792  
PRINT " NEGATIVE LIMIT SWITCH FOUND "  
PRINT " ZERO POSITION AT NEG SWITCH "  
CLR 8467  
JOG INC R6.61  
PRINT " MOVING TO OFFSET POSITION "  
INH -792  
PRINT " AT OFFSET POSITION"  
JOG RES R0  
RES R0  
PRINT " ZERO POSITION REGISTER AT HOME POSITION"  
SET BIT 136  
clr BIT 137  
GOTO MAIN1  
REM START THE CAMMING  
\_CAMMING  
AXIS0 EXC (5,-5) : REM set excess error limits (0.01 is about 5 deg of motor rotation, less than .1 for footplate)  
DIM LA(1) : REM Dimension 1 long arrays  
DIM LA0(69) : REM LAO has 69 elements  
PRINT "SLOWLY MOVE FLYWHEEL FORWARD UNTIL THE FOOTPLATES BEGIN MOVING"  
INTCAP AXIS0 10 : REM arms capture of axis0 position when HS inp 4 rises (designated by 10)  
INH 777: REM wait for flag 777 to be set (flag 777 is set when inp 4 trips intcap)  
enc1 res -4134 : REM resets encoder to -3700 so it is zero at BDC on the right.
set bit 138
PRINT "Index detected. Encoder reset."
CAM DIM R1 : REM Define 1 cam segments
CAM SEG R(0,10000,LA0) : REM Define cam segment range and source
CAM SCALE R (1/1000) : REM scales cam output back to revolutions
CAM SRC R1 : REM Define cam source as ENC1
CAM SRC RES : REM resets the cam source to 0
set bit 137
_loop
IF (p6160 = 0) THEN CAM ON R
IF (BIT 790) THEN GOTO MAIN1: REM Start camming
GOTO loop
ENDP

Program 1 - Emergency stop

PROGRAM
'Program 1
'TODO: edit your program here
pboot
MAIN
CLR BIT777
INTCAP AXIS0 10
WHILE (NOT BIT777)
IF (NOT BIT0) THEN SET BIT8467
WEND
SET 32
P0=P6916

WHILE (P6916 < (P0 + 500))

IF (NOT BIT0) THEN SET BIT8467

WEND

CLR 32

GOTO MAIN

ENDP
Chapter 3: Does the newly developed elliptically based robotic gait trainer (EBRGT) produce a gait pattern that mimics level surface walking?

Abstract

Restoration of gait is a major goal of rehabilitation for persons with hemiparesis following stroke. A current focus for gait restoration after paresis associated with stroke is on task-specific, repetitive rehabilitation techniques. Methods to administer this type of training are either physically demanding for the therapists or very expensive to implement. Our solution to this problem was to create a low-cost robotic device that has the ability to simulate walking. The purpose of this study was to validate the newly developed elliptically-based robotic gait trainer (EBRGT) by determining the extent to which normal gait is simulated in a healthy adult population when walking on the device. Twenty normal, healthy adult subjects (mean age=33.9 ± 12.2, range=19-58) were asked to walk at a self-selected pace on both the EBRGT and a treadmill while kinematic and electromyographic (EMG) data were recorded. Simple linear regressions were performed for the sagittal plane hip, knee, ankle, and foot with respect to horizon (foot) joint angles between walking on the treadmill and the EBRGT. Hip and foot angles had high correlations, indicating strong linear relationships between the two conditions. The knee
and ankle angles had poor relationships between conditions. The EMG data were integrated separately for 7 gait phases: Initial loading (il), mid-stance (mst), terminal stance (tst), pre-swing (psw), initial swing (isw), mid-swing (msw), and terminal swing (tsw). Vastus Lateralis (VL) muscle activity was significantly different between the two devices during 6 of 7 phases. The VL exhibited a greater amount of activity during the entire gait cycle during EBRGT walking, but the timing of peak EMG was similar between conditions. The Biceps Femoris (BF), Tibialis Anterior (TA) and Lateral Gastrocnemius (LG) muscles had 4 or more phases where the EMG was not significantly different between conditions. The EBRGT produced similar kinematics at the foot and hip when compared to treadmill ambulation. It also resulted in some major muscle groups in the lower extremity firing in a dynamic pattern that was somewhat similar to treadmill walking. The significance of these findings in the rehabilitation of patients who have suffered stroke is currently under investigation. The device is currently being tested in individuals that have hemiparesis secondary to stroke.

Introduction

Stroke is one of the leading causes of disability in the United States with 750,000 individuals affected each year (Williams et al., 1999). Many of the individuals who survive stroke are left with severe disability (Rosamond et al., 2007). These deficits include hemiparesis, a weakness on one side of the body, which can impair their ability to walk. In a study by Lord et al. (2004), over 90% of the individuals who had suffered
stroke considered the ability to get out and about in the community to be important and it was considered essential by 40% of the participants. Restoration of gait is a central goal for rehabilitation of persons following stroke.

The current focus for gait restoration after stroke is on task specific, repetitive rehabilitation techniques. Task specific training for walking involves the individual, with assistance, moving their lower-limbs in a pattern similar to normal gait while bearing part or all of their body weight. Over-ground gait training, body weight supported treadmill training (BWSTT), and more recently, robotic gait training have been used to administer task specific gait training in patients with stroke. Body-weight supported treadmill training has been shown to improve gait in hemiparetic stroke subjects (Sullivan et al., 2007) and may result in better outcomes than over ground or conventional physical therapy (Visintin et al., 1998, Ada et al., 2003, Salbach et al., 2004, Dean et al., 2000, Richards et al., 1993).

BWSTT also has the potential to improve cardiovascular fitness. Unfortunately, it has failed to become widely accepted because of the high staffing costs required to perform the treatment (Morrison and Backus, 2007). The field of robot-assisted motor rehabilitation has emerged as an alternative to therapist-assisted training. In this type of training, a machine guides the lower extremities in a walking like pattern while the body weight of the individual is supported. Unfortunately, these interventions are often not available due to the cost and complexity of the devices.

In response to the potential benefits of task specific training in rehabilitation of gait after stroke and the need for affordable, simple interventions our group designed a
new semi-robotic gait trainer. The gait trainer is both affordable and simple to use and has features that encompass characteristics of both BWSTT and robotic gait training.

The platform for the new device is a commercially available elliptical trainer. The elliptical trainer was chosen for three reasons 1) studies have shown that sagittal plane hip and knee kinematics correlate well with sagittal plan kinematics during level surface walking (Burnfield et al., 2010, Bradford 2007), thus it has features that mimic gait, 2) elliptical trainers have a mechanical linkage between the left and right sides, similar to a bicycle, thus patients with hemiparesis could help advance their affected limb with their unaffected lower extremity and 3) it employs a distal control mechanism to help guide the lower limbs, but still requires the subject to control their trunk position, including weight shifting.

The Elliptically Based Robotic Gait Trainer (EBRGT) was developed in our lab by significantly modifying a NordicTrack CXT 910 elliptical trainer. The goal was to develop a device that would guide the user’s lower extremity in a pattern that mimics level surface walking while still requiring effort from the user. Studies have shown that the in the sagittal plane, hip and knee kinematics during elliptical trainer use has a good correlation with joint kinematics during level surface walking (Burnfield et al., 2010, Bradford et al., 2010). However, ankle kinematics during elliptical training have a poor correlation to level surface walking (Burnfield et al., 2010). Patients with hemiparesis due to stroke often have difficulty both advancing the paretic lower extremity and lifting the toe during swing to avoid toe drag (Olney and Richards, 1999). In theory the EBRGT should help the user advance their affected leg while also guiding the toe during swing to simulate level surface walking.
The purpose of this study was to validate the newly developed elliptically-based robotic gait trainer (EBRGT) by evaluating its ability to mimic walking in a normal healthy, adult population.

Methods

Participants

Twenty healthy adults (12 female, 8 male) participated in the study. They had a mean age of 33.9 ± 12.2 years (range=19-58), a mean height of 66.35 ± 3.6 inches (range=60-74), and a mean mass of 72.3 ± 15.5 kg (range=58.9-115.7). Subjects were recruited as a sample of convenience from the university population and surrounding community. Subjects were free from any musculoskeletal, cardiovascular, or neurological impairment that might have affected their ability to perform the activity. This study was approved by the institutional review board at Virginia Commonwealth University and all subjects gave written informed consent before participating in the study.
Device Design

The footplates of the elliptical trainer were modified to adjust the sagittal plane ankle kinematics during use of the elliptical trainer to a pattern that was intended to better mimic level surface walking. The mounts for both the left and right footplates were modified so that each footplate was attached to the ski (Figure 1) using a single axis bearing. This allowed the footplates to tip up and down in the sagittal plane. The new footplate was mounted several inches off of the ski to allow for adequate range of motion to mimic level surface walking. The modified elliptical trainer or EBRGT is depicted in Figure 2.

The motion of the footplates was controlled using a mechanical push/pull mechanism coupled with a servo motor. The control signal was derived from an optical encoder placed on the flywheel/crank (Figure 1) of the elliptical trainer. One complete turn of the flywheel/crank represented one complete gait cycle. The position of the flywheel was aligned with the events of level surface walking. Thus the position of the footplate was determined by the position of the flywheel. For example, the most anterior position of the right footplate is correlated with initial contact during level surface walking. As noted in the control block diagram in figure 3, each footplate is driven by its own controller however; the controllers receive identical input from a single optical encoder on the flywheel.

The device works as follows: The user applies force to the footplates and the flywheel begins moving, as the flywheel turns, an encoder provides flywheel position
input to the controllers, based on the position of the flywheel; the controllers determine
the new position for the footplates and send out a command for the new footplate
position. Thus no motion will occur without input from the user and the gait velocity is
controlled by the user.

Procedures

Electromyography (EMG) and video data were collected during a single one hour
session in a motion analysis laboratory. Before data collection, subjects walked on the
elliptically based robotic gait trainer (EBRGT) until they felt comfortable with the
activity, typically less than three minutes. After habituation, self-adhesive Ag/Cl
electrodes with a sensor area of 13.2mm² (Blue Sensor M, Ambu, Denmark) were placed
over 4 muscles of the right lower extremity: vastus lateralis (VL), tibialis anterior (TA),
biceps femoris (BF), and lateral gastrocnemius (LG) (Delagi and Perotto, 1980). Before
application of the electrodes, the areas were shaved and cleaned with alcohol. Electrodes
were spaced with a 34mm center-to-center distance. A reference electrode was placed
over the right fibular head. Volitional sub-maximal contractions of each muscle were
elicited to verify suitable electrode placement. Four channels of EMG data were
collected simultaneously at 1000 hz using the Myosystem 1200 (Noraxon, Scottsdale,
Az).

Reflective markers were placed on the subject’s right side with double sided tape.
Markers were placed over the lateral border of the acromion, the greater trochanter,
lateral epicondyle of the femur, fibular head, lateral malleolus, lateral aspect of the calcaneus, and lateral aspect of the head of the fifth metatarsal. Landmark locations were palpated using standard techniques. Once the reflective markers were in place, a 14 second static video trial was captured with the subject standing in a neutral position. A high speed digital camera (Bassler Scout, Bassler Inc., Exton, PA) placed 10 feet away from the EBRGT and perpendicular to the subject’s right side was used to capture video at 120hz. Since the participants were normal healthy adults it was assumed that they had a symmetric gait and thus only the right side of each participant was analyzed.

Each participant walked on both the EBRGT and a treadmill (Cateye EC-T220, Boulder, CO) with the order of presentation randomized. They were asked to walk on each device for a total of about 6 minutes. At the end of 2 minutes, 14 seconds of video data was collected to ensure 10 gait cycles were captured. The participant was asked to stop once the video data was collected and the EMG leads were attached to the electrodes. A resting EMG was then collected with the patient in a sitting position. After the resting EMG was collected, the participant was asked to return to the same device and walk for an additional 3 minutes. At the end of the second bout of 2 minutes, 30 seconds of EMG data was collected. While walking on both the EBRGT and treadmill, the participants were told to walk at comfortable pace that they could maintain for about 6 minutes. Participants wore comfortable shorts and a t-shirt or tank top and their own athletic shoes. No strap was provided to keep the participants feet in contact with the footplates, however the subjects were instructed to keep their entire foot in contact with the footplate.
Kinematic data processing

Reflective markers were tracked and digitized using Max TRAQ 2D software (Innovision Systems Inc., Columbiaville, MI). Data was then exported and processed using custom MATLAB software (The MathWorks, Natick, MA). The kinematic data were first low-pass filtered using a fourth-order-zero phase lag Butterworth filter with a cutoff frequency of 6Hz (Winter, 1990). Orthopedic sagittal plane joint angles for the hip, knee, and ankle were calculated. Sagittal plane thigh angles with respect to a vertical line passing through the hip were also calculated. This was done to examine the action of the thigh body segment without the interaction of the trunk position. The sagittal plane angle of the foot with respect to the horizon (foot) was also calculated since this was the parameter that we were controlling with the modifications to the elliptical trainer.

Ten gait cycles were then extracted and time normalized with 0-100% representing a full gait cycle: 0% of the gait cycle represented initial contact. Initial contact on the treadmill was identified as the most positive x-position of the lateral malleolus marker. Since no ‘initial contact’ actually occurs on the EBRGT, the most positive x-position of the lateral malleolus marker also represented 0% of the gait cycle; this marked the transition where the distal part of the lower extremity changed directions from moving anteriorly, to moving in a posterior direction. The EBRGT gait cycle was further broken up into two phases: one where the foot was moving in an anterior direction (swing) and one where the foot was moving a posterior direction (stance). The ten gait cycles for each activity were averaged. The EBRGT gait cycle was further time
normalized for comparison to treadmill walking. During typical level surface walking, swing phase represents 40% of the gait cycle and stance phase represents 60% of the gait cycle. During EBRGT walking, the ratio of stance to swing is 50:50. In order to compare between treadmill and EBRGT walking, the EBRGT data was re-sampled. The stance portion of the gait cycle was re-sampled with 60 points instead of 50 and the swing phase was re-sampled with 40 points instead of 50. The data from all subjects was averaged for each condition. Cadence, speed, and step length were also calculated from the kinematic data.

**EMG data processing**

EMG data was captured and stored on a PC for post-processing using MotionMonitor V7.0 (Innovative Sports Training, Chicago, IL). To determine beginning of the gait cycle within the EMG data, a footswitch was used during treadmill walking and an optical encoder was used during EBRGT walking. The signal of each was sampled at 1000hz and synchronized with the EMG data. EMG data was high pass filtered with a 20 hz cutoff frequency and low pass filtered with a 200hz cutoff frequency. A root-mean-square (RMS) with a 25ms window was performed. The data was then exported in plain text format for further processing with custom MATLAB software (The Math Works, Natick, MA).

In order to determine if significant activity was present, each EMG signal was tested to see if it contained data that was greater than the baseline value plus 5 times the
standard deviation of baseline. After determining if there was activity present, the EMG
signal for each subject was normalized to the maximum EMG signal value over both
conditions. This was done to normalize the EMG so that it could be easily compared
between subjects. Electrodes were not removed between conditions. The 50-50 gait
cycle of the EBRGT was again time normalized so that the same phases of the gait cycle
could be compared between EBRGT and treadmill (see kinematic data analysis for
details). The time normalized cycles were averaged for 10 gait cycles for each
participant and each condition.

Area under the curve (AUC) for processed EMG was calculated for each of the 7
phases of gait: initial loading (il), mid-stance (mst), terminal stance (tst), pre-swing
(psw), initial swing (isw), mid-swing (msw), and terminal swing (Hidler and Wall, 2005).

Data analysis

All statistical analysis was performed using JMP 8 (SAS Corp., Cary, NC). Joint
angles at critical points (Burnfield et al., 2010) during the gait cycle were compared
between the two conditions using paired t-tests. Simple linear correlations between joint
angles during EBRGT and treadmill walking were computed for each subject. AUC
between conditions for each of the 7 phases of gait were compared using paired t-tests. A
Bonferonni adjustment was used to account for multiple comparisons. Alpha was set at
0.05 for all tests.
RESULTS

Temporal and spatial gait parameters

The stride length while walking on the EBRGT was 35.2±1.4cm, which was much less than the 59.1±7.1cm mean stride length on the treadmill (Table 1). The stride length of the EBRGT is fixed, but the footplates allow for some shifting of the foot forward and backward, which may explain some variability. Cadence while walking on the EBRGT was also slightly slower when compared to the treadmill. The shorter stride length and slower cadence resulted in the walking speed on the EBRGT to be almost half that of the treadmill.

Kinematics

Thigh, hip, knee, ankle and foot angles all revealed similar movement patterns between EBRGT and Treadmill walking (Figure 4). However, there were differences in the magnitude of the angles for all joints.

Thigh (thigh relative to a vertical axis) angles during both conditions were similar in phase and magnitude until the last 25% of the gait cycle (Figure 4A). During that
portion of swing phase, there was a much larger peak in thigh flexion angles during EBRGT walking than during treadmill walking. While hip (trunk relative to thigh) angles showed a similar phasing during both EBRGT and treadmill walking, the magnitude of hip flexion was consistently greater during EBRGT for the entire gait cycle (Figure 4B).

Knee angles were similar during each activity with a slightly later peak during EBRGT (Figure 4C). The magnitude of knee flexion was also greater during EBRGT than treadmill walking from midswing through initial loading (75%-12% of the gait cycle).

Ankle angles showed similar plantar/dorsiflexion reversals for both conditions with similar phasing but large differences in magnitude (Figure 4D). Similar to the knee, the largest magnitude differences in dorsiflexion occurred between midswing and initial contact (75-100% of gait cycle) and then again during initial loading (0-12% of gait cycle). A short period of plantarflexion that occurs within initial loading during treadmill walking is absent during EBRGT walking.

Foot angles showed very similar patterns for both activities in terms of magnitude and phasing. The foot angles were negative during midstance while EBRGT walking, where during treadmill walking the foot angle was zero.

There were statistically significant differences between joint angles at critical points (Table 2) when comparing treadmill walking and EBRGT for the hip, thigh, knee, and ankle. The foot angles relative to horizon were not significantly different at 3 of the
4 points tested. While there was a significant difference in the minimum foot angle value during loading response, the actual difference between the angles was less than 5 degrees.

Simple linear correlations (Table 3) revealed excellent correlation coefficients for hip, thigh, and foot angles between the two walking conditions. Correlation of knee joint angles between treadmill and EBRGT walking revealed a moderate correlation coefficient. Ankle joint angles between the two conditions were very poorly correlated.

*Electromyography*

Qualitative analysis of the EMG profiles (Figure 5) of walking on the treadmill and walking on the EBRGT revealed both similarities and differences in muscle activity between the activities. The vastus lateralis muscle (VL) showed peak activity at the beginning and end of the gait cycle during both walking conditions (Figure 5A); however, there was a much higher level of activation for the entire gait cycle during EBRGT walking. The tibialis anterior muscle (TA) EMG profile (Figure 5B) was again very similar in phasing of peak activity when comparing the treadmill and EBRGT walking. The peak that occurs during initial swing was similar in timing and magnitude, while the peak that occurred at initial contact was less than half the magnitude during EBRGT walking as compared to treadmill.

Biceps femoris (BF) muscle EMG profile was not similar between the two activities. There was a moderate level of BF activity during EBRGT walking over the whole gait cycle with a relative peak at about 40% of the gait cycle and again at 90% of
the gait cycle. During treadmill walking there was a peak also around 90% of the gait cycle, but the magnitude of activation was almost twice that of the peak during EBRGT walking.

During treadmill walking there is a burst of activity in lateral gastrocnemius muscle (LG) between 30 and 60% of the gait cycle. During the rest of the cycle there is little to no activity in the LG. During EBRGT walking there is a slight rise in activation from 10-30% of the gait cycle, but the rise occurs earlier than the peak during treadmill walking and is only about 1/3 of the magnitude.

AUC analysis revealed that for 3 of the 4 muscles observed there were more non-significant differences than significant over the 7 phases of the gait cycle (Figure 6, Table 4). The VL had the most gait phases with significant differences between EBRGT walking and treadmill walking. Only 1 of the 7 phases of gait, initial loading (il) was not statistically different (Table 4). During all of the other gait phases there was a much greater AUC for EBRGT than treadmill walking (Figure 6A). For example, during terminal swing, there was nearly a 6 fold difference between values.

For the TA, AUC for only 2 of the 7 phases, il and tsw, were significantly different when comparing treadmill and EBRGT walking (Table 4). The AUC for these two periods was at least double the value during treadmill walking as compared to EBRGT (Figure 6B).

AUC for three gait phases was significantly different between conditions for the BF; tst, psw, and tsw (Table 4). During tst and psw, there was at least 3x greater AUC
during EBRGT than Treadmill walking (Figure 6C). While during tsw, there was significantly greater muscle activity during treadmill walking than EBRGT.

The LG only had one phase of gait, tst, where there was a significant difference between EBRGT and treadmill walking (Table 4). During tst, the AUC was much greater during treadmill walking than EBRGT walking (Figure 6D).

Discussion

Analysis of sagittal plane joint kinematics revealed some similarities as well as differences between treadmill and EBRGT walking. The hip and thigh had very similar movement profiles between the two walking conditions (r=0.95 and r=0.96 respectively), however there was statistically more flexion for both measures during EBRGT walking than treadmill walking. The hip and thigh angles were both calculated in order to examine the effects of trunk positioning on the hip angle; the hip angle minus the thigh angle yields the trunk position. When subjects were walking on the EBRGT, they appeared to be leaning forward in order grasp the handles of the gait trainer. During the development of the EBRGT, the footplates were elevated to allow for range of motion of the footplates. This caused the handles of the gait trainer to be at a lower elevation relative to the user. Due to the position of the handles, the subjects were leaning forward with their trunk to reach the handles, thus increasing hip joint flexion. While thigh angles were still consistently more flexed during EBRGT than during treadmill walking, there was only a 7.2 degree difference at peak extensions for the thigh versus a 13.6 degree
difference between conditions at the hip. This means that when abstracted from the movement of the trunk, the thigh angles more closely resemble treadmill walking. It may be helpful to redesign the handles in the future to reduce the forward tilt of the trunk so that EBRGT walking better mimics hip movements during level surface walking.

While the thigh angles during EBRGT walking closely mimic level surface walking during most of the gait cycle, there is still about a 20 degree difference in peak hip flexion during midswing. As stated by Burnfield et al. (2010), there is a larger vertical translation of the foot during the EBRGT gait cycle, than during level surface walking. There are two ways to compensate for this movement: either the center of mass can translate further in the vertical direction, or the joints of the lower extremity can flex more to shorten the distance between the hip and the foot. While the movement of the center of mass was not measured in this study, it is suspected that movement of the center of mass, as with most activities, would be minimized (Lu et al., 2007). It is believed that the reason for the increased thigh flexion during midswing is a result of the vertical translation of the foot, which also peaks during midswing when the crank/flywheel (Figure 1) is in the highest position. There is a pattern of increased flexion (dorsiflexion at the ankle) at all of the joints except the foot angle when comparing EBRGT walking to treadmill walking. This is again attributed to the vertical translation of the foot.

It was expected that there would be perfect correlation for foot angles when comparing the two conditions, since the movement of the footplate was designed to mimic level surface walking. The correlation was good (r=0.95), but there were some differences in foot angles between the two conditions. The footplates of the EBRGT were constructed of a flat piece of aluminum with a rough surface. For this study, we did
not use any sort of strap to keep the subjects foot in contact with the footplate, we just asked that the participants keep their heels down and in contact with the footplates. It is possible that the differences in foot angles were due to the participants not staying in contact with the footplate. In future studies, straps or some other system may be implemented to help control the movement of the foot.

Burnfield et al. (2010) compared joint kinematics and muscle activity between four different elliptical trainers and level surface walking (Table 2). Our kinematic results are similar to the results of that study with the exception of ankle kinematics during elliptical use. This was expected since we made modifications to how the foot articulates during use of our modified elliptical trainer. This also provides evidence that our modifications only affected the ankle and did not cause changes at the other joints in terms of kinematics.

There were both differences and similarities in muscle activity between the EBRGT and treadmill walking. The muscle with the greatest differences was the VL. This may be explained by the increased knee flexion measured during EBRGT walking (Figure 4C). A flexed knee during weight bearing causes the moment arm of the body weight to increase as the knee flexes. This means that the knee extensors would have to exert a larger force to keep the knee from collapsing. A study by Lu et al. (2007) also found that during elliptical training the knee extensor moment required while walking on an elliptical trainer was 9x greater during swing and 3x greater during stance than walking. This supports our findings since the VL is part of the group of muscles that extends the knee. A different elliptical trainer without modifications was used in the
study by Lu et al. (2007); however the joint kinematics they reported were similar to this study.

The difference in TA muscle activity during tsw and ic may be explained by the fact that the EBRGT dorsiflexes the foot for the user and thus assists the action of the TA, requiring less muscles activity. The BF muscle activity showed differences during tst, psw, and tsw. The differences during terminal stance may be explained by the increased hip flexion during use of the EBRGT. During treadmill walking, the trunk is nearly perpendicular so that little muscle activity is required to maintain posture. During EBRGT walking it was noted that the participants were always leaning forward. The BF may have been firing at a higher level during tst and psw during EBRGT walking in order to maintain posture since the line of gravity for the trunk would no longer be passing through the hip. The decreased level of BF activity at tsw during EBRGT walking may be explained by the foot always remaining in contact with the footplate. During treadmill walking the BF would assist in decelerating the lower leg and foot at the end of swing in preparation for initial contact. Since there is no true swing during EBRGT walking, there may be no need to decelerate the lower leg and foot. The LG only exhibited a statistical difference during one phase of gait when comparing EBRGT and treadmill walking: tst. During treadmill walking, the LG is active to assist in push off, where the lower extremity and body are propelled forward. Lu et al. (2007) found that the required plantarflexor moment was much smaller during EBRGT walking than level surface walking, which supports our findings since the LG is a plantarflexor. However, the peak posteriorly directed reaction forces in the Lu et al. (2007) study were not different between the two activities suggesting that this force may have come from a different
group of muscles, such as the hip extensors which include the BF. The footplate trajectory of the EBRGT during this phase of gait moves the users foot into plantarflexion and thus may reduce the action of the LG muscle.

It has been shown that muscle activity changes with gait speed (Murray et al., 1984, den Otter et al., 2004). In this study, the participants were instructed to walk at a comfortable pace on both the elliptical trainer and the treadmill. This resulted in the participants walking with a different cadence as well as a different stride length and thus at a different gait speed. Participants walked slower on the EBRGT than on the treadmill. In general muscle activity decreases with decreasing gait speed (Murray et al., 1984, den Otter et al., 2004). If gait speed were the reason for muscle activity differences there would have been a decrease in gain for all the muscles during EBRGT walking relative to treadmill walking and this was not the case in this study.

Limitations of this study include the use of two dimensional kinematics and EMG recorded from a small number of muscles of the lower extremity. While walking is primarily a sagittal plane activity, there are movements that occur in all three dimensions. Only four muscles were monitored in the lower extremities, it may be useful to monitor more muscles to provide a better picture of the activity of walking on the EBRGT as it relates to treadmill walking.
CONCLUSION

The EBRGT was able to mimic some aspects of treadmill walking while there were some notable differences. There was a general increase in flexion across all lower extremity joints for most of the gait cycle during EBRGT walking relative to treadmill walking. While this was a notable difference, it does not necessarily mean that the EBRGT is not a useful tool for retraining gait. There were also notable differences in muscle activity. Again, while the muscle activity of the EBRGT did not perfectly simulate level surface walking all four muscle groups of the lower extremity did fire in a cyclic, coordinated pattern which is often lacking in patients recovering from hemiparesis due to stroke (Knuttson and Richards, 1979). Task specific training for recovery of gait post stroke is focused on practicing the task at which you want to improve, however, for gait, no one has determined the degree to which the activity needs to precisely simulate normal gait. Studies are ongoing to determine the feasibility and efficacy of using the EBRGT to retrain gait in patients with hemiparesis due to stroke.
Table 1. Temporal and spatial gait parameters during EBRGT and treadmill walking.

<table>
<thead>
<tr>
<th>Measure</th>
<th>EBRGT</th>
<th>Tread</th>
</tr>
</thead>
<tbody>
<tr>
<td>Step Length (cm)</td>
<td>35.2 (1.4)</td>
<td>59.1 (7.1)</td>
</tr>
<tr>
<td>Cadence (strides/sec)</td>
<td>0.84 (0.14)</td>
<td>1.0 (0.08)</td>
</tr>
<tr>
<td>Speed (cm/sec)</td>
<td>59.5 (10.4)</td>
<td>119.4 (21.0)</td>
</tr>
</tbody>
</table>
### Table 2. Summary of joint angles at critical points and significance results from paired t-test (p=0.05/18=0.0028, n=19). **NS** = instances where there was no statistical difference between the two conditions. Data from Burnfield et al. (2010) also included in table for comparison. These authors compared level surface walking to walking on four different elliptical trainers; SportsArt, Life Fitness, Octane, and TRUE.

<table>
<thead>
<tr>
<th>Joint</th>
<th>Phase</th>
<th>Tread</th>
<th>EBRGT</th>
<th>Sig.</th>
<th>(Burnfield et al. 2010)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>Walking</td>
<td>SportsArt</td>
<td>Life Fitness</td>
<td>Octane</td>
</tr>
<tr>
<td>Hip</td>
<td>IC</td>
<td>27.7(4.6)</td>
<td>47.1(4.7)</td>
<td>&lt;0.0001</td>
<td>31.5(7.2)</td>
</tr>
<tr>
<td></td>
<td>TSt peak ext</td>
<td>0.82(3.4)</td>
<td>14.4(6.2)</td>
<td>&lt;0.0001</td>
<td>-7.3(7.6)</td>
</tr>
<tr>
<td>MSw peak flex</td>
<td>33.0(3.8)</td>
<td>59.8(5.0)</td>
<td>&lt;0.0001</td>
<td>34.4(4.9)</td>
<td>54.1(6.7)</td>
</tr>
<tr>
<td>Thigh</td>
<td>IC</td>
<td>22.5(4.2)</td>
<td>34.5(3.7)</td>
<td>&lt;0.0001</td>
<td>23.3(4.2)</td>
</tr>
<tr>
<td></td>
<td>TSt peak ext</td>
<td>-10.7(3.3)</td>
<td>-3.5(4.0)</td>
<td>&lt;0.0001</td>
<td>-14.7(4.4)</td>
</tr>
<tr>
<td>MSw peak flex</td>
<td>26.7(3.6)</td>
<td>47.5(4.1)</td>
<td>&lt;0.0001</td>
<td>26.3(6.4)</td>
<td>40.7(4.0)</td>
</tr>
<tr>
<td>Knee</td>
<td>IC</td>
<td>4.9(3.2)</td>
<td>38.6(7.0)</td>
<td>&lt;0.0001</td>
<td>3.7(5.6)</td>
</tr>
<tr>
<td></td>
<td>LR final position</td>
<td>15.1(9.7)</td>
<td>23.8(8.6)</td>
<td>0.0004</td>
<td>19.3(6.8)</td>
</tr>
<tr>
<td></td>
<td>TSt peak ext</td>
<td>6.0(4.4)</td>
<td>16.2(3.3)</td>
<td>&lt;0.0001</td>
<td>6.2(5.6)</td>
</tr>
<tr>
<td>MSw peak flex</td>
<td>68.0(7.0)</td>
<td>44.9(9.0)</td>
<td>&lt;0.0001</td>
<td>66.8(7.1)</td>
<td>72.4(5.3)</td>
</tr>
<tr>
<td>Ankle</td>
<td>IC</td>
<td>1.1(0.57)</td>
<td>20.9(4.5)</td>
<td>&lt;0.0001</td>
<td>3.0(3.7)</td>
</tr>
<tr>
<td></td>
<td>LR peak PF</td>
<td>-4.9(2.0)</td>
<td>8.2(5.2)</td>
<td>&lt;0.0001</td>
<td>-2.9(3.1)</td>
</tr>
<tr>
<td></td>
<td>TSt peak DF</td>
<td>16.4(3.3)</td>
<td>11.4(4.5)</td>
<td>0.0004</td>
<td>14.8(3.2)</td>
</tr>
<tr>
<td></td>
<td>MSw final position</td>
<td>2.3(3.0)</td>
<td>23.2(46)</td>
<td>&lt;0.0001</td>
<td>3.4(2.3)</td>
</tr>
<tr>
<td>Foot</td>
<td>IC</td>
<td>15.7(4.5)</td>
<td>14.0(2.8)</td>
<td>0.1201</td>
<td>14.0(2.8)</td>
</tr>
<tr>
<td></td>
<td>LR min angle</td>
<td>0.75(2.0)</td>
<td>5.5(4.8)</td>
<td>0.0002</td>
<td>0.75(2.0)</td>
</tr>
<tr>
<td></td>
<td>TSt max angle</td>
<td>-0.9(1.8)</td>
<td>-3.7(4.7)</td>
<td>0.0168</td>
<td>-0.9(1.8)</td>
</tr>
<tr>
<td></td>
<td>MSw final position</td>
<td>-7.4(4.4)</td>
<td>-8.7(3.1)</td>
<td>0.2641</td>
<td>-7.4(4.4)</td>
</tr>
</tbody>
</table>
Table 3. Results of simple linear correlations between joint angles during EBRGT and Treadmill walking. Correlations were performed for each subject and the mean and standard deviation of all subject data is presented. An r value of 1 represents a perfect linear correlation and r value of zero represents no correlation at all.

<table>
<thead>
<tr>
<th>Joint</th>
<th>$R^2$, mean (SD)</th>
<th>r</th>
</tr>
</thead>
<tbody>
<tr>
<td>hip</td>
<td>0.90(0.04)</td>
<td>0.95</td>
</tr>
<tr>
<td>Thigh</td>
<td>0.93(0.03)</td>
<td>0.96</td>
</tr>
<tr>
<td>knee</td>
<td>0.41(0.16)</td>
<td>0.64</td>
</tr>
<tr>
<td>ankle</td>
<td>0.06(0.07)</td>
<td>0.24</td>
</tr>
<tr>
<td>foot</td>
<td>0.90(0.03)</td>
<td>0.95</td>
</tr>
</tbody>
</table>
Table 4. Summary of significant differences between EBRGT and Tread AUC values for each phase of gait (p≤0.5/28=0.0018, VL n=17, TA n=18, BF n=19, LG n=17). ‘*’ denotes a significant difference.

<table>
<thead>
<tr>
<th>Muscle</th>
<th>il</th>
<th>mst</th>
<th>tst</th>
<th>psw</th>
<th>isw</th>
<th>msw</th>
<th>tsw</th>
</tr>
</thead>
<tbody>
<tr>
<td>vl</td>
<td>-</td>
<td>*</td>
<td>*</td>
<td>*</td>
<td>*</td>
<td>*</td>
<td>*</td>
</tr>
<tr>
<td>ta</td>
<td>*</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>*</td>
</tr>
<tr>
<td>bf</td>
<td>-</td>
<td>-</td>
<td>*</td>
<td>*</td>
<td>-</td>
<td>-</td>
<td>*</td>
</tr>
<tr>
<td>lg</td>
<td>-</td>
<td>-</td>
<td>*</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
</tbody>
</table>
Figure 1. The commercially available elliptical trainer before modifications.
Figure 2. EBRGT (elliptical trainer after modifications) with a normal healthy subject.
Figure 3. Control block diagram for modified elliptical gait trainer.
Figure 4. Sagittal plane joint angles (degrees) for the thigh (A), hip (B), knee (C), ankle (D), and foot (E) for one complete gait cycle, starting with initial contact. For the thigh, hip and knee; flexion is represented by positive angles and extension is represented by negative angles. For the ankle, dorsiflexion is represented by positive angles and plantarflexion is represented by negative angles. The foot angles are calculated relative to the horizontal; positive angles represent when the toe is above the horizon and negative angles represent when the toe is below the horizon. Error bars represent one standard deviation from the mean and the vertical black line demarcates the transition from stance to swing.
Figure 5. Ensemble averaged EMG profiles during treadmill and EBRGT walking for the vastus lateralis muscle (A), the tibialis anterior muscle (B), the biceps femoris muscle (C), and the lateral gastrocnemius muscle (D). The plots represent one full gait cycle starting with initial contact. Shaded area represents 1 standard deviation from the mean. Vertical black line demarcates transition from stance to swing phase.
Figure 6. Area under the curve (AUC) plots across the 7 phases of gait for the VL (A), TA (B), BF (C), and LG (D) for both walking conditions. Instances of significant difference demarcated with a ‘*’ (p=0.05/28=0.0018, n=17, n=18, n=19, n=17 respectively) Values shown are an average of all subjects. Error bars represent 1 standard deviation.
Works Cited


Appendix A

Custom MATLAB software: Kinematics

% This program calculates joint angles for the hip, knee, ankle, and foot
% with respect to horizon. Then these are broken up into gait cycles and
% averaged. In the ebrgt trials there are 18 columns, in the treadmill
% trials there are 14 columns. These represent the x,y coordinates of 9 and
% 7 points respectively. The points represent: shoulder, greater trochanter, lateral epicondyle, fibular head, lateral maleolus, calcaneous, head of the fifth metatarsal, back of footplate, front of footplate. The last two don't exist in the treadmill trials.

Subjects performed at self selected speed.

clear all
filename = input('Enter name of first file with extension', 's');
file = xlsread(filename);
len=length(file);

filenameb = input('Enter name of baseline file with extension', 's');
fileb = xlsread(filenameb);
lenb=length(fileb);

syb = fileb(11:lenb,1);
sxb = -(fileb(11:lenb,2));
hyb = fileb(11:lenb,3);
hxb = -(fileb(11:lenb,4));
lyb = fileb(11:lenb,5);
lxb = -(fileb(11:lenb,6));
fyb = fileb(11:lenb,7);
fxb = -(fileb(11:lenb,8));
ayb = fileb(11:lenb,9);
axb = -(fileb(11:lenb,10));
cyb = fileb(11:lenb,11);
cxb = -(fileb(11:lenb,12));
myb = fileb(11:lenb,13);
mxb = -(fileb(11:lenb,14));
%byb = fileb(11:lenb,15);%
%bxb = -(fileb(11:lenb,16));
%fryb = fileb(11:lenb,17);
%frxb = -(fileb(11:lenb,18));

sy = file(11:len,1);
sx = -(file(11:len,2));
hy = file(11:len,3);
hx = -(file(11:len,4));
ly = file(11:len,5);
lx = -(file(11:len,6));
fy = file(11:len,7);
fx = -(file(11:len,8));
ay = file(11:len,9);
ax = -(file(11:len,10));
cy = file(11:len,11);
cx = -(file(11:len,12));
my = file(11:len,13);
mx = -(file(11:len,14));
%by = file(11:len,15);
%bx = -(file(11:len,16));
%fry = file(11:len,17);
%frx = -(file(11:len,18));

[B,A] = butter (4,0.4);
sy = filtfilt (B,A, sy);
sx = filtfilt (B,A, sx);
hy = filtfilt (B,A, hy);
hx = filtfilt (B,A, hx);
ly = filtfilt (B,A, ly);
lx = filtfilt (B,A, lx);
fy = filtfilt (B,A, fy);
fx = filtfilt (B,A, fx);
ay = filtfilt (B,A, ay);
ax = filtfilt (B,A, ax);
cy = filtfilt (B,A, cy);
cx = filtfilt (B,A, cx);
my = filtfilt (B,A, my);
mx = filtfilt (B,A, mx);
%by = filtfilt (B,A, by);
%bx = filtfilt (B,A, bx);
%fry = filtfilt (B,A, fry);
%frx = filtfilt (B,A, frx);

%The following code calculates hip angle as the angle between the thigh segment and the horizon. When the thigh is perpendicular to horizon, the angle is zero.
clear hip
hip = atand((lx-hx)./(hy-ly));

%the following code calculates the hip angles as the angle between the trunk and thigh. The hip angle is zero when the trunk and thigh are parallel

clear hip_o
hip_o =(atand((sx-hx)./(sy-hy))+ atand((lx-hx)./(hy-ly)));
The following code calculates the knee angle as the angle between the thigh segment and lower leg segment.

```matlab
kn = 180 - ((90 - hip) + (90 + atand((ax-fx)./(fy-ay))));
```

The following code calculates the ankle angle as the angle between lower leg segment and the foot segment. The ankle angle is zero when the lower leg and foot are perpendicular.

```matlab
ak = (180/pi).*atan2((my-cy),(mx-cx)) + (180/pi).*atan2((fx-ax),(fy-ay));
```

The following calculates the foot wrt horizon angle.

```matlab
fh = (180/pi).*atan2((my-cy),(mx-cx));
```

The following calculates the footplate wrt horizon angle.

```matlab
fp = atand((fry-by)./(frx-bx));
```

Baseline angle calculations follow.

The following code calculates hip angle as the angle between the thigh segment and the horizon. When the thigh is perpendicular to horizon, the angle is zero.

```matlab
hipb = atand((lxb-hxb)./(hyb-lyb));
```

The following code calculates the hip angles as the angle between the trunk and thigh. The hip angle is zero when the trunk and thigh are parallel.

```matlab
hip_ob = (atand((sxb-hxb)./(syb-hyb)) + atand((lxb-hxb)./(hyb-lyb)));
```

The following code calculates the knee angle as the angle between the thigh segment and lower leg segment.

```matlab
knb = 180 - ((90 - hipb) + (90 + atand((AXB-fxb)./(fyb-ayb))));
```

The following code calculates the ankle angle as the angle between lower leg segment and the foot segment. The ankle angle is zero when the lower leg and foot are perpendicular.

```matlab
akb = atand((myb-cyb)./(mxb-cxb)) + atand((fxb-axb)./(fyb-ayb));
```

The following calculates the foot wrt horizon angle.

```matlab
fhb = atand((myb-cyb)./(mxb-cxb));
```
hipbm = hipb(length(hipb));
hip_obm = hip_ob(length(hip_ob));
knbm = knb(length(knb));
akbm = akb(length(akb));
fhbm = fhb(length(fhb));

 hippoc= hipb(length(hipb));
hip=hipb-hipnorm;
thigh=thighb-knee_norm;
ank=akb-ankle_norm;

%Normalize joint angles by subtracting baseline.
%Input normalization values from tread for knee and ankle
hip_norm = input('Enter normalization value for hip');
hip_norm = input('Enter normalization value for thigh');
knee_norm = input('Enter normalization value for the knee');
ankle_norm = input('Enter normalization value for the ankle');
fh_norm = input('Enter normalization value for the foot');

hip_ooff = hip_obm-hip_onorm;
hip_off = hipbm-hip_norm;
kn_off = knbm-knee_norm;
ak_off = akbm-ankle_norm;
fh_off = fhbm-fh_norm;

kip = hip -hip_off;
hip_on = hip_o -hip_ooff;
k = kn-kn_off;
ank = ak-ak_off;
fh = fh-fh_off;

%The following breaks each joint angle up into gait cycles beginning
with initial contact and creates a new column for each new gait cycle.

%The following code finds the points where ax is greatest, which
represents initial contact
clear high;
k=1;
m=1;
for i=16:1:(length(ax)-15);
    if ax(i)> ax(i-2) && ax(i)> ax(i-5) && ax(i)>ax(i+2) &&
        ax(i)>ax(i+5) && m==1;
        high(k)= i;
        k=k+1;
        m=2;
    elseif m==2 && i>(high(k-1)+30)
        m=1;
    end
end

%The following code determines the average speed in strides/second
for b=2:length(high);
cycle(b)=high(b)-high(b-1);
ave_frames= mean(cycle);
ave_speed = 1/(ave_frames/30);
max_ax = max(ax);
min_ax = min(ax);

% step length calculation in arbitrary units
sl = max_ax - min_ax;

% calibration using leg length measure
ll = fyb - ayb;
ll = mean(ll);
ll_cm = input('enter lower leg length in cm');
calib = ll_cm / ll;

% step length in cm
sl_cm = sl * calib;

L = {'ave speed (strides/sec)', ave_speed};

clear yy
clear xx
clear x1
clear ln
clear j
clear x2
clear hip_cy
clear x3
clear x4
clear yy_m
clear yy2
clear yy3
clear a
clear b

for j = 1:1:(length(high) - 1);
    ln = (high(j + 1) - high(j));
    x1 = 1:1:ln;
    xx = spline(x1, hipn(high(j):(high(j + 1) - 1)));
    a = floor(ln / 2);
    b = ceil(ln / 2);
    x2 = ln / 100:ln / 100:ln;
    x3 = a / 60:a / 60:a;
    x4 = (b / 40) + (a); (b) / 40: (b) + (a);
    yy = ppval(xx, x2);
    % resample so that ebrgt can be compared with treadmill (make 50/50
    % gait cycle match up with 60/40 of treadmill
    yy2 = ppval(xx, x3);
    yy3 = ppval(xx, x4);
    yy_m(1:60) = yy2;
    yy_m(61:100) = yy3;
    hip_m(:, j) = yy_m';
    yy = yy';
    hip_cy(:, j) = yy;
    hip_m1 = hip_m;
end
clear yy
clear xx
clear x1
clear ln
clear j
clear x2
clear hipo_cy
clear x3
clear x4
clear yy_m
clear yy2
clear yy3
clear a
clear b
for j=1:1:(length(high)-1);
    ln=(high(j+1)-high(j));
    x1=1:1:ln;
    xx=spline(x1,hip_on(high(j):(high(j+1)-1)));
    a=floor(ln/2);
    b=ceil(ln/2);
    x2=ln/100:ln/100:ln;
    x3=a/60:a/60:a;
    x4=(b/40)+(a):(b)/40:(b)+(a);
    yy=ppval(xx,x2);
    %resample so that ebrgt can be compared with treadmill (make 50/50 gait cycle match up with 60/40 of treadmill
    yy2=ppval(xx,x3);
    yy3=ppval(xx,x4);
    yy_m(1:60)=yy2;
    yy_m(61:100)=yy3;
    yy=yy';
    hipo_m(:,j)=yy_m';
    hipo_cy(:,j)=yy;
    hipo_m1=hipo_m;
end

clear yy
clear xx
clear x1
clear ln
clear j
clear x2
clear knn_cy
clear x3
clear x4
clear yy_m
clear yy2
clear yy3
clear a
clear b
for j=1:1:(length(high)-1);
    ln=(high(j+1)-high(j));
    x1=1:1:ln;
    xx=spline(x1,knn(high(j):(high(j+1)-1)));
    a=floor(ln/2);
    b=ceil(ln/2);
    x2=ln/100:ln/100:ln;
    x3=a/60:a/60:a;
    x4=(b/40)+(a):(b)/40:(b)+(a);
    yy=ppval(xx,x2);
    %resample so that ebrgt can be compared with treadmill (make 50/50 gait cycle match up with 60/40 of treadmill
    yy2=ppval(xx,x3);
    yy3=ppval(xx,x4);
    yy_m(1:60)=yy2;
    yy_m(61:100)=yy3;
    yy=yy';
    hipo_m(:,j)=yy_m';
    hipo_cy(:,j)=yy;
    hipo_m1=hipo_m;
end
yy3=ppval(xx,x4);
yy_m(1:60)=yy2;
yy_m(61:100)=yy3;
knn_m(:,j)=yy_m';
yy=yy';
knn_cy(:,j)=yy;
knn_m1=knn_m;
end

clear yy
clear xx
clear x1
clear ln
clear j
clear x2
clear ak_cy
clear x3
clear x4
clear yy_m
clear yy2
clear yy3
clear a
clear b
for j=1:1:(length(high)-1);
    ln=(high(j+1)-high(j));
    x1=1:1:ln;
    xx=spline(x1,akn(high(j):(high(j+1)-1)));
    a=floor(ln/2);
    b=ceil(ln/2);
    x2=ln/100:ln/100:ln;
    x3=a/60:a/60:a;
    x4=(b/40)+(a):(b)/40:(b)+(a);
    yy=ppval(xx,x2);
    %resample so that ebrgt can be compared with treadmill (make 50/50
gait cycle match up with 60/40 of treadmill
    yy2=ppval(xx,x3);
    yy3=ppval(xx,x4);
    yy_m(1:60)=yy2;
    yy_m(61:100)=yy3;
    ak_m(:,j)=yy_m';
    yy=yy';
    ak_cy(:,j)=yy;
    ak_m1=ak_m;
end

clear yy
clear xx
clear x1
clear ln
clear j
clear x2
clear fh_cy
clear x3
clear x4
clear yy_m
clear yy2
clear yy3
clear a
clear b
for j=1:1:(length(high)-1);
  ln=(high(j+1)-high(j));
  x1=1:1:ln;
  xx=spline(x1,fhn(high(j):(high(j+1)-1)));a=floor(ln/2);
b=ceil(ln/2);
  x2=ln/100:ln/100:ln;
  x3=a/60:a/60:a;
  x4=(b/40)+(a):(b)/40:(b)+(a);
  yy=ppval(xx,x2);
  %resample so that ebrgt can be compared with treadmill (make 50/50
gait cycle match up with 60/40 of treadmill
  yy2=ppval(xx,x3);
  yy3=ppval(xx,x4);
  yy_m(1:60)=yy2;
  yy_m(61:100)=yy3;
  fhm(:,j)=yy_m';
  yy=yy';
  fh_cy(:,j)=yy;
  fh_m1=fhm;
end
hip_cy = mean(hip_cy,2);
hip_m = mean(hip_m,2);
hipo_cy=mean(hipo_cy,2);
hipo_m=mean(hipo_m,2);
knn_cy = mean(knn_cy,2);
knn_m=mean(knn_m,2);
ak_cy=mean(ak_cy,2);
ak_m=mean(ak_m,2);
fh_cy=mean(fh_cy,2);
fh_m=mean(fh_m,2);

%The following code finds the max and min for each angle
[hip_max,hip_max_index] = max(hip_m);
[hip_min, hip_min_index] = min(hip_m);
[hipo_max, hipo_max_index] = max(hipo_m);
[hipo_min, hipo_min_index] = min(hipo_m);
[knn_max, knn_max_index] = max(knn_m);
[knn_min, knn_min_index] = min(knn_m);
[ak_max, ak_max_index]=max(ak_m);
[ak_min, ak_min_index]=min(ak_m);
[fh_max, fh_max_index]=max(fh_m);
[fh_min, fh_min_index] = min(fh_m);
hip_max = mean(hip_max);
hip_max_index = mean(hip_max_index);
hip_min = mean(hip_min);
hip_min_index = mean(hip_min_index);
hipo_max = mean (hipo_max);
hipo_max_index = mean(hipo_max_index);
hipo_min = mean(hipo_min);
hipo_min_index = mean (hipo_min_index);
knn_max = mean(knn_max);
knn_max_index = mean(knn_max_index);
knn_min = mean(knn_min);
knn_min_index = mean(knn_min_index);
ak_max = mean(ak_max);
ak_max_index = mean(ak_max_index);
ak_min = mean(ak_min);
ak_min_index = mean(ak_min_index);
fh_max = mean(fh_max);
fh_max_index = mean(fh_max_index);
fh_min = mean(fh_min);
fh_min_index = mean(fh_min_index);

max_min = {'joint' 'maximum' 'index' 'minimum' 'index'; 'hip' hip_max
hip_max_index hip_min hip_min_index; 'hipo' hipo_max hipo_max_index
hipo_min hipo_min_index;
'knee' knn_max knn_max_index knn_min knn_min_index; 'ankle' ak_max
ak_max_index ak_min ak_min_index; 'foot' fh_max fh_max_index fh_min
fh_min_index};

%The following code finds critical even joint angles recorded during
%elliptical training

%hip - initial contact (ic), peak ext during terminal stance (tst),
peak
%flexion during midswing (msw).
hipo_ic = hipo_m(1);
hipo_tst = min(hipo_m(31:50));
hipo_msw = max(hipo_m(76:87));

hipo_crit = {'ic' 'tst_min' 'msw_max'; hipo_ic hipo_tst hipo_msw};

%thigh (or hip relative to vertical axis)
hip_ic = hip_m(1);
hip_tst = min(hip_m(31:50));
hip_msw = max(hip_m(76:87));

hip_crit = {'ic' 'tst_min' 'msw_max'; hip_ic hip_tst hip_msw};

%knee - initial contact (ic), loading response final position (LR),
peak
%ext during terminal stance (tst), peak flex during initial swing (isw)

knn_ic = knn_m(1);
knn_lr = knn_m(12);
knn_tst = min(knn_m(31:50));
knn_isw = max(knn_m(51:67));

knn_crit = {'ic' 'lr_final' 'tst_min' 'isw_max'; knn_ic knn_lr knn_tst
knn_isw};
%ankle - initial contact (ic), loading response peak plantar flexion (lr),
%peak dorsiflexion during terminal stance (tst), final position of midswing
%(msw)

ak_ic = ak_m(1);
ak_lr = min(ak_m(1:12));
ak_tst = max(ak_m(31:50));
ak_msw = ak_m(87);

ak_crit = {'ic' 'lr_min' 'tst_max' 'msw_final'; ak_ic ak_lr ak_tst
ak_msw};

%foot wrt horizon - same angles as ankle
fh_ic = fh_m(1);
fh_lr = min(fh_m(1:12));
fh_tst = max(fh_m(31:50));
fh_msw = fh_m(87);

foot_crit = {'ic' 'lr_min' 'tst_max' 'msw_final'; fh_ic fh_lr fh_tst
fh_msw};

filename3 = input('Enter name of output file for angles at critical
events','s');
xlswrite (filename3, max_min, 01)
xlswrite (filename3, hip_crit, 02)
xlswrite (filename3, hipo_crit, 03)
xlswrite (filename3, knn_crit, 04)
xlswrite (filename3, ak_crit, 05)
xlswrite (filename3, foot_crit, 06)
xlswrite (filename3, L, 07)
Appendix B

Custom MATLAB software: Electromyography

%This program reads in a four channel EMG file with a marker. There are 6 columns of data. Column 1 is frame #, 2 is index, 3 is footswitch, 4 is vl, 5 is ta, 6 is bf, 7 is lateral gastroc

clear all

filename = input('Enter name of file with extension','s');
file = xlsread(filename);

baseline = input ('Enter name of the baseline file corresponding to trial','s');
fileb = xlsread (baseline);

frame = file(:,1);
index = file(:,2);
footswitch = file(:,3);

vl = file(7:30006,4);
ta = file(7:30006,5);
bf=file(7:30006,6);
lg=file(7:30006,7);

vlb = fileb(7:30006,4);
tab = fileb(7:30006,5);
bfb=fileb(7:30006,6);
lgb=fileb(7:30006,7);

vlb_ave = mean(vlb);
vlb_sd = std(vlb);
tab_ave = mean (tab);
tab_sd = std (tab);
bfb_ave = mean (bfb);
bfb_sd = std (bfb);
lgb_ave = mean (lgb);
lgb_sd = std (lgb);

vl  = vl - vlb_ave;
ta = ta-tab_ave;
bf = bf - bfb_ave;
lg = lg - lgb_ave;
%This part of the program determines at which frame # the signal goes high, 
%which designates heel strike. Finds the frame where the signal goes from 
%negative to positive and steps back 250pts to find actual heel strike 
k=1;
m=1;
for i=6:1:(length(index)-1000)
    if index(i) < (-4.5) && index(i-5)>(-4.5)&& m==1;
        high(k)= (i);
        k=k+1;
        m=2;
    end
    if m==2 && i>(high(k-1)+1000)
        m=1;
    end
end

%The following code determines the cadence  (speed of each subject can be 
%calculated by determining stride length from kinematic data)

for k=1:length(k)
    ptsperstride(k) = high(k+1)-high(k);
end
avgptsperstride = mean(ptsperstride);
shift = avgptsperstride*0.1634;

strides = 1/(avgptsperstride/1000);
cadence = strides*2;
shift = round(shift);
high = high+shift;

%This part of the program breaks each EMG signal into individual gait 
%cycles 

v1_1=v1(high(1):high(2));
v1_2=v1(high(2):high(3));
v1_3=v1(high(3):high(4));
v1_4=v1(high(4):high(5));
v1_5=v1(high(5):high(6));
v1_6=v1(high(6):high(7));
v1_7=v1(high(7):high(8));
v1_8=v1(high(8):high(9));
v1_9=v1(high(9):high(10));
v1_10=v1(high(10):high(11));

v1_1=v1(high(1):high(2));
ta_2=ta(high(2):high(3));
ta_3=ta(high(3):high(4));
ta_4=ta(high(4):high(5));
ta_5=ta(high(5):high(6));
ta_6=ta(high(6):high(7));
ta_7=ta(high(7):high(8));
ta_8=ta(high(8):high(9));
ta_9=ta(high(9):high(10));
ta_10=ta(high(10):high(11));

bf_1=bf(high(1):high(2));
bf_2=bf(high(2):high(3));
bf_3=bf(high(3):high(4));
bf_4=bf(high(4):high(5));
bf_5=bf(high(5):high(6));
bf_6=bf(high(6):high(7));
bf_7=bf(high(7):high(8));
bf_8=bf(high(8):high(9));
bf_9=bf(high(9):high(10));
bf_10=bf(high(10):high(11));

lg_1=lg(high(1):high(2));
lg_2=lg(high(2):high(3));
lg_3=lg(high(3):high(4));
lg_4=lg(high(4):high(5));
lg_5=lg(high(5):high(6));
lg_6=lg(high(6):high(7));
lg_7=lg(high(7):high(8));
lg_8=lg(high(8):high(9));
lg_9=lg(high(9):high(10));
lg_10=lg(high(10):high(11));

fs_1=footswitch(high(1):high(2));
fs_2=footswitch(high(2):high(3));
fs_3=footswitch(high(3):high(4));
fs_4=footswitch(high(4):high(5));
fs_5=footswitch(high(5):high(6));
fs_6=footswitch(high(6):high(7));
fs_7=footswitch(high(7):high(8));
fs_8=footswitch(high(8):high(9));
fs_9=footswitch(high(9):high(10));
fs_10=footswitch(high(10):high(11));

%This part of the program makes each gait cycle the same length.
%vl
ln=length(vl_1);
x1=1:1:ln;
xx=spline(x1,vl_1);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
vl_norm(:,1)=yy;
```matlab
ln=length(vl_2);
x1=1:1:ln;
xx=spline(x1,vl_2);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
vl_norm(:,2)=yy;

ln=length(vl_3);
x1=1:1:ln;
xx=spline(x1,vl_3);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
vl_norm(:,3)=yy;

ln=length(vl_4);
x1=1:1:ln;
xx=spline(x1,vl_4);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
vl_norm(:,4)=yy;

ln=length(vl_5);
x1=1:1:ln;
xx=spline(x1,vl_5);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
vl_norm(:,5)=yy;

ln=length(vl_6);
x1=1:1:ln;
xx=spline(x1,vl_6);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
vl_norm(:,6)=yy;

ln=length(vl_7);
x1=1:1:ln;
xx=spline(x1,vl_7);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
vl_norm(:,7)=yy;

ln=length(vl_8);
x1=1:1:ln;
xx=spline(x1,vl_8);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
vl_norm(:,8)=yy;

ln=length(vl_9);
x1=1:1:ln;
xx=spline(x1,vl_9);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
vl_norm(:,9)=yy;
```

ln=length(vl_10);
x1=1:1:ln;
xx=spline(x1,vl_10);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
vl_norm(:,10)=yy;

%BF

ln=length(bf_1);
x1=1:1:ln;
xx=spline(x1,bf_1);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
bf_norm(:,1)=yy;

ln=length(bf_2);
x1=1:1:ln;
xx=spline(x1,bf_2);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
bf_norm(:,2)=yy;

ln=length(bf_3);
x1=1:1:ln;
xx=spline(x1,bf_3);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
bf_norm(:,3)=yy;

ln=length(bf_4);
x1=1:1:ln;
xx=spline(x1,bf_4);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
bf_norm(:,4)=yy;

ln=length(bf_5);
x1=1:1:ln;
xx=spline(x1,bf_5);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
bf_norm(:,5)=yy;

ln=length(bf_6);
x1=1:1:ln;
xx=spline(x1,bf_6);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
bf_norm(:,6)=yy;

ln=length(bf_7);
x1=1:1:ln;
xx=spline(x1,bf_7);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
bf_norm(:,7)=yy;

ln=length(bf_8);
x1=1:1:ln;
xx=spline(x1,bf_8);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
bf_norm(:,8)=yy;

ln=length(bf_9);
x1=1:1:ln;
xx=spline(x1,bf_9);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
bf_norm(:,9)=yy;

ln=length(bf_10);
x1=1:1:ln;
xx=spline(x1,bf_10);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
bf_norm(:,10)=yy;

%TA

ln=length(ta_1);
x1=1:1:ln;
xx=spline(x1,ta_1);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
ta_norm(:,1)=yy;

ln=length(ta_2);
x1=1:1:ln;
xx=spline(x1,ta_2);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
ta_norm(:,2)=yy;

ln=length(ta_3);
x1=1:1:ln;
xx=spline(x1,ta_3);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
ta_norm(:,3)=yy;

ln=length(ta_4);
x1=1:1:ln;
xx=spline(x1,ta_4);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
ta_norm(:,4)=yy;
ln=length(ta_5);
x1=1:1:ln;
xx=spline(x1,ta_5);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
ta_norm(:,5)=yy;

ln=length(ta_6);
x1=1:1:ln;
xx=spline(x1,ta_6);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
ta_norm(:,6)=yy;

ln=length(ta_7);
x1=1:1:ln;
xx=spline(x1,ta_7);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
ta_norm(:,7)=yy;

ln=length(ta_8);
x1=1:1:ln;
xx=spline(x1,ta_8);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
ta_norm(:,8)=yy;

ln=length(ta_9);
x1=1:1:ln;
xx=spline(x1,ta_9);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
ta_norm(:,9)=yy;

ln=length(ta_10);
x1=1:1:ln;
xx=spline(x1,ta_10);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
ta_norm(:,10)=yy;

%LG

ln=length(lg_1);
x1=1:1:ln;
xx=spline(x1,lg_1);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
lg_norm(:,1)=yy;

ln=length(lg_2);
x1=1:1:ln;
xx=spline(x1,lg_2);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
lg_norm(:,2)=yy;

ln=length(lg_3);
      x1=1:1:ln;
      xx=spline(x1,lg_3);
      x2=ln/2500:ln/2500:ln;
      yyy=ppval(xxx,xx);
      lg_norm(:,3)=yy;

ln=length(lg_4);
      x1=1:1:ln;
      xx=spline(x1,lg_4);
      x2=ln/2500:ln/2500:ln;
      yyy=ppval(xxx,xx);
      lg_norm(:,4)=yy;

ln=length(lg_5);
      x1=1:1:ln;
      xx=spline(x1,lg_5);
      x2=ln/2500:ln/2500:ln;
      yyy=ppval(xxx,xx);
      lg_norm(:,5)=yy;

ln=length(lg_6);
      x1=1:1:ln;
      xx=spline(x1,lg_6);
      x2=ln/2500:ln/2500:ln;
      yyy=ppval(xxx,xx);
      lg_norm(:,6)=yy;

ln=length(lg_7);
      x1=1:1:ln;
      xx=spline(x1,lg_7);
      x2=ln/2500:ln/2500:ln;
      yyy=ppval(xxx,xx);
      lg_norm(:,7)=yy;

ln=length(lg_8);
      x1=1:1:ln;
      xx=spline(x1,lg_8);
      x2=ln/2500:ln/2500:ln;
      yyy=ppval(xxx,xx);
      lg_norm(:,8)=yy;

ln=length(lg_9);
      x1=1:1:ln;
      xx=spline(x1,lg_9);
      x2=ln/2500:ln/2500:ln;
      yyy=ppval(xxx,xx);
      lg_norm(:,9)=yy;

ln=length(lg_10);
      x1=1:1:ln;
xx=spline(x1,lg_10);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
lg_norm(:,10)=yy;

%fs

ln=length(fs_1);
x1=1:1:ln;
xx=spline(x1,fs_1);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
fs_norm(:,1)=yy;

ln=length(fs_2);
x1=1:1:ln;
xx=spline(x1,fs_2);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
fs_norm(:,2)=yy;

ln=length(fs_3);
x1=1:1:ln;
xx=spline(x1,fs_3);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
fs_norm(:,3)=yy;

ln=length(fs_4);
x1=1:1:ln;
xx=spline(x1,fs_4);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
fs_norm(:,4)=yy;

ln=length(fs_5);
x1=1:1:ln;
xx=spline(x1,fs_5);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
fs_norm(:,5)=yy;

ln=length(fs_6);
x1=1:1:ln;
xx=spline(x1,fs_6);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
fs_norm(:,6)=yy;

ln=length(fs_7);
x1=1:1:ln;
xx=spline(x1,fs_7);
x2=ln/2500:ln/2500:ln;
yy=ppval(xx,x2);
fs_norm(:,7)=yy;
% The following code averages all of the gait cycles for each signal, which
% leaves one array.
vl_avg = mean(vl_norm, 2);
bf_avg = mean(bf_norm, 2);
ta_avg = mean(ta_norm, 2);
lg_avg = mean(lg_norm, 2);
fs_avg = mean(fs_norm, 2);

vl_thresh = 1:2500;
ta_thresh = 1:2500;
bf_thresh = 1:2500;
lg_thresh = 1:2500;

for p = 1:2500;
    vl_thresh(p) = 20;
    ta_thresh(p) = 20;
    bf_thresh(p) = 20;
    lg_thresh(p) = 20;
end

% The following code multiplies the baseline standard deviation by 5, which creates a threshold to determine if significant muscle activity is present in the signal.
v1_t = 1:2500;
ta_t = 1:2500;
bf_t = 1:2500;
lg_t = 1:2500;
for w=1:2500;
v1_t(w) = 5*v1b_sd;
ta_t(w) = 5*tab_sd;
bf_t(w) = 5*bfb_sd;
lg_t(w) = 5*lgb_sd;
end
gc = 0:100/2499:100;

figure(1)
subplot(2,2,1), plot(gc,vl_avg)
title ('vastus lateralis')
hold on
subplot(2,2,1), plot(gc,vl_t,'r')
subplot(2,2,2), plot(gc,bf_avg)
title ('biceps femoris')
hold on
subplot(2,2,2), plot(gc,bf_t,'r')
subplot(2,2,3), plot(gc,ta_avg)
title ('tibialis anterior')
hold on
subplot(2,2,3), plot(gc,ta_t,'r')
subplot(2,2,4), plot(gc,lg_avg)
title ('lateral gastrocnemius')
hold on
subplot(2,2,4), plot(gc,lg_t,'r')

%Normalize emg to peak emg during activity to so that comparison
between
%subjects can be made

vl_max = max (vl_avg);
vl_n = (vl_avg / vl_max)*100;
ta_max = max (ta_avg);
ta_n = (ta_avg/ta_max)*100;
bf_max = max (bf_avg);
bf_n = (bf_avg/bf_max)*100;
lg_max = max(lg_avg);
lg_n = (lg_avg/lg_max)*100;

%Determine when each signal is on, based on 20% max being considered
on.
vl_onset=1:2500;
ta_onset=1:2500;
bf_onset=1:2500;
lg_onset=1:2500;

for kk = 1:2500
    if vl_n(kk) >20
        vl_onset(kk)= 50;
    else
        vl_onset(kk)=0;
    end
    if ta_n(kk) >20
        ta_onset(kk)= 50;
    else
        ta_onset(kk)=0;
end
end
if bf_n(kk) >20
    bf_onset(kk)= 50;
else
    bf_onset(kk)=0;
end
if lg_n(kk) >20
    lg_onset(kk)= 50;
else
    lg_onset(kk)=0;
end
end

figure (2)
subplot(2,2,1), plot(gc,vl_n)
title ('vastus lateralis')
hold on
subplot(2,2,1), plot(gc,vl_thresh,'r')
subplot(2,2,1), plot(gc,vl_onset)
subplot(2,2,2), plot(gc,bf_n)
title ('biceps femoris')
hold on
subplot(2,2,2), plot(gc,bf_thresh,'r')
subplot(2,2,2), plot(gc,bf_onset)
subplot(2,2,3), plot(gc,ta_n)
title ('tibialis anterior')
hold on
subplot(2,2,3), plot(gc,ta_thresh,'r')
subplot(2,2,3), plot(gc,ta_onset)
subplot(2,2,4), plot(gc,lg_n)
title ('lateral gastrocnemius')
hold on
subplot(2,2,4), plot(gc,lg_thresh,'r')
subplot(2,2,4), plot(gc,lg_onset)

%Following code calculates the area under the curve separately for the 7
%phases of gait

auc_vl = 1:2499;
auc_ta =1:2499;
auc_bf = 1:2499;
auc_lg = 1:2499;

for ii= 1:2499;
    if vl_n(ii)<0
        vl_n(ii)=0;
    end
    if ta_n(ii)<0
        ta_n(ii)=0;
    end
    if bf_n(ii)<0

bf_n (ii)= 0;
end
if lg_n(ii) <0
  lg_n(ii) = 0;
end
auc_vl(ii)=0.5*abs(vl_n(ii+1)+vl_n(ii))*(100/2499);
auc_ta(ii)=0.5*abs(ta_n(ii+1)+ta_n(ii))*(100/2499);
auc_bf(ii)=0.5*abs(bf_n(ii+1)+bf_n(ii))*(100/2499);
auc_lg(ii)=0.5*abs(lg_n(ii+1)+lg_n(ii))*(100/2499);
end

il_vl = sum(auc_vl(1:249));
il_ta = sum(auc_ta(1:249));
il_bf = sum(auc_bf(1:249));
il_lg = sum(auc_lg(1:249));

mst_vl = sum(auc_vl(250:599));
mst_ta = sum(auc_ta(250:599));
mst_bf = sum(auc_bf(250:599));
mst_lg = sum(auc_lg(250:599));

tst_vl = sum(auc_vl(600:1012));
tst_ta = sum(auc_ta(600:1012));
tst_bf = sum(auc_bf(600:1012));
tst_lg = sum(auc_lg(600:1012));

psw_vl = sum(auc_vl(1013:1249));
psw_ta = sum(auc_ta(1013:1249));
psw_bf = sum(auc_bf(1013:1249));
psw_lg = sum(auc_lg(1013:1249));

isw_vl = sum(auc_vl(1250:1674));
isw_ta = sum(auc_ta(1250:1674));
isw_bf = sum(auc_bf(1250:1674));
isw_lg = sum(auc_lg(1250:1674));

msw_vl = sum(auc_vl(1675:2074));
msw_ta = sum(auc_ta(1675:2074));
msw_bf = sum(auc_bf(1675:2074));
msw_lg = sum(auc_lg(1675:2074));

tsw_vl = sum(auc_vl(2075:2499));
tsw_ta = sum(auc_ta(2075:2499));
tsw_bf = sum(auc_bf(2075:2499));
tsw_lg = sum(auc_lg(2075:2499));

%This following code finds the peak for each channel of averaged EMG.
%Finds the maximum value before normalization and the index of that value.

[peak_vl,vl_time]=max(vl_avg);
[peak_ta,ta_time]=max(ta_avg);
[peak_bf,bf_time]=max(bf_avg);
[peak_lg,lg_time]=max(lg_avg);
vl_time = vl_time*(100/2499);
ta_time = ta_time*(100/2499);
bf_time = bf_time*(100/2499);
lg_time = lg_time*(100/2499);

%The following code calculates the total duration (%GC) that each channel of EMG is on. 'On' is defined as over 20% of peak.

dur_vl = (sum(vl_onset)/125000)*100;
dur_ta = (sum(ta_onset)/125000)*100;
dur_bf = (sum(bf_onset)/125000)*100;
dur_lg = (sum(lg_onset)/125000)*100;

%The following code calculates the total area under the curve for all four channels of emg.
auc_tot_vl=sum(auc_vl);
auc_tot_ta=sum(auc_ta);
auc_tot_bf=sum(auc_bf);
auc_tot_lg=sum(auc_lg);

filename2 = input('Enter name output file with extension','s');
xlswrite    (filename2, vl_avg, 01, 'a1')
xlswrite    (filename2, ta_avg, 01, 'b1')
xlswrite    (filename2, bf_avg, 01, 'c1')
xlswrite    (filename2, lg_avg, 01, 'd1')
xlswrite    (filename2, vl_t', 01, 'f1')
xlswrite    (filename2, ta_t', 01, 'g1')
xlswrite    (filename2, bf_t', 01, 'h1')
xlswrite    (filename2, lg_t', 01, 'i1')
xlswrite    (filename2, vl_n, 02, 'a1')
xlswrite    (filename2, ta_n, 02, 'b1')
xlswrite    (filename2, bf_n, 02, 'c1')
xlswrite    (filename2, lg_n, 02, 'd1')
xlswrite    (filename2, vl_thresh', 02, 'f1')
xlswrite    (filename2, ta_thresh', 02, 'g1')
xlswrite    (filename2, bf_thresh', 02, 'h1')
xlswrite    (filename2, lg_thresh', 02, 'i1')
xlswrite    (filename2, il_vl, 03, 'a1')
xlswrite    (filename2, il_ta, 03, 'b1')
xlswrite    (filename2, il_bf, 03, 'c1')
xlswrite    (filename2, il_lg, 03, 'd1')
xlswrite    (filename2, mst_vl, 03, 'a2')
xlswrite    (filename2, mst_ta, 03, 'b2')
xlswrite    (filename2, mst_bf, 03, 'c2')
xlswrite    (filename2, mst_lg, 03, 'd2')
xlswrite    (filename2, tst_vl, 03, 'a3')
xlswrite (filename2, tst_ta, 03, 'b3')
xlswrite (filename2, tst_bf, 03, 'c3')
xlswrite (filename2, tst_lg, 03, 'd3')
xlswrite (filename2, psw_vl, 03, 'a4')
xlswrite (filename2, psw_ta, 03, 'b4')
xlswrite (filename2, psw_bf, 03, 'c4')
xlswrite (filename2, psw_lg, 03, 'd4')
xlswrite (filename2, isw_vl, 03, 'a5')
xlswrite (filename2, isw_ta, 03, 'b5')
xlswrite (filename2, isw_bf, 03, 'c5')
xlswrite (filename2, isw_lg, 03, 'd5')
xlswrite (filename2, msw_vl, 03, 'a6')
xlswrite (filename2, msw_ta, 03, 'b6')
xlswrite (filename2, msw_bf, 03, 'c6')
xlswrite (filename2, msw_lg, 03, 'd6')
xlswrite (filename2, tsw_vl, 03, 'a7')
xlswrite (filename2, tsw_ta, 03, 'b7')
xlswrite (filename2, tsw_bf, 03, 'c7')
xlswrite (filename2, tsw_lg, 03, 'd7')
xlswrite (filename2, auc_tot_vl, 03, 'a8')
xlswrite (filename2, auc_tot_ta, 03, 'b8')
xlswrite (filename2, auc_tot_bf, 03, 'c8')
xlswrite (filename2, auc_tot_lg, 03, 'd8')
xlswrite (filename2, peak_vl, 04, 'a1')
xlswrite (filename2, vl_time, 04, 'a2')
xlswrite (filename2, peak_ta, 04, 'b1')
xlswrite (filename2, ta_time, 04, 'b2')
xlswrite (filename2, peak_bf, 04, 'c1')
xlswrite (filename2, bf_time, 04, 'c2')
xlswrite (filename2, peak_lg, 04, 'd1')
xlswrite (filename2, lg_time, 04, 'd2')
xlswrite (filename2, dur_vl, 05, 'a1')
xlswrite (filename2, dur_ta, 05, 'b1')
xlswrite (filename2, dur_bf, 05, 'c1')
xlswrite (filename2, dur_lg, 05, 'd1')
CHAPTER 4: It is feasible for ambulatory individuals with hemiparesis post stroke to practice walking on the elliptical based robotic gait trainer (EBRGT)

Abstract

Stroke is one of the leading causes of disability in the United States with 750,000 individuals affected each year (1). A major residual effect of stroke is impaired walking ability. Restoration of gait is a major goal of rehabilitation for persons with stroke. The current focus for gait restoration due to paralysis associated with stroke is on task specific, repetitive rehabilitation techniques. Our group developed a low-cost robotic device that has the ability to simulate walking. The purpose of this study was to determine the feasibility of using the newly developed elliptically based robotic gait trainer (EBRGT) to retrain gait in a hemiparetic stroke population. Six adult subjects (mean age=64.5±8.8 years) with varying levels of hemiparesis secondary to stroke (mean gait speed=76.0±24.8 cm/s) walked on the EBRGT at their preferred speed. All 6 subjects were able to ambulate on the EBRGT with minimal assistance. Some required early assistance to spin the flywheel, but later became independent as data collection progressed. Based on a survey taken at the end of the study, all 6 subjects said that they were interested in using the device again. Kinematic results were similar to normal
healthy adult subjects walking on the EBRGT. It is feasible for ambulatory patients with hemiparesis secondary to stroke to walk on the newly developed EBRGT with minimal assistance. Ongoing studies are being performed to determine if gait training on the EBRGT can improve gait parameters in individuals with hemiparesis resulting from stroke.

Introduction

Stroke is one of the leading causes of disability in the United States with 750,000 individuals affected each year (Williams et al., 1999). Many of the subjects that survive stroke are left with severe disability (Rosamond et al., 2007). These deficits include hemiparesis, a weakness on one side of the body, which can impair their ability to walk. In a study by Lord et al. (2004), over 90% of the subjects who had suffered stroke considered the ability to get out and about in the community to be important and it was considered essential by 40% of the participants. Restoration of gait is a central goal in rehabilitation of persons following stroke.

The current focus for gait restoration after stroke is on task specific, repetitive rehabilitation techniques. Task specific training for walking involves the subject, with assistance, moving their lower-limbs in a pattern similar to normal gait while bearing part or all of their body weight. Over-ground gait training, body weight supported treadmill training (BWSTT), and more recently, robotic gait training have been used to administer task specific gait training in patients with stroke. Body-weight supported treadmill training has been shown to improve gait in hemiparetic stroke subjects (Sullivan et al.,
2007) and may result in better outcomes than over ground or conventional physical therapy (Visintin et al., 1998, Ada et al., 2003, Salbach et al., 2004, Dean et al., 2000, Richards et al., 1993).

BWSTT also has the potential to improve cardiovascular fitness. Unfortunately, it has failed to become widely accepted because of the high staffing costs required to perform the treatment (Morrison and Backus, 2007). The field of robot-assisted motor rehabilitation has emerged as an alternative to therapist-assisted training (Winchester and Querry, 2006). In this type of training, a machine guides the lower extremities in a walking like pattern while the body weight of the subject is supported. Unfortunately, these interventions are often not available due to the cost and complexity of the devices.

In response to the potential benefits of task specific training in rehabilitation of gait after stroke and the need for affordable, simple interventions our group designed a new semi-robotic gait trainer. The gait trainer is both affordable and simple to use and has features that encompass characteristics of both BWSTT and robotic gait training.

The platform for the new device is a commercially available elliptical trainer. The elliptical trainer was chosen for three reasons 1) studies have shown that sagittal plane hip and knee kinematics correlate well with sagittal plan kinematics during level surface walking (Burnfield et al., 2010, Bradford et al., 2007), thus it has features that mimic gait, 2) elliptical trainers have a mechanical linkage between the left and right sides, similar to a bicycle, thus patients with hemiparesis could help advance their affected limb with their unaffected lower extremity and 3) it employs a distal control
mechanism to help guide the lower limbs, but still requires the subject to control their trunk position and lateral weight shifting.

While this gait trainer has been tested with a normal healthy adult population (Chapter 3), it has not been tested in the target population: patients with hemiparesis secondary to stroke. One study examined the feasibility of using elliptical trainers to train subjects with hemiparesis due to stroke and found that the activity was feasible and safe (Jackson et al., 2010). The base for our device is an elliptical trainer and retains some of the features of a commercially available elliptical; however, we have also made modifications that warrant testing.

To date, no one has performed gait analysis of subjects with hemiparesis while using an elliptical trainer. Movement analysis has been performed on normal healthy adults using elliptical trainers in several studies (Lu et al., 2007, Burnfield et al., 2010, Bradford et al., 2007), but it is unknown whether subjects with hemiparesis will perform like normal healthy adults. The purpose of this study was to determine the feasibility of subjects with stroke walking on the EBRGT and if their movement patterns are similar to normal healthy adults using the same device.

Methods

Participants

Six adult subjects (mean age=64.5±8.8 years) with hemiparesis secondary to stroke (mean gait speed=76.0±24.8 cm/s) were recruited through established relationships
with physical therapy rehabilitation clinics and physicians at both the Medical College of Virginia Hospital (Richmond, VA) and the Sheltering Arms Rehabilitation Center (Richmond, VA). All participants were able to ambulate at least 10 meters with or without assistive devices and were free from any other neurological, orthopedic, or unstable cardiovascular disease. After informed consent was obtained, patient history and evaluation were performed (See Table1). All participants had hip, knee, and ankle passive range of motion that were within normal limits.

Device Design

The footplates of the elliptical trainer were modified to adjust the sagittal plane ankle kinematics during use of the elliptical trainer to a pattern that was intended to better mimic level surface walking. The mounts for both the left and right footplates were modified so that each footplate was attached to the ski (Figure 1) using a single axis bearing. This allowed the footplates to tip up and down in the sagittal plane. The new footplate was mounted several inches off of the ski to allow for adequate range of motion to mimic level surface walking. The modified elliptical trainer or EBRGT is depicted in Figure 2.

The motion of the footplates was controlled using a mechanical push/pull mechanism coupled with a servo motor. The control signal was derived from an optical encoder placed on the flywheel/crank (Figure 1) of the elliptical trainer. One complete turn of the flywheel/crank represented one complete gait cycle. The position of the flywheel was aligned with the events of level surface walking. Thus the position of the
footplate was determined by the position of the flywheel. For example, the most anterior position of the right footplate is correlated with initial contact during level surface walking. As noted in the control block diagram in figure 3, each footplate is driven by its own controller however; the controllers receive identical input from a single optical encoder on the flywheel.

The device works as follows: The user applies force to the footplates and the flywheel begins moving, as the flywheel turns, an encoder provides flywheel position input to the controllers, based on the position of the flywheel; the controllers determine the new position for the footplates and send out a command for the new footplate position. Thus no motion will occur without input from the user and the gait velocity is controlled by the user.

**Procedures**

Video data, electromyography (EMG), and over ground temporal gait data were collected during a single one hour session in a motion analysis laboratory. Before data collection, subjects walked on the elliptically based robotic gait trainer (EBRGT) until they felt comfortable with the activity. If participants wore an ankle-foot orthosis (AFO), it was removed before walking on the EBRGT. After habituation to the EBRGT, self-adhesive Ag/Cl electrodes with a sensor area of 13.2mm² (Blue Sensor M, Ambu, Denmark) were placed over 4 muscles of both lower extremities: vastus lateralis muscle (VL), tibialis anterior muscle (TA), biceps femoris muscle (BF), and lateral gastrocnemius muscle (LG) (Delagi and Perotto, 1980). Before application of the electrodes, the area was
shaved and cleaned with alcohol. Electrodes were spaced with a 34mm center-to-center distance. A reference electrode was placed over the fibular head on the same side of the body as the other electrodes. Four channels of EMG data were collected simultaneously at 1000 Hz using the Myosystem 1200 (Noraxon, Scottsdale, Az). EMG was only collected for 3 of the 6 participants. Reflective markers were placed on both the left and right sides of the subject with double sided tape. Markers were placed over the lateral border of the acromion, the greater trochanter, lateral epicondyle of the femur, fibular head, lateral malleolus, the lateral aspect of the calcaneus, and the lateral aspect of the head of the fifth metatarsal. Landmark locations were palpated using standard techniques. Two high speed digital cameras (Bassler Scout, Bassler Inc., Exton, PA) placed 10 feet away from the EBRGT and perpendicular to the subject’s left and right side were used to capture video at 120hz.

Each participant walked on the EBRGT for a total of either about 1.5 minutes or 3.5 minutes, depending on if EMG data was collected. During use of the EBRGT, participants wore a Biodex™ unweighing harness (Biodex, Shirley, NY) which was attached to an overhead support system for safety. No body weight was supported by the harness. A toe strap was used to keep the participant’s feet in contact with the footplates and they were instructed to keep their heels in contact with the footplates. Participants were instructed to find a comfortable pace and maintain that pace during data collection. Subjects were given the option to hold onto a stationary handle in front of them or the reciprocating handles that were mechanically linked to the ski. At the end of the first minute, 14 seconds of video data was collected from the left and right side simultaneously. For three of the six subjects, EMG data was collected after video data
was collected. Once video data collection was completed, the participant was asked to stop and leads were attached to the 4 sets of electrodes on one lower extremity. The participant was asked to walk for another minute. During that time, 30 seconds of EMG data were captured. This process was repeated for the opposite lower extremity. A 30 second baseline EMG signal was also captured with the subject resting in a chair with their feet flat on the ground before walking on the EBRGT.

Over ground temporal gait data was collected using the GAITRite™ system (CIR Systems Inc, Havertown, PA). This is a carpet like walkway with instrumentation to detect footfalls and software to calculate temporal-spatial gait parameters. Participants were told to walk at a comfortable pace down the 10m walkway. The GAITRite™ instrumented mat was centered within the walkway. Each participant made 4 passes and the results were averaged.

After the participants walked on the EBRGT, an acceptability questionnaire was administered (See Appendix).

**Kinematic data processing**

Reflective markers were tracked and digitized using Max TRAQ 2D software (Innovision Systems Inc., Columbiaville, MI). Data was then exported and processed using custom MATLAB software (The MathWorks, Natick, MA). The kinematic data were first low-pass filtered using a fourth-order-zero phase lag Butterworth filter with a cutoff frequency of 6Hz (Winter, 1990). Orthopedic sagittal plane joint angles for the hip, knee, and ankle were calculated. Sagittal plane thigh angles with respect to a vertical
line passing through the hip were also calculated. This was done to examine the action of the thigh body segment without the interaction of the trunk position. The sagittal plane angle of the foot with respect to the horizon was also calculated since this was the parameter that we were controlling with the modifications to the elliptical trainer.

Ten gait cycles were extracted and time-normalized with 0-100% representing a full gait cycle: 0% of the gait cycle represented initial contact. Since no initial contact occurs on the EBRGT, the most positive x-position of the lateral malleolus marker also represented 0% of the gait cycle; this marked the transition where the distal part of the lower extremity changed directions from moving in an anterior direction to moving in a posterior direction. This represents the transition from ‘swing’ to ‘stance’.

**EMG data processing**

EMG data was captured and stored on a PC for post-processing using MotionMonitor V7.0 (Innovative Sports Training, Chicago, IL). To determine beginning of the gait cycle within the EMG data, an optical encoder was used during EBRGT walking. The signals of both the optical encoder and EMG were sampled at 1000 Hz and synchronized with the EMG data. EMG data was high pass filtered with a 20 Hz cutoff frequency and low pass filtered with a 200 Hz cutoff frequency. A root-mean-square (RMS) with a 25ms window was performed. The data was then exported in plain text format for further processing with custom MATLAB software (The Math Works, Natick, MA).
In order to determine if significant muscle activity was present, each EMG signal was tested to see if it contained data that was greater than 5x the standard deviation of baseline EMG recordings for the same muscle. After determining if there was activity present, the EMG signal for each subject was normalized to the maximum EMG signal during the activity so that data could be compared between subjects. The signal was then broken up into gait cycles and time normalized for averaging as described for kinematic data. The time-normalized cycles were averaged for 10 gait cycles for each participant.

Results

All six subjects with hemiparesis were able to walk on the EBRGT safely with little or no assistance. On occasion, assistance was needed to initiate movement. One participant needed help keeping the heel of their affected leg from turning inwards (tibial external rotation). Note that this participant had little to no sensation in their affected lower extremity. All subjects held onto the reciprocating handles of the EBRGT. Three subjects were only able to hold onto one handle due to limitations of their affected upper extremity. One or two assistants were required to help the participants safely mount and dismount the EBRGT. Once mounted, only one assistant tended to the subject as needed.

Temporal and spatial gait parameters

The step length while walking on the EBRGT was fixed at 36cm. Subjects were allowed to self-select their most comfortable pace (Table 2). Mean cadence was 0.64 ±
0.12 strides/sec with a range of 0.44-0.81 strides/second during EBRGT walking. During over ground walking mean cadence was 0.73±0.15 strides/sec with a range of 0.40-0.87 strides/sec. Mean over ground step length including both the left and right side of all participants was 50.4±11.3 cm with a range of 27.6-62.3 cm.

Kinematics

Kinematic data of the participants with hemiparesis were compared to data from a group of normal healthy subjects who walked on the same device in a previous study (Chapter #3). Thigh, hip, knee, ankle and foot angles all revealed similar movement patterns between participants with stroke and normal healthy adults (Figure 5). Thigh, hip, and foot angles were the most similar between groups. Knee and ankle angles (Figure 5C and 5D) of participants with stroke displayed more variation with respect to the normal healthy subjects than other joint angles. Four of the six participants with stroke displayed greater knee flexion than the normal healthy adults during the stance phase of gait (0-60%). During swing phase (61-100%), the knee angle of 5 of the 6 participants with hemiparesis had a lower peak flexion angle than the normal healthy adults.

Electromyography

EMG data for the participants with hemiparesis were also compared to data from a group of normal healthy subjects who walked on the same device in a previous study
EMG profiles for participants with hemiparesis were similar to the normal healthy subjects. However, participant #4 did not have any TA EMG activity on their affected side during EBRGT walking. Figure 6 depicts the EMG profile for participant #2 and an average EMG profile from a group of normal healthy subjects. Participant #2 had activity in all four muscles measured, VL, TA, BF, and LG, during EBRGT walking. The VL EMG profile (Figure 5A) for participant #2 seemed to have a higher level of activity from 30-60% of the gait cycle when compared to normal healthy adults. The other notable difference was in the LG (Figure 5D). In the last 40% of the gait cycle there seems to be a higher level of activation of the LG for participant #2 than for the normal healthy subjects. The other two participants (3 and 4), had LG and VL EMG profiles that more closely resembled the pattern of the normal healthy subjects.

Questionnaire results

All six participants filled out the acceptability questionnaire, of those, two required assistance from a researcher to physically fill in the answers. All six participants answered yes when asked if they would use the device again. Five of the six participants agreed or strongly agreed that the device was easy to use. One of the six participants disagreed with the statement that the device was easy to use. Four of the six participants experienced pain greater than 0 on a visual analog scale (VAS). Most of the participants listed the location of their pain as being muscle fatigue pain in the knee extensors. One participant listed the pain as being in the knee joint and one participant listed pain as
being located in the shoulder. All participants listed the effort while walking on the EBRGT as being moderate to high.

Discussion

All six participants with hemiparesis were able to safely walk on the EBRGT with movement patterns and muscle firing patterns that were similar to normal healthy adults walking on the same device. The purpose of the device is to help patients with hemiparesis practice a weight bearing, gait like movement. In a previous study gait kinematics and muscle firing patterns of normal healthy adults were compared between walking on the EBRGT and walking on a treadmill (Chapter 3). That study found many similarities between the two activities. In the current study, the joint kinematics and EMG of participants with hemiparesis were compared to normal healthy subjects walking on the EBRGT. Both kinematics and EMG analysis revealed similar results for the participants with hemiparesis when compared to normal healthy subjects. This suggests that participants with hemiparesis secondary to stroke perform gait like movements and muscle firing patterns when walking on the EBRGT.

Overall, the participants exhibited EMG profiles for the VL, TA, BF, and LG that were similar to normal healthy subjects walking on the EBRGT. There were some slight differences in participant 2’s VL EMG profiles when compared to normal healthy subjects. Participant 2 was not able to grip both of the handles of the gait trainer due to limitations of their affected upper extremity. Only gripping one handle made it more difficult for the participant to balance and may have required his lower extremities to
assist more in stabilization. This may have contributed to some of the differences seen in muscle firing patterns. Patient 4 did not exhibit any significant TA activity where normal healthy subjects did have some TA activity (Chapter 3) during EBRGT walking. This may be due to the altered motor control that occurs after stroke (Knuttson and Richards, 1979).

While kinematic analysis revealed joint angle profiles for the participants with stroke that were similar to normal healthy adults, there was some variability at the knee and ankle. The EBRGT controls the foot movements, so it is expected that the foot angles would be very similar when comparing between groups walking on the EBRGT. The differences between joint kinematics between the individuals with stroke and healthy normal adults may be due to the impaired motor control and spasticity often present in individuals with stroke (Olney and Richards, 1996).

Self-selected cadence while walking on the EBRGT was only slightly slower than over-ground cadence; However, this resulted in a much slower gait speed than preferred over ground gait speeds since the step length of the EBRGT was fixed at 36cm. Some gait training studies have found positive outcomes from having subjects with hemiparesis post stroke practice walking at fast gait speeds (Pohl et al., 2002, Sullivan et al., 2002). While subjects preferred to walk at slower cadences and thus gait speeds, a benefit of this device is that the gait speed is controlled by the user and not limited by the device. Users could be coached to practice walking at a faster cadence and thus at a faster speed.

The questionnaire was not administered anonymously thus the results must be interpreted carefully. The goal of the questionnaire was to determine the acceptability of
the EBRGT by users with hemiparesis due to stroke. All of the participants said that they would use the gait training device again. The one participant that had knee joint pain stated that they often had this pain during walking and other daily activities. The one participant that had shoulder pain was one of two participants that could only hold onto one handle of EBRGT. This same participant had the slowest over-ground gait speed and thus may have been the lowest functioning (Perry et al., 1995) and described the device as being difficult to use. Adjustments may need to be made to the device or the exercise to make it easier for lower functioning users to more easily perform the activity. Off-loading some body weight or redesigning the handles may have made the activity easier for this participant. In future designs it may be beneficial to have a handle that is more central to the user’s body and stationary instead of moving to make it easier to stabilize with one hand.

The muscle fatigue experienced in the quadriceps was expected as the EBRGT elicits longer duration and stronger activations of the quadriceps relative to level surface walking (Chapter 3). Another study where patients with stroke walked on an elliptical trainer found similar results in terms of fatiguing the knee extensors (Jackson et al., 2010). In future designs, changing the trajectory of the footplate may change the direction of the ground reaction force and thus change the required knee extensor force. It may also be necessary to off-load some body weight of the user to reduce quadriceps muscle fatigue.
Conclusion

This study presents a newly developed, robotic gait trainer to help patients with stroke practice a gait like movement with minimal manual assistance during gait rehabilitation. It is feasible for ambulatory patients with hemiparesis secondary to stroke to walk on the newly developed EBRGT with minimal assistance. The new device was readily accepted by this group of participants, who all expressed an interest in using the device again. Studies are being conducted to determine the effects of gait training with the EBRGT on functional mobility outcomes in individuals with hemiparesis.
Table 1. Participant characteristics.

<table>
<thead>
<tr>
<th>Subject</th>
<th>Gender</th>
<th>Age</th>
<th>Side of Hemiparesis</th>
<th>Height (in)</th>
<th>Weight (lbs)</th>
<th>Preferred Gait speed</th>
<th>Time Since Stroke</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>F</td>
<td>69</td>
<td>L</td>
<td>64</td>
<td>144</td>
<td>92.4</td>
<td>3.5 months</td>
</tr>
<tr>
<td>2</td>
<td>M</td>
<td>64</td>
<td>R</td>
<td>72</td>
<td>198</td>
<td>89.1</td>
<td>45 months</td>
</tr>
<tr>
<td>3</td>
<td>M</td>
<td>78</td>
<td>L</td>
<td>70</td>
<td>200</td>
<td>96.8</td>
<td>3 months</td>
</tr>
<tr>
<td>4</td>
<td>M</td>
<td>58</td>
<td>R</td>
<td>69</td>
<td>168</td>
<td>53.9</td>
<td>7 months</td>
</tr>
<tr>
<td>5</td>
<td>F</td>
<td>68</td>
<td>R</td>
<td>60</td>
<td>134</td>
<td>30.8</td>
<td>4 months</td>
</tr>
<tr>
<td>6</td>
<td>M</td>
<td>50</td>
<td>L</td>
<td>69</td>
<td>200</td>
<td>93.1</td>
<td>11 months</td>
</tr>
</tbody>
</table>
Table 2. Self-selected cadence during EBRGT walking.

<table>
<thead>
<tr>
<th>Subject</th>
<th>strides/sec</th>
</tr>
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<tbody>
<tr>
<td>1</td>
<td>0.70</td>
</tr>
<tr>
<td>2</td>
<td>0.59</td>
</tr>
<tr>
<td>3</td>
<td>0.73</td>
</tr>
<tr>
<td>4</td>
<td>0.58</td>
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<tr>
<td>5</td>
<td>0.45</td>
</tr>
<tr>
<td>6</td>
<td>0.82</td>
</tr>
<tr>
<td>mean</td>
<td>0.64</td>
</tr>
<tr>
<td>st. dev.</td>
<td>0.12</td>
</tr>
<tr>
<td>range</td>
<td>0.44-0.81</td>
</tr>
</tbody>
</table>
**Table 3.** Results from acceptability questionnaire.

<table>
<thead>
<tr>
<th>Subject</th>
<th>The gait training device was easy to use during your session today:</th>
<th>You felt discomfort while using the gait training device:</th>
<th>VAS of pain</th>
<th>How much effort did using the gait training device require?</th>
<th>Would you use the gait training device again?</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Agree</td>
<td>Occasionally</td>
<td>5</td>
<td>High</td>
<td>Yes</td>
</tr>
<tr>
<td>2</td>
<td>Strongly agree</td>
<td>Very frequently</td>
<td>2</td>
<td>Moderate</td>
<td>Yes</td>
</tr>
<tr>
<td>3</td>
<td>Agree</td>
<td>Occasionally</td>
<td>5</td>
<td>High</td>
<td>yes</td>
</tr>
<tr>
<td>4</td>
<td>Agree</td>
<td>never</td>
<td>0</td>
<td>moderate</td>
<td>yes</td>
</tr>
<tr>
<td>5</td>
<td>Disagree</td>
<td>Occasionally</td>
<td>5</td>
<td>high</td>
<td>yes</td>
</tr>
<tr>
<td>6</td>
<td>agree</td>
<td>rarely</td>
<td>0</td>
<td>moderate</td>
<td>yes</td>
</tr>
</tbody>
</table>
Figure 1. The commercially available elliptical trainer before modifications.
Figure 2. EBRGT (elliptical trainer after modifications) with a normal healthy subject.
Figure 3. Control block diagram for modified elliptical gait trainer.
Figure 4. A study participant walking on the EBRGT.
**Figure 5.** Sagittal plane joint angles for the thigh (A), hip (B), knee (C), ankle (D), and foot (E) for one complete gait cycle, starting with initial contact. Only the hemiparetic side of each participant is displayed. For the thigh, hip and knee; flexion is represented by positive angles and extension is represented by negative angles. For the ankle, dorsiflexion is represented by positive angles and plantarflexion is represented by negative angles. The foot angles are calculated relative to the horizontal; positive angles represent when the toe is above the horizon and negative angles represent when the toe is below the horizon. Normals (normal healthy adults) error bars represent one standard deviation from the mean.
Figure 6. Ensemble averaged EMG profiles of participant #2 (right hemiparesis) during EBRGT walking for the vastus lateralis muscle (A), the tibialis anterior muscle (B), the biceps femoris muscle (C), and the lateral gastrocnemius muscle (D). The plots represent one full gait cycle starting with initial contact. Data from healthy normals (Chapter #3) are shown with error bars representing one standard deviation.
Works Cited


Appendix
Appendix A

Acceptability Questionnaire

The gait training device was easy to use during your session today:

Strongly agree
Agree
Undecided
Disagree
Strongly disagree

You felt discomfort while using the gait training device:

Always
Very Frequently
Occasionally
Rarely
Very Rarely
Never

If you experienced discomfort while using the gait training device, please rate the intensity (Place a mark on the line below indicating your pain):

0-----------------------------------------------10
No pain                                      Worst pain possible

How much effort did using the gait training device require?

Exhausting
High
Moderate
Low

Would you use the gait training device again?

Yes
No
Maybe
Undecided

If you would like, please elaborate on your experience using the gait training device:

_________________________________________________________________________________________________
_________________________________________________________________________________________________
_________________________________________________________________________________________________
_________________________________________________________________________________________________

_________________________________________________________________________________________________
Chapter 5: Gait training with the EBRGT produces improved gait parameters in ambulatory individuals with hemiparesis post-stroke: A case series.

Abstract

Stroke is one of the leading causes of disability in the United States with 750,000 individuals affected each year (1). A major residual effect of stroke is impaired walking ability. Restoration of gait is a major goal of rehabilitation for persons with stroke. The current focus for gait restoration due to paralysis associated with stroke is on task specific, repetitive rehabilitation techniques. Our group developed a low-cost robotic device that has the ability to simulate walking. The purpose of this study was to determine the feasibility of using the newly developed elliptically based robotic gait trainer (EBRGT) to retrain gait in a hemiparetic stroke population. Four ambulatory adult subjects with hemiparesis secondary to stroke trained on the EBRGT 3x/week for 4 or 8 weeks. All 4 subjects exhibited an increase in over ground gait speed after the intervention and were able to improve their training session distances. It is feasible for ambulatory patients with hemiparesis secondary to stroke to walk on the newly developed EBRGT with minimal assistance. Results suggest positive, clinically significant outcomes in measurable gait parameters and warrant larger controlled studies.
Introduction

Stroke is one of the leading causes of disability in the United States with 750,000 individuals affected each year (Williams et al., 1999). Many of the subjects that survive stroke are left with severe disability (Rosamond et al., 2007). These deficits include hemiparesis, a weakness on one side of the body, which can impair their ability to walk. In a study by Lord et al. (2004), over 90% of the subjects who had suffered stroke considered the ability to get out and about in the community to be important and it was considered essential by 40% of the participants. Restoration of gait is a central goal in rehabilitation of persons with stroke.

The current focus for gait restoration after stroke is on task specific, repetitive rehabilitation techniques. Task specific training for walking involves the subject, with assistance, moving their lower-limbs in a pattern similar to normal gait while bearing part or all of their body weight. Over-ground gait training, body weight supported treadmill training (BWSTT), and more recently, robotic gait training have been used to administer task specific gait training in patients with stroke. Body-weight supported treadmill training has been shown to improve gait in hemiparetic stroke subjects (Sullivan et al., 2007) and may result in better outcomes than over ground or conventional physical therapy (Visintin et al., 1998, Ada et al., 2003, Salbach et al., 2004, Dean et al., 2000, Richards et al., 1993).

BWSTT also has the potential to improve cardiovascular fitness. Unfortunately, it has failed to become widely accepted because of the high staffing costs required to perform
the treatment (Morrison and Backus, 2007). The field of robot-assisted motor rehabilitation has emerged as an alternative to therapist-assisted training (Winchester and Querry, 2006). In this type of training, a machine guides the lower extremities in a walking like pattern while the body weight of the subject is supported. Unfortunately, these interventions are often not available due to the cost and complexity of the devices.

In response to the potential benefits of task specific training in rehabilitation of gait after stroke and the need for affordable, simple interventions our group designed a new semi-robotic gait trainer (Chapter #2). The gait trainer is both affordable and simple to use and has features that encompass characteristics of both BWSTT and robotic gait training.

The Elliptically Based Robotic Gait Trainer (EBRGT) was developed in our lab from a Nordic Track CXT 910 elliptical trainer. While this gait trainer has been tested with a normal healthy adult population (Chapter #3), it has not been tested in the target population: patients with hemiparesis secondary to stroke. One study examined the feasibility of using elliptical trainers to train subjects with hemiparesis due to stroke and found that the activity was feasible and safe but did not exhibit improved gait speed or function (Jackson et al., 2010). The base for our device is an elliptical trainer and retains some of the mechanics of a commercially available elliptical (paper #1). Other studies have tested a robotic device that employs a similar distal control mechanism and have found the activity to be safe and even efficacious in improving gait and function in individuals with stroke (Werner et al., 2002, Pohl et al., 2007, Tong et al., 2006).
While our device is similar to commercially available elliptical trainers and other robotic devices it has some differences and thus testing in the target population is warranted. The purpose of this study was to assess the feasibility of implementing a gait training protocol using the EBRGT to improve functional walking ability in ambulatory individuals with hemiparesis secondary to stroke.

Methods

Participants
Four individuals with hemiparesis secondary to stroke were recruited through established relationships with physical therapy rehabilitation clinics and physicians at both the Medical College of Virginia Hospital (Richmond, VA) and the Sheltering Arms Rehabilitation Center (Richmond, VA). All participants had some walking impairment (decreased walking speed or use of an assistive device or orthosis) but were able to walk at least 10m with supervision. Participants were free from any other neurological, orthopedic, or unstable cardiovascular disease and were able to follow instructions. After informed consent was obtained, patient history and evaluation were performed (See Table 1). All participants had hip, knee, and ankle passive range of motion that were within normal limits.
Intervention

Training took place in a movement analysis laboratory setting in the Department of Physical Therapy at Virginia Commonwealth University. Participants attended training sessions 3x/week for at least 4 weeks and had the option to continue for 8 weeks. During each training session, the participants walked on the EBRGT (Chapter 3 and 4) as long as they were able to maintain a rating of perceived exertion (RPE 6-20) of ‘somewhat hard’ or less (Borg, 1982). Rest periods were permitted at the participant’s discretion in order to maintain the appropriate RPE. Heart rate was monitored and exercise was terminated if a maximal training HR value was exceeded (Karoven et al., 1957). Contact time with the gait trainer was limited to 30mins on any given training day.

During use of the EBRGT, participants wore a Biodex™ unweighing harness (Biodex, Shirley, NY) which was attached to an overhead support system for safety (Figure 1). No body weight was supported by the harness. Straps were used to secure the participant’s toes to the footplates and they were also instructed to keep their heels in contact with the footplates. A physical therapist was present during training to monitor the participant and provide any assistance if needed. See Figure 1 for setup during gait training sessions.
Outcome Measures

Measures of gait and functional ability were measured within one week prior to gait training and within one week after the 4 week intervention. Gait speed was determined using the 10m walk test (10MWT). This was measured using an instrumented walkway (GAITRite®, CIR systems; Clifton, New Jersey). The walkway resembles a long piece of carpet. It was placed in the middle of a 14m walkway. Participants were instructed to walk down the walkway at a comfortable pace 4 times and an average of the last two trials was used as the participant’s gait speed. Participants were allowed to use assistive devices or orthoses they typically used for community ambulation and were allowed rest periods if needed. The 10MWT is one of the most widely used methods of measuring walking ability both in the clinic and in research (Richards et al., 1999, Richards and Olney, 1996). The validity of gait speed as a measure of walking ability has been extensively studied (daCunha et al., 2002, Meada et al., 2000, Wolf et al, 1999). An ICC of 0.95 has been reported for gait speed in a stroke population (daCunha-Filho 2003). Gait speed has been found to be a strong predictor of community ambulation (Perry 1995).

Temporal and spatial gait parameters were also measured using the instrumented walkway (GAITRite®, CIR systems; Clifton, New Jersey). The GAITRite® data processing software was used to determine cadence, stride length, step length, swing, and stance times. The GAITRite® system has been found to be a valid tool for measuring both averaged and individual step parameters of gait (Webster et al., 2005). Gait
symmetry was calculated in terms of step length ratio (SLR), swing time ratio (SwR), and stance time ratio (StR). Each of these was calculated by dividing the value for the paretic extremity by that of the non-paretic extremity (Balasubramanian et al., 2007, Patterson et al., 2008).

Mobility status was determined by using the locomotor portion of the Functional Independence Measure (FIM-L). This test is widely used in rehabilitation to assess mobility. The test items have reliability coefficients of 0.86 to 0.97 (Stineman et al., 1996). See appendix for description of categories.

Results

All four subjects with hemiparesis were able to walk on the EBRGT safely with little or no assistance. After one or two training sessions, the only physical assistance necessary was to don/doff the safety harness and to mount and dismount the EBRGT safely. Some verbal coaching was used to remind the participants to keep their heels from coming up or to keep from crouching or leaning forward. All participants opted to hold onto the oscillating handles of the EBRGT. Participants 1 and 2 completed 4 weeks of training, 3x/week. After 4 weeks of training, participants 3 and 4 opted to continue for an additional 4 weeks, making their total intervention 8 weeks.

Participant 1 was present for 9 of 12 scheduled training sessions. Two sessions were missed due to lack of transportation, and one session was missed due to chest pain.
The participant was asked to get an approval from their primary care physician before returning to training. The participant was released and returned to the study. Participant 1 did not opt to continue the study for additional 4 weeks due to transportation issues.

Participant 2 was present for 11 of 12 scheduled training sessions. One training session was missed due to the device being down. Participant 2 required the most assistance mounting and dismounting the EBRGT as he required two assistants. He opted to use the oscillating handles of the EBRGT even though he could only hold onto the left handle due to severe impairment of his right upper extremity. Participant 2 did not opt to continue for 8 weeks due to travel time to the study site. However, he expressed that he enjoyed being able to exercise without the fear of tripping and falling.

Participant 3 was present for 22 of 24 scheduled training sessions. One was missed due to device failure and one was due to a scheduling conflict.

Participant 4 was present for 20 of 24 scheduled training sessions. Two missed training sessions were due to scheduling conflicts and the other two were due to illnesses unrelated to the training. One adverse event due to training was noted with participant 4; anterior knee pain began at the end of the first week of training. Training was continued as normal and ice was applied post exercise along with an elastic wrap just below the inferior border of the patella. Pain resolved by the end of the third week. Participant 4 and his wife noted that he was tripping less when walking at home and in the community. This was also noted by the investigators when walking with the participant to and from the study site to his transportation.
**Distance Walked**

All four participants increased their total distance walked during a single training session over the course of the intervention (Figure 2). If a participant missed that particular visit, the data from the previous visit was used. Modest improvements in total distance were observed during the first two weeks of the intervention with larger improvement generally occurring after the first two weeks. Both participants (3 and 4), that continued the treatment for 8 weeks, seemed to have a plateau in training distance during the last two weeks. Distance walked during training session was calculated from average speed and time data for each session. Participants increased their distance walked by increasing both speed and duration.

**Gait Speed**

All participants had an increase in over ground gait speed over the course of the gait training intervention. Participant 1 saw the smallest increase in over ground gait speed with about a 3% increase in speed. Participant 4 had the greatest increase in over ground gait speed with an 85% increase. While there was an overall increase in speed for participant 3, his greatest increase was in the first two weeks and then his gait speed plateaued. Participant 4 had the greatest increase in gait speed during the second half of
his 8 week intervention. Participant 4 also had the lowest initial gait speed at 53.4 cm/sec, but ended up in a similar range to the other participants.

**FIM-L**

There was no change in the FIM-L score for any of the participants.

**Over ground temporal and spatial gait parameters**

The changes in temporal and spatial gait parameters from pre and post training varied between participants (Table 2). Small, if any, changes were noted for participant 1. Some of the largest changes were noted for participant 4. His cadence increased by almost 18 steps/minute (22%) and his step length increased for both his paretic and non-paretic sides. Participants 3 also had an increase in cadence and step length for both sides, but to a lesser extent than participant 4. Participant 2 did not increase cadence, but did increase step length on both his paretic and non-paretic sides.

All but one participant had a slight increase in single limb support times for their paretic side. The same slight increase was seen for their non-paretic sides also. This led to a slight decrease in all participants double limb support time.
Selected measures of gait symmetry are presented below in Table 3. The largest gait asymmetries were noted for participant 2 and participant 4. Participant 2 had larger temporal asymmetries while participant 4 had larger spatial asymmetries. Participant 4 had improved gait symmetry for both SLR and SwR after the gait training intervention. Participant 2 did not have any improvements in gait symmetry.

Discussion

Feasibility of implementing a gait training intervention using the EBRGT

The main purpose of this study was to determine if it was feasible to improve gait parameters and mobility outcome in ambulatory individuals with hemiparesis due to stroke using the EBRGT. Our results suggest that gait training with the EBRGT may have positive benefits on mobility and gait parameters. All four participants were able to train on the EBRGT with minimal assistance. The main assistance needed was to help the participants mount and dismount the EBRGT. Some simple modifications may make the EBRGT even easier for this group to use. To mount the EBRGT the users had to step up onto a platform using a step. The addition of a handrail may reduce the amount of assistance needed as there was nowhere for the participants to hold on while they were stepping up to mount the device.
Only one adverse event/effect was noted as a result of training. Participant 4 developed anterior knee pain during training that continued between training sessions. The knee pain resolved with some icing and did not prevent the participant from performing normal daily activity or training on the EBRGT.

While there were some positive outcomes of this study, participants were only able to complete very modest amounts of training at any given session. The maximum time spent walking varied between participants from 6.3-15.0 minutes during a single session. If the main goal is mass practice of the gait cycle, it may be necessary to power the flywheel of the elliptical so that less input is required from the user to spin the flywheel. Alternatively or in addition, some body weight could be off-loaded to reduce the amount of energy expended to maintain posture against gravity. Reducing the input required by the user may increase the number of gait cycles the user can perform before fatiguing.

There were no changes in FIM-L due to the intervention. The categories used in the FIM-L do not provide high levels of resolution. In the future, a test with higher resolution would be more appropriate to use in a higher functioning sub-acute and chronic stroke population where modest improvements are expected (Jorgensen et al., 1995).
All participants increased their gait training distance during the course of the intervention. While endurance was not directly measured, this may be an indication of improved endurance due to the intervention. Participants self-reported anecdotal evidence of improved endurance. Participant 4 reported being able to attend football games and other activities out in the community that previously he would not have attended due to tripping and tiring easily. Participant 3 also reported being able to walk longer distances without fatiguing: such as parking in regular parking spots instead of parking in the handicap designated spots. Participant 2, who walks regularly at the mall, reported walking 2 miles instead of his typical 1 mile at the mall.

All four participants exhibited an increase in self-selected over ground gait speed after either 4 or 8 weeks of gait training using the EBRGT. Two of the four approached or exceeded the minimal detectable change (MDC) of 30cm/s for gait speed (Fulk and Echternach, 2008). Participant 4 even progressed from a gait speed that indicated limited community ambulation (<80cm/s) to a gait speed that indicated unlimited community ambulation (>80cm/s) (Perry et al., 1995). It is important to note that participant 4 was sedentary prior to this study and without a control, it is difficult to determine whether the
improvements were due to a change in activity level or due to the specificity of the activity. Participant 3 had a plateau in gait speed close to 120cm/sec. This is not surprising since that speed is within the range of self-selected walking speeds for normal healthy adults (Perry, 1992).

Jackson et al. (2010) performed a case study where ambulatory individuals with stroke trained on an elliptical trainer 2 or 3 times/week for 8 weeks. The participants in this study did not exhibit an increase in self-selected over ground gait speed at the end of the intervention. There are a few reasons that could explain the differences in findings. One explanation for this could be the differences in training parameters. In the current study, no assistance was provided to the participants while training on the EBRGT, whereas in the Jackson et al. (2010) study assistance was provided by an assistant either moving the mechanically linked handles to assist the participant or by offloading some body weight. Assistance was provided in the Jackson et al (2010) study so that the individuals could practice walking for 20 minutes continuously. Another factor may be the level of chronicity of the participants. While both studies involved participants with chronic stroke (> 6 months post stroke), the Jackson et al. study enrolled participants who were 3 years post stroke and greater. The currently study only involved one participant who fell in that same range.

While some studies have demonstrated that the speed at which the participants practice gait influences gait speed as an outcome measure (Sullivan et al., 2002, Pohl et al., 2002), this study did not focus on increasing speed during training sessions. The participants in our study were told to exercise at a perceived exertion of ‘somewhat hard’ or less for as long as possible during a 30 minute window. During training, participants
typically trained at speeds ranging between 0.44m/s and 0.89m/s. It is interesting that all participants improved over ground gait speed and all participants ended up with self-selected over ground speeds that exceeded the 0.89m/s. Therefore, there is a mechanism contributing to improved self-selected over ground gait speed other than training speed. The step length of the gait trainer is fixed at about 36cm, thus in order to achieve a speed of 0.89m/s, the user would have to adopt a cadence of 148 steps/min. A cadence of that magnitude exceeds the self-selected level surface walking cadence of all of the participants in the study. Practicing gait cycles at a higher rate may have contributed to increased over-ground gait speed. The cadence during training was not directly measured and all participants did not train at the maximum speed and thus cadence.

Temporal and spatial gait parameters

Participant 1 had little or no changes in temporal or spatial gait parameters. This was not surprising since she had only mild effects on her gait due to stroke to start with and she only attended 9 of 12 scheduled training sessions. During her training sessions, she only reached maximum of 6.3 mins of training tolerated in a 30 minute window. Participant 1 exhibited a minimal increase in gait speed due solely to increased cadence and not an increase step length. Both paretic and non-paretic step length actually decreased slightly.

All of the other participants (2,3,4) had clinically significant gains in step length of over 5 cm (Patterson et al., 2008b). In addition, double limb support times decreased.
Decreased double support time may be an indicator of improved balance. Base of support is largest when both feet are in contact with the ground, such as in double support. An individual with impaired balance may show an increased double support time as it is more stable relative to single support where only one foot is in contact with the ground and the base of support is much narrower.

Participant 1 and participant 3 did not exhibit significant gait asymmetries (Balasubramanian et al, 2007) while participants 2 and 4 did. While gait asymmetry does not correlate well with gait speed it does correlate with stroke severity and paretic limb function (Balasubramanian et al, 2007). Participant 4 had the largest spatial gait asymmetry in terms of step length. Not only did participant 4 increase his step length bilaterally, but he also improved his step length asymmetry from 1.53 (indicating a longer paretic than non-paretic step length) to 1.25 after the gait training intervention. Improved step length symmetry indicates improved paretic limb function (Balasubramanian et al, 2007); however without including more biomechanical measures, it is unclear what these improvements were.

Participant 2 had a very small step length asymmetry that did not improve with training. He also had a larger swing time and stance time ratio asymmetries that did not improve with training. Participant 2 was reluctant to bear weight on his paretic limb, which is indicated by short stance times on the paretic limb relative to the non-paretic stance times. This did not improve with training and this compensatory strategy was maintained even though gait speed and step lengths were increased with training, which is similar to results of other studies of gait training in individuals in chronic stroke (Patterson et al., 2008b, Patterson et al., 2008, Balasubramanian et al., 2007).
Limitations

Limitations of this study included the absence of a control group and a low number of subjects. Some of the outcome measures may contain bias due to the principle investigators performing the tests. Results are not generalizable due to the small number of participants. Some of the effects of the device were inferred from other measures, it may provide more insight into the effects of training on the device to take direct measures of balance and endurance in futures studies.

Conclusion

It is feasible and safe for ambulatory patients with hemiparesis secondary to stroke to train on the newly developed EBRGT with minimal assistance. Results suggest positive, clinically significant outcomes in measurable gait parameters. Some study participants improved gait speeds to an extent that suggests improvements in mobility. The EBRGT may provide a safe and affordable training option for patients after stroke. Randomized controlled trials comparing the EBRGT to conventional gait training methods are warranted.
Table 1. Participant characteristics.

<table>
<thead>
<tr>
<th>Participant</th>
<th>Gender</th>
<th>Age</th>
<th>Side of Hemiparesis</th>
<th>Height (in)</th>
<th>Weight (lbs)</th>
<th>Months Since Stroke</th>
<th>FIM-L</th>
<th>Anti-spasticity Medication</th>
<th>Assistive devices</th>
<th>Orthosis</th>
<th>exercise</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>F</td>
<td>69</td>
<td>L</td>
<td>64</td>
<td>144</td>
<td>3.5 months</td>
<td>7</td>
<td>No</td>
<td>No</td>
<td>None</td>
<td>no</td>
</tr>
<tr>
<td>2</td>
<td>M</td>
<td>64</td>
<td>R</td>
<td>72</td>
<td>198</td>
<td>45 months</td>
<td>6</td>
<td>Yes</td>
<td>Single point cane</td>
<td>AFO</td>
<td>walks 7 days/week at mall for 1 mile</td>
</tr>
<tr>
<td>3</td>
<td>M</td>
<td>78</td>
<td>L</td>
<td>70</td>
<td>200</td>
<td>3 months</td>
<td>7</td>
<td>No</td>
<td>No</td>
<td>None</td>
<td>stationary bike 20 mins/day</td>
</tr>
<tr>
<td>4</td>
<td>M</td>
<td>58</td>
<td>R</td>
<td>69</td>
<td>168</td>
<td>7 months</td>
<td>6</td>
<td>No</td>
<td>Single point cane</td>
<td>AFO prescribed, but not used</td>
<td>no</td>
</tr>
</tbody>
</table>
Table 2. Selected temporal and spatial gait parameters for each participant before and after intervention. SL=step length, P=paretic, NP=non-paretic.

<table>
<thead>
<tr>
<th>Gait parameter</th>
<th>Participant 1</th>
<th>Participant 2</th>
<th>Participant 3</th>
<th>Participant 4</th>
</tr>
</thead>
<tbody>
<tr>
<td>cadence (steps/min)</td>
<td>92.4</td>
<td>96.5</td>
<td>98.1</td>
<td>97.5</td>
</tr>
<tr>
<td>SL-P (cm)</td>
<td>59.7</td>
<td>58.4</td>
<td>58</td>
<td>67.7</td>
</tr>
<tr>
<td>SL-NP (cm)</td>
<td>60.2</td>
<td>59.6</td>
<td>51</td>
<td>57.2</td>
</tr>
<tr>
<td>Single limb support-P (%GC)</td>
<td>35.2</td>
<td>33.9</td>
<td>25.4</td>
<td>27.2</td>
</tr>
<tr>
<td>Single limb support-NP (%GC)</td>
<td>36.6</td>
<td>37</td>
<td>40.2</td>
<td>41.5</td>
</tr>
<tr>
<td>Double limb support (%GC)</td>
<td>28.6</td>
<td>27.8</td>
<td>34.2</td>
<td>31</td>
</tr>
<tr>
<td>Stance-P (%GC)</td>
<td>63.4</td>
<td>63</td>
<td>60</td>
<td>58.5</td>
</tr>
<tr>
<td>Stance-NP (%GC)</td>
<td>64.8</td>
<td>66.1</td>
<td>74.6</td>
<td>72.8</td>
</tr>
<tr>
<td>Swing-P (%GC)</td>
<td>33.4</td>
<td>37</td>
<td>40.1</td>
<td>41.5</td>
</tr>
<tr>
<td>Swing-NP (%GC)</td>
<td>35.3</td>
<td>34</td>
<td>25.5</td>
<td>27.2</td>
</tr>
</tbody>
</table>
Table 3. Gait symmetry measures for step length, stance time, and swing time. Step length ratio=SLR, Stance time ratio=StR, Swing time ratio=SwR. A ratio of 1 indicates perfect symmetry. A ratio greater than 1 represents the paretic parameter being greater than the non-paretic parameter. A ratio of less than 1 indicates the paretic parameter being less than the non-paretic parameter.

<table>
<thead>
<tr>
<th>Participant #</th>
<th>0 weeks</th>
<th>4 weeks</th>
<th>8 weeks</th>
<th>0 weeks</th>
<th>4 weeks</th>
<th>8 weeks</th>
<th>0 weeks</th>
<th>4 weeks</th>
<th>8 weeks</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.99</td>
<td>0.98</td>
<td>n/a</td>
<td>0.98</td>
<td>0.95</td>
<td>n/a</td>
<td>0.95</td>
<td>1.09</td>
<td>n/a</td>
</tr>
<tr>
<td>2</td>
<td>1.14</td>
<td>1.18</td>
<td>n/a</td>
<td>0.80</td>
<td>0.80</td>
<td>n/a</td>
<td>1.57</td>
<td>1.52</td>
<td>n/a</td>
</tr>
<tr>
<td>3</td>
<td>0.96</td>
<td>0.98</td>
<td>1.00</td>
<td>0.94</td>
<td>0.94</td>
<td>0.92</td>
<td>1.13</td>
<td>1.04</td>
<td>1.16</td>
</tr>
<tr>
<td>4</td>
<td>1.53</td>
<td>1.38</td>
<td>1.25</td>
<td>0.92</td>
<td>0.94</td>
<td>0.92</td>
<td>1.23</td>
<td>1.14</td>
<td>1.15</td>
</tr>
</tbody>
</table>
**Figure 1.** A study participant walking on the EBRGT.
Figure 2. Distance walked by each participant during selected training sessions over the course of the intervention.
Figure 3. Overground gait speed measured pre and post intervention and every two weeks during the intervention.
Works Cited


Pohl M, Merholz J, Ritschel C, Ruckriem S. Speed dependent treadmill training in ambulatory hemiparetic stroke patients: A randomized controlled trial. Stroke. 2002;33;553-558


Tong RK, Ng MF, Li Leonard SL. Effectiveness of gait training using an electromechanical gait trainer, with and without functional electric stimulation, in subacute stroke: A randomized controlled trial. Arch Phys Med Rehabil 2006;87:1298-1304.


Appendix
Appendix A

Functional Independence Measure – Locomotor

(FIM-L)

Independent: Another person is not required for the activity

- 7 – Complete independence: All tasks are safely performed without modification, assistive devices, or aids, and within reasonable time
- 6 – Modified independence: Activity requires any one or more than one of the following: an assistive device, more than reasonable time or with safety considerations.

Dependent: Another person is required for either supervision or physical assistance for the tasks to be performed

Modified Dependence: The subject expends half or more of the effort.

- 5 – Supervision or setup: The subject requires no more help than standby, cuing or coaxing, without physical contact, or needs assistive devices.
- 4 – Minimal contact assistance: With physical contact the subject requires no more help than touching, and the subject expends 75% or more of the effort
- 3 – Moderate assistance: The subject requires more help than touching, or expends half (50%) or more (up to 75%) of the effort

Complete dependence: The subject expends less than 50% of the effort. Maximal or total assistance is required, for the activity. The levels of assistance required are:

- 2 – Maximal assistance: The subject expends less than 50% of the effort, but at least 25%.
- 1 – Total assistance: The subject expends less than 25% of the effort
Chapter 6: Conclusion of dissertation

The purpose of this dissertation research was to determine the feasibility of using a newly developed gait trainer to re-train gait in individuals with hemiparesis resulting from stroke. Robotic gait training may improve gait and thus mobility in individuals with hemiparesis. Unfortunately, existing devices are too expensive for widespread acceptance. The elliptically based robotic gait trainer (EBRGT) was developed as a solution to this problem.

The EBRGT is a low cost solution robotic gait training solution (Chapter 1) that produces gait like movements in both healthy normal adults and adults with hemiparesis resulting from stroke (Chapters 3 and 4). Modifications to the device may improve the gait pattern and make the device more accessible to individuals with hemiparesis. Chapter 5 of this research demonstrated that ambulatory individuals with hemiparesis resulting from stroke may exhibit positive functional gait outcomes as a result of training on the EBRGT.

Future Research

Further research is warranted to determine the efficacy of the EBRGT in individuals with hemiparesis resulting from stroke. A controlled trial is necessary to
determine if the positive outcomes are due to the EBRGT or some other factor not controlled for in the current study.

The current study only included individuals who were ambulatory after stroke. In order to broaden the application of the new device, it should be tested in other populations. Robotic gait devices have been shown to improve gait and mobility outcomes in non-ambulatory individuals with hemiparesis secondary to stroke.

Some modifications to the device were indicated by the current study. A new design for the handles of the EBRGT may help improve joint kinematics at the hip to better simulate level surface walking. Accessibility to the device may also be improved by adding handrails to the steps used to mount the device. This may help reduce the number assistants needed for some individuals from 2 to 1.
Vita

Jessica Cortney Bradford was born on March 10, 1982 in Mecklenburg County, North Carolina and is an American Citizen. She graduated from Lloyd C. Bird High School, Chesterfield, VA in 2000. In 2004, she received her Bachelor of Science from North Carolina State University, Raleigh, NC in Biomedical Engineering and Biological Engineering. While working as a research assistant from 2004-2006 at a small biotechnology company, she completed her Master of Science in Biomedical Engineering at Virginia Commonwealth University, Richmond, VA.