“Though human ingenuity may make various inventions which, by the help of various machines answering the same end, it will never devise any inventions more beautiful, nor more simple, nor more to the purpose than Nature does; because in her inventions nothing is wanting, and nothing is superfluous.”

Leonardo da Vinci

University of Alberta

External Sensors for the Feedback Control of Functional Electrical Stimulation Assisted Walking

by

Lisa Lovse

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Examining Committee

Dr. Richard Stein, Biomedical Engineering

Dr. Arthur Prochazka, Biomedical Engineering

Dr. Vivian Mushahwar, Cell Biology

Dr. Ken Fyfe, Mechanical Engineering

For my mom.

Abstract

Functional electrical stimulation (FES) is a rehabilitative technology that can be used to improve walking in individuals with mobility impairments due to neurologic injury or disease. Feedback is essential for efficient FES-assisted walking. The overall goal of my project was to investigate external sensors to provide feedback for FES-assisted walking. The current study evaluated accelerometers, force sensitive resistors, segment orientation angles, and segment angular velocities to determine which were appropriate for determining the activation and deactivation of six major muscles used for walking. The results demonstrated that the segment orientation angles were the most appropriate sensors. Using the segment angle of the thigh, shank, and foot, the activation and deactivation times of the six muscles investigated could be determined within 6% of the step cycle. The shank segment angle performed the best for determining the activation and deactivation times when only one sensor was desired.

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List of Symbols and Abbreviations

θ_F – foot angle

θ_S – shank angle

θ_T – thigh angle

a - acceleration

AB – able-bodied

Act – activation

ANOVA – analysis of variance

Ar – radial acceleration

ASIS – anterior superior iliac spine

At – tangential acceleration

BF – biceps femoris

CNS – central nervous system

CPG – central pattern generator

DC – direct current

Deact – deactivation

EMG – electromyography

F – fast

f – force

FES – functional electrical stimulation

FSR – force sensitive resistor

g – acceleration of gravity (9.81 m/s^2)

Hz – hertz (cycles/second)

k – spring constant

m – mass

MEMS – microelectromechanical system

MG – medial gastrocnemius

MI – mobility impaired

MS – multiple sclerosis

N – normal

NMES – neuromuscular electrical stimulation

RF – rectus femoris

RGO – reciprocating gait orthosis

RMS – root mean squared

S – slow

SCI – spinal cord injury

Sol – soleus

Std Dev – standard deviation

TA – tibialis anterior

VL – vastus lateralis

VS – very slow

WA – WalkAide

x - displacement

Chapter 1 – Introduction

1.1 Prevalence of Paralysis

Paralysis is the devastating outcome of numerous neurological injuries or diseases. Nearly 1 in 50 people in America live with paralysis; an estimated 6 million people [1]. The leading cause of paralysis is stroke, accounting for 29% of those living with paralysis, followed by spinal cord injury (SCI) at 23%, and multiple sclerosis (MS) at 17%[1]. In Canada stroke is also one of the leading causes of disability along with SCI and MS [2]. There are an estimated 300,000 Canadians living with the effects of stroke, with 50,000 new stroke cases in Canada each year [3]. In 2002, there were an estimated 36,000 Canadians living with SCI with 1,382 new spinal cord injury admissions that year [4]. The MS Society of Canada estimates that there are currently 55,000 to 75,000 patients with MS in Canada [5]. This is one of the highest prevalence rates of MS in the world. Worldwide, MS is one of the most common disabling neurological conditions in young adults [6]. The typical age of onset is between 20 and 50 years of age [6]. Approximately 80% of all spinal cord injuries occur to individuals under the age of 30 years, making it a young adults disease [4]. The paralysis that results from these neurologic disorders can be particularly devastating to the quality of life of those affected due to the age of occurrence as well as the devastating consequences of the disorders.

1.2 Significance of Stroke, MS, and SCI

A stroke is a sudden loss of brain function caused by either a blockage of an artery to the brain or by bleeding in or around the brain caused by a ruptured blood vessel [6]. The effects of stroke depend on the area of the brain that was affected and how much of the brain was affected. These effects can include paralysis of one side of the body, vision difficulties, disorientation, and trouble walking. After a stroke, 25% of those afflicted recover with minor impairment or disability, while 40% are left with moderate to severe impairment [6]. A stroke can affect the ability of an individual to voluntarily activate the muscles needed for walking.

MS is a disabling disease resulting from inflammation and damage of the nerve cells that make up the brain and the spinal cord. Symptoms of this disease include vision difficulties, muscle weakness, loss of balance and coordination, pain, extreme fatigue, and bladder and bowel problems. The primary type of MS is characterized by a steady, slow progression. As the disease progresses, the ability of those afflicted to walk is compromised.

The spinal cord is the relay through which motor and sensory information travels between the body and the brain. Paralysis can result when the spinal cord is damaged by traumatic injury or disease; this is known as Spinal Cord Injury (SCI). SCIs affect the conduction of sensory and motor signals across the site of the lesion, with descending motor commands from supraspinal systems as well as ascending sensory feedback to supraspinal systems being hindered. As a result, a SCI can result in partial or total paralysis of two or all four limbs, as well as loss of sensation, bowel and bladder control, and independent respiration among other things. Similar to stroke and MS, SCIs can result in partial or total loss of voluntary control of the muscles necessary for walking. However, in all these conditions the motor neurons below the level of the lesion remain intact and form a viable connection with the muscle they innervate. As a result, it is possible to use functional electrical stimulation (FES) as an interface with the nervous system to restore function to muscles and generate functional walking. Walking is one of the most desired goals of people with SCI as well as stroke and MS [7]. Restoring the ability to walk in individuals suffering with these neurological disorders can help improve their independence and quality of life.

1.3 Gait

Walking is a cyclical motion often referred to as gait. One cycle of walking is termed the gait cycle. The gait cycle can be divided into two distinct phases: the stance phase and the swing phase as shown in Figure 1.1. Each of these phases can be further subdivided [8]. The stance phase begins with initial contact often called heel contact for walking in a healthy subject. This is the moment when the foot contacts the ground. The loading response or foot flat follows initial contact where weight is rapidly transferred onto the outstretched limb. After loading response, the body progresses over the stance limb and full weight bearing occurs at what is called mid-stance. This proceeds into terminal stance or heel off when the body moves ahead of the limb and weight is transferred onto the forefoot. During pre-swing or toe off there is rapid unloading of the limb as weight is transferred to the contralateral limb. The swing phase involves acceleration of the leg during initial swing where the thigh begins to advance as the foot comes up off the floor. During mid-swing the thigh continues to advance as the knee extends and the foot clears the ground. At terminal swing the limb decelerates as it prepares to contact the ground again.

1.3.1 Major Muscles Used During Gait

Many muscles are used during gait. These muscles can be organized into several groups based on their function and the joint that they control including: ankle dorsiflexors, ankle plantarflexors, knee extensors, knee flexors, hip extensors, and hip flexors. Six major muscles that represent these groups are the ankle dorsiflexor muscle, the tibialis anterior (TA); the ankle plantar flexor muscles, the soleus (Sol) and the medial gastrocnemius (MG); the knee extensor muscle, vastus lateralis (VL); the dual action knee extensor and hip flexor muscle, rectus femoris (RF); and the knee flexor muscle, biceps femoris (BF). The tibialis anterior muscle is located on the anterior or front of the lower leg, just lateral to the crest of the tibia bone. It is used to lift the toes during the swing phase of walking as well as to gently lower the toes to the ground for the foot flat stage of stance. Together, the soleus and gastrocnemius muscles are more commonly referred to as the calf muscles or triceps surae. They are located on the posterior or back of the lower leg. These are powerful muscles, vital in walking as they produce the required push off propulsive force. The calf muscles are active during the stance phase of gait, with their activity peaking around the middle of the gait cycle, just prior to swing [9]. The vastus lateralis and rectus femoris muscles are two of the four quadriceps muscles. The vastus lateralis muscle is the largest part of the quadriceps muscle, located laterally on the anterior of the thigh. This muscle is responsible for extending and stabilizing the knee, with its activity beginning during mid-swing and progressing into the stance phase. Its major peak of activity is during the loading response of gait. The rectus femoris muscle is located along the midline of the anterior thigh. This is a dual action muscle that is involved in extending the knee at the end of swing and beginning of stance, as well as flexing the hip during initial swing to bring the thigh forward. The biceps femoris muscle is also referred to as the lateral hamstrings and is located on the posterior side of the thigh. Its major activity beings after mid-swing and continues into the loading response phase of gait, with its peak of activity around 4% of the gait cycle. The location of these six muscles (TA, Sol, MG, VL, RF, BF) is shown in Figure 1.2.

1.4 Functional Electrical Stimulation (FES)

FES is a rehabilitative technology that uses electrical currents applied to peripheral nerves to restore function to either sensory or motor systems. As a subset of neuromuscular electrical stimulation (NMES), FES uses electrical stimulation to generate functional and purposeful movements for long term management of these movements, as opposed to using electrical stimulation on a short term basis to achieve a reduction in paralytic impairments and increased voluntary functional activities [10]. In the case of surface FES, the stimulating current is applied to electrodes on the skin just above the motor point or to electrodes placed over the nerve innervating the desired muscle. Stimulation at these points provides the greatest amount of motor excitation with the minimal amount of stimulation current. An electrical field is established between two electrodes, causing the ions in the tissue between the electrodes to form a current. This ionic current causes an ionic flow across the nerve, influencing the transmembrane potential. If the potential exceeds a certain threshold, an action potential is generated which causes the contraction of the muscle [11]. It has been suggested that only regeneration of neural tissue could elicit the equivalent function that FES can in paralyzed muscle [12]. A small stimulus of a few milliwatts can generate considerable action, such as a hundred Newton-meter torque in the lower limb. These low levels of current can safely be applied to neural tissue [12].

The first clinical application of FES to improve walking occurred in 1961 when Liberson et al used a single channel stimulator to stimulate the peroneal nerve of 7 hemiplegic patients suffering from “foot drop” [13]. Foot drop is a common symptom of hemiplegia that is characterized by a lack of dorsiflexion during the swing phase of walking. Liberson observed a considerable improvement of gait in all seven of the hemiplegic patients to which peroneal nerve stimulation was applied. Peroneal nerve stimulation causes a contraction of the ankle dorsiflexor muscle, the tibialis anterior muscle, thus lifting the toe. Liberson provided this stimulation in sync with the swing phase of walking using a heel switch. This is widely accepted as the beginning of functional electrical stimulation or “functional electrotherapy” as described by Liberson. Prior to this, electrotherapy had been applied as a treatment series for a certain time to provide a lasting therapeutic effect. The specific uses of electrotherapy prior to Liberson’s study included prevention of muscle atrophy when voluntary contractions were impossible or undesirable, to maintain denervated fibers in a state of vitality until the onset of nerve regeneration, to increase the strength of muscle in cases of muscular imbalance when voluntary contraction is impossible or undesirable, and to reeducate movements by demonstrating to the patient the contraction of a muscle [13]. Around the same time, Liberson was developing a single-channel FES device to correct foot drop in hemiparetic patients, Kantrowitz was applying surface FES to the quadriceps and gluteus maximus muscles to produce standing in a paraplegic patient with a complete SCI. The stimulation produced full extension in the lower extremities, allowing a "swing-through" gait similar to the gait achieved with long-leg braces [14]. Over the past 50 years there have been continuing efforts to develop single channel stimulators to enhance gait in the hemiparetic population as well as an expansion of application of these stimulators to the SCI population and other populations with central nervous system lesions. There is also a strong effort to develop multichannel stimulators to correct more complex gait anomalies in these same subject populations [10]. In many cases, electrical stimulation of one or more muscles can provide a more natural movement than can be achieved using a traditional mechanical orthosis [10].

1.4.1 Benefits of FES

While the primary aim of FES assisted gait is to increase functional upright mobility, there are therapeutic benefits that can result from chronic use as well. FES gait may decrease muscle spasm, increase blood flow to stimulated areas, increase muscle strength in paralyzed muscles, and improve cardiovascular fitness [15]. However, these benefits are dependent on the nature of the injury. The greatest functional improvement with the use of FES gait training has been seen in persons with incomplete SCI. FES gait training in these subjects has improved walking ability, allowing users to walk faster, further, and for longer, as well as allowing them to be more independent in the community as they require less assistance [15]. Subjects with incomplete SCIs could particularly benefit from FES due to their partial preservation of sensation and proprioception [16]. While individuals with incomplete SCI or stroke are the most successful candidates for FES gait enhancement, rudimentary stepping can be successful for short distance ambulation in a select number of patients with complete paraplegia [10].

1.5 FES Systems for Walking

The majority of FES walking devices available today are based on one or both of the technologies introduced by Liberson and Kantrowitz almost 50 years ago [14]. These devices use one of three methods of stimulation: transcutaneous, percutaneous, or implanted [17]. Transcutaneous or surface stimulation uses self adhesive or nonadhesive electrodes placed on the skin above the motor points or major nerves. This method is the least invasive, but the most costly in terms of time to don and doff the electrodes. Percutaneous simulation involves wire electrodes inserted into the muscles close to the motor axons with an epidermal needle. This invasive technique is not useful outside of the research environment, but does allow access to deeper muscles that are inaccessible with surface stimulation. The final method involves implanted electrodes attached to nerves or muscles close to the motor points. These electrodes can also be implanted in the spinal cord.

1.5.1 Categories of FES Walking Devices

There are generally three categories of FES walking devices: single channel FES systems, multichannel FES systems, and hybrid systems.

1.5.1.1 Single Channel FES Systems

The first category of devices involves one or possibly two channels of stimulation and typically uses surface stimulation. These devices are generally targeted to subjects that are able to stand and walk but with difficulty due to weakness in the muscles required for efficient gait. Foot drop stimulators, such as the device first introduced by Liberson, are the most common single channel FES devices. As mentioned previously, foot drop results from a weakness in, or inability to voluntarily activate, the ankle muscles required to dorsiflex the foot during the swing phase of gait. Foot drop stimulators usually stimulate the common peroneal nerve causing contraction of the tibialis anterior muscle, thus dorsiflexing the foot. Stimulation of the common peroneal nerve can also trigger the flexor withdrawal reflex, which is undesirable in this context but as will be discussed later can be useful for multichannel FES systems. For foot drop stimulators, stimulation must be applied at the appropriate time: during the swing phase. Several control methods have been used to decide when to turn the stimulator on and off. The user or a therapist can press a hand switch when stimulation should be turned on; however, this requires constant attention. A heel switch, similar to that used by Liberson, is the most commonly used as it provides more automatic control. When the heel lifts off the ground prior to swing the stimulation is turned on to lift the toe during swing. Once the heel contacts the floor again at the end of swing and beginning of stance, the heel switch turns the stimulation off. There are drawbacks to this method; the user is limited in the footwear and cannot walk barefoot. Also, there must be a connection from the sensor beneath the heel to the stimulator, typically placed below the knee. As a result other control methods have been explored such as the one used in the commercially available WalkAide (Innovative Neurotronics, Austin TX). The WalkAide uses a tilt sensor to determine when to turn stimulation of the common peroneal nerve on and off. The tilt sensor measures the orientation of the leg with respect to the vertical. When the lower leg is tilted back at the end of stance the tilt sensor turns on stimulation and when the leg is tilted forward at the end of swing it turns the stimulation off. The stimulator, tilt sensor, and control electronics are contained in a compact package about the size of a pager and worn on the leg just below the knee. The device is attached with a soft cuff that also contains the electrodes [18]. Some other widely known devices for correcting foot drop include the Fepa, the MikroFES, and the Odstock 2. These devices are all small, fairly reliable, and simple to use.

1.5.1.2 Multi-channel FES Systems

For more complicated gait deficits, multichannel FES systems are used. This second category of FES walking devices use four or more channels of stimulation and are used to generate standing and/or stepping in complete paraplegics who would otherwise be unable to walk. There are generally two strategies for producing a stepping motion. Stimulation can be applied to an afferent nerve, most commonly the common peroneal nerve, to elicit the flexion withdrawal reflex. This stimulation is applied suddenly in order to trigger the flexor withdrawal reflex, which will simultaneously flex the hip and the knee as well as cause ankle dorsiflexion. The common peroneal nerve is also targeted for foot drop stimulators, but in that instance the stimulation is limited to only cause ankle dorsiflexion and avoid triggering the flexor withdrawal reflex. Inducing the flexion withdrawal reflex produces a motion in the stimulated leg similar to the swing phase of gait. Alternatively, the hip and knee flexor muscles, and ankle dorsiflexor muscles can be stimulated individually to produce a swing phase. There are benefits and drawbacks to both techniques. While utilizing the flexion withdrawal reflex is attractive as it only requires one channel of stimulation per leg to produce hip, knee and ankle flexion, this reflex is susceptible to habituation, is highly variable, and may be difficult to elicit in some subjects [11]. Stimulation of the hip and knee flexor muscles, and ankle dorsiflexor muscles individually allows for greater fine-tuning of stimulation responses. However, accessing the hip flexors is difficult if surface stimulation is being used. Successful ambulation has been achieved using both techniques. Surface, percutaneous and implanted stimulation strategies have been explored with these devices. The most commonly known, multichannel device that uses surface stimulation is the Parastep system (Therapeutic Alliances Inc, Fairborn OH). The Parastep system was designed based on the system originally developed by Kralj et al in the 1980s, the Ljubljana FES Walking System. This system used a minimum of 4 channels of surface stimulation to produce a simple reciprocal gait pattern in paralyzed subjects. The gait pattern was split into 3 phases: right swing, double stance, and left swing. Transitions between phases were controlled using 2 pushbuttons attached to the left and right handles of a walking frame, canes, or crutches. When the system was active and no button was being pressed the knee extensors were stimulated to generate stance. A button press would result in stimulation of the ipsilateral peroneal nerve, which would elicit the flexion withdrawal reflex, generating the swing phase for that leg. The subject would remain in swing as long as the button was pressed. This system was applied to 50 complete SCI subjects, 25 of which learned to walk with the system. However the walking speed with this system was slow at 12-18 m/min, and the distance that could be covered was limited to 100-200 m due to fatigue in the quadriceps [19].

Based on the same principles as the Ljubljana FES system, the Parastep system was developed incorporating two additional channels of stimulation to enhance trunk stability. As with the Ljubljana FES system, it uses the flexion withdrawal reflex for the swing phase and stimulation of the knee extensors for stance, as well as pushbutton controllers on a walker that is used for balancing support. The additional channels target the paraspinals or gluteus maximius muscles to improve trunk stability. The Parastep system was the first multichannel FES walking device approved by the FDA and made commercially available. It allows people with traumatic T4-T12 complete or near complete SCI to stand and ambulate short distances. The Parastep system has been evaluated in several multi-centre studies. Over 400 people have used the Parastep system for standing and short distance ambulation in the clinic or at home. However the ambulation performance differed widely and was not predictable. Average walking speeds ranged from 5 m/min to 14.5 m/min, with mean maximal ambulation distances between 118 m to 444 m [20]. In addition to the Ljubljana FES walking system and the Parastep system, there are other surface FES systems that have been developed to allow people with paraplegia to stand and walk. The 8-channel neuroprosthesis WALK! has been used to ascend and descend stairs as well as for standing and walking [21]. Bijak et al have developed the Vienna FES system, that uses 8-channels of stimulation to allow people with paraplegia to walk short distances [22, 23].

A multichannel, FES system was developed by Kobetic and Triolo that used percutaneous and later implanted electrodes with the aim of producing a more natural gait in people with paraplegia [20, 24]. Their approach involved electrodes surgically implanted in the major muscles required for standing and walking with percutaneous intramuscular electrodes inserted in additional muscles to fine tune the stimulation patterns. With 48 channels of stimulation, they were able to produce movements approaching normal gait [24]. However, because 48-channels of stimulation would not be practical outside the research laboratory, they also investigated simpler 8- and 16-channel systems. The 16-channel system offered the best performance, increasing gait speed from 0.1 m/s for an 8-channel system to 0.4 m/s[24]. The muscles stimulated included the hamstrings, gluteus maximus, and the posterior portion of adduction magnus to produce hip extension; the tensor fasciae latae, and either sartorius or iliopsoas for hip flexion; the vastus lateralis/intermedius for knee extension; and the tibialis anterior and peroneous longus for dorsiflexion. For all systems, an assistive device such as a rolling walker was needed to provide balance support to the user, and a 4-button key pad was used by the patient to control transitions between walking phases. There was reluctance by users to wear the FES system with the percutaneous electrodes because of the wires that protruded through the skin, demonstrating the limitations of this system that restricted its use to the laboratory.

There have also been fully implanted FES systems developed. There are a small number of people with implanted 16-channel walking systems as described above [24]. Most of these implants were performed in Cleveland[25]. Some other commonly known multichannel, implanted FES-walking devices include the LARSI system, the FESmate, and the Praxis24 system [17]. The Praxis24 system uses 24 implanted electrodes.

In addition to enabling a swing-through gait pattern, it also allows bladder voiding [14].

1.5.1.3 Hybrid FES Systems

The third category involves hybrid devices that combine FES with mechanical braces. The braces are used to improve stability by reducing the number of degrees of freedom of the body as well as reducing energy consumption by providing support during stance. When using mechanical braces upper body strength is required for standing up and for forward progression during walking. Bracing can be combined with both surface and implanted stimulation. Two common hybrid devices are the HAS system that combines surface FES with active braces, and the RGO-2 or modified reciprocating gait orthosis system that combines surface FES with passive braces. The RGO-2 is an orthosis-based design for restoring standing and limited ambulation in people with complete paraplegia [20, 26]. The mechanical orthosis is a hip-knee-ankle-foot orthosis with a locking mechanism in the knee joint and a coupled cable mechanism that links the 2 hip joints. The coupling mechanism for the hip joints prevents bilateral flexion of the hips during standing and provides a reciprocating motion of the two legs when in motion. The FES component of the system uses surface electrodes over the hamstrings and quadriceps of each thigh. The stimulation sequence causes hip flexion in the ipsilateral leg and hip extension in the contralateral leg for propulsion. In 70 participants at

Louisiana State University, where the device was developed, there was a 75% success ratio for the fitting and training of people with paraplegia to use the orthosis[26]. Users were able to walk at least 180 m on different surfaces with walker support and an average speed of 0.22 m/s. As donning and doffing of the system is cumbersome and time consuming, only 6 people with paraplegia were reported as using the RGO-2 system in daily life tasks [20].

Kobetic and Triolo have recently developed a new hybrid FES walking device that incorporates a novel variable constraint hip mechanism that can either reciprocally couple the hip joints as in the RGO-2, individually lock them, or allow them to move freely [27]. This method allows hip and trunk stabilization when the joints are coupled, and increased hip flexion when the joints are uncoupled during the swing phase of gait to improve step length. The coordination of joint coupling and locking with muscle activation is based on sensor information that is fed to a gait event detector. This system uses 16 channels of stimulation, with 8 implanted intramuscular electrodes per leg. This new hybrid system has been successfully tested on non-disabled volunteers as well as an individual with SCI.

1.5.2 Automatic Control of FES Walking Systems

No matter the device, all forms of FES-assisted gait are slow, awkward, unnatural looking, and energy demanding. The current FES systems for subjects with complete paraplegia are not intended to replace the wheelchair, but instead to allow the user to stand and walk short distances. In an attempt to improve speed and energy consumption, Popovic et al compared automatic and hand-controlled walking in people with paraplegia [28]. In the study, a three-step procedure was used to synthesis the control strategy. The first step involved a simulation of walking using a fully customized model of the potential user to determine the muscle activation pattern needed for a selected walking pattern. In the second step, an artificial neural network was used to generate the rules that would govern the FES system in real time. The neural network was trained using kinematic and dynamic data for the input and the simulated muscle activation patterns as the output. An able-bodied subject trained to walk at different speeds using under-elbow crutches generated the kinematic and dynamic data. The final step transferred the machine-determined rules into a programmable stimulator. For the walking trials in 6 subjects with paraplegia, two 8-channel stimulators were used bilaterally for stimulation of the tibialis anterior and lateral gastrocnemius/soleus muscles to control the ankle joint, the hamstrings and vastus lateralis/medialis muscles to control the knee joint, the gluteus and rectus femoris muscles to control hip extension and flexion, and the gluteus, the tensor fasciae latae, and the adductor longus muscles for control of hip abduction and adduction. The controller used four goniometers attached to body segments to measure joint angles, 2 dual-axis accelerometers attached to the hip joints and 4 pressure sensitive resistors at the heel and metatarsal zone to provide automatic control. The lowest energy cost was for near-ballistic walking, but this was still approximately two times the value of the energy cost for an able bodied population. This was however a significant improvement over the energy cost for hand-controlled walking, which was approximately five times the value for an able bodied population. Even though the near-ballistic walking was most efficient, it was not preferred by any of the users. Five of the six users selected the automatically controlled, slow walking speed as their preferred mode because they found it difficult to synchronize their trunk and arms with the externally controlled leg movements for the faster speeds. This study demonstrates that automatic control using closed-loop feedback can improve the speed and efficiency of FES gait over the pushbutton control that is most commonly used. However, systems that are more acceptable for subjects still need to be developed.

1.6 Walking Control in Normal Physiological System

In the normal physiological system, stepping is controlled by descending drive from supraspinal systems as well as central pattern generators (CPGs) located within the spinal cord. These systems receive input from various sensory receptors within the body. While it has been shown that the spinal locomotor generator, or CPG, can function even when all sensory feedback has been abolished, in the absence of sensory feedback any obstacle or incline is not corrected for and can lead to instability in the gait pattern and potentially a fall. The Central Nervous System (CNS) can be viewed as a complex hierarchically structured controller, with muscles acting as the motors of the human body moving the skeleton and any external loads [29-32]. The CNS controller connects to the sensors and muscular actuators via peripheral nerves. During normal locomotion, one or more of the sensory systems in the human body detects biomechanical events. These sensory inputs are interpreted by the CNS controller and cause the CPG rhythm to be modulated. At a low level, these inputs provide feedback to spinal reflexes, while at a higher level the inputs provide the individual with information about the relative position and orientation of their body segments with respect to each other, the position of their body in the environment, the force exerted on the environment, and obstacles in the environment.

1.6.1 Sensors Used in Normal Physiological Control of Walking

The many sensory systems in the human body provide information about the environment as well as internal and external feedback to the CNS for motor control. Two important classes of sensors in the human body necessary for motor control are the exteroceptive sensors and the proprioceptive sensors [29, 30]. Exteroceptive sensors are the human senses such as vision and the tactile senses. This class of sensors includes the cutaneous skin receptors. Proprioceptors sense movement and internal forces in the body. There are several proprioceptors in the body including muscle spindles that sense the length and shortening velocity in muscles, tendon organs that sense the tension in tendons, joint sensors that provide information about joint angles, and the vestibular system that detects inertial sensory information due to movement of the head. Afferent feedback from muscle proprioceptors and cutaneous receptors is used to continuously shape and regulate the activity of the CPG and motor neurons during locomotion.

1.6.2 Neural Mechanisms for Control of Walking In The Normal Physiological System

Considerable progress in understanding the neural mechanisms that regulate walking in mammals has been made in the past few decades. Computer simulations are an important tool to determine the relative importance of certain proposed mechanisms as they allow the isolation of individual mechanisms in a functional context. This cannot be done using direct experimental approaches as individual neuronal mechanisms or sets of mechanisms cannot be isolated in behaving animals. Computer simulations of stepping of the cat hind limb have been used to assess the relative importance of sensory feedback and the CPG. In their planar model, Yakovenko et al demonstrated that stretch reflexes only seem to contribute significantly when central activation levels are low [33]. Ekeberg and Pearson developed a three-dimensional cat hind limb model to study the relative importance of two sensory signals involved in the termination of stance: unloading of leg, and hip extension [34]. The conclusion from their simulation suggests that the coordination of stepping in the hind legs depends critically on load-sensitive signals from each leg. Taking both of these modeling studies into account, there seems to be an importance for both an open loop pattern generation component as well as a closed loop sensory feedback component to control locomotion. Guevremont et al have demonstrated this when developing a controller for generating over ground locomotion in a cat model [35]. In this study three controllers were implemented: an open loop rhythmic controller, a closed loop feedback based controller, and a combined controller that united aspects of the two previous controllers. The conclusion of the study was that the combined controller was the best solution for restoring robust over ground locomotion after SCI.

1.7 Sensory Feedback for FES Walking Systems

Many of the FES devices presented could benefit greatly from the incorporation of sensors to provide feedback for the modulation of stimulation intensity, as well as for determining the appropriate time for turning stimulation on and off. While there have been some attempts to incorporate sensors to help control FES systems, many systems in use still rely on push-button control which requires a conscious effort from the user to control standing and walking. In general, there are two objectives for using sensory signals in artificial human motor control [30]: for exchange of information with the user to the FES system, and to provide feedback to the artificial control system. In the first instance, sensory signals can be used to determine the intention of the user or to provide sensory feedback to the user through such approaches as auditory signals. In the second instance, sensory signals are used for the coordination of multiple body segments during movement or to control muscles individually.

Sensors are essential for feedback. There are two categories of sensors that can be used to provide feedback to control artificial walking systems: natural sensors and artificial sensors.

1.7.1 Natural Sensors

Natural sensors are the sensory neurons already contained within the body such as the proprioceptors and cutaneous skin receptors. Nerve cuff electrodes implanted around the nerve to record the desired sensory signal, have been developed and tested in animals and demonstrated to be feasible in humans for the control of peroneal nerve stimulation [36, 37].

This method is attractive if a completely implanted system is desired; however, because of its invasiveness it may not be applicable to the larger population.

Electromyography (EMG) has also been explored as a non-invasive way to provide feedback for the control of FES [20, 38, 39]. An electromyogram is the electrical signal associated with the contraction of a muscle [40]. Graupe and Kohn investigated EMG while developing the Parastep FES system as a means of replacing the pushbuttons to turn stimulation on and off, as well as to determine if stimulation levels needed to be increased. EMG signals were recorded from above the level of the spinal lesion of Parastep users to control the activation of stimulation. This was done by using a computer to discriminate between EMG patterns in the upper trunk muscles during standing up and the initiation of stepping. Following a training period of 18 months, the control paradigm was found to make decision errors less than once every 75-100 steps [41]. Graupe et al also recorded response-EMG from the stimulated muscles below the level of the spinal lesion to determine the effectiveness of the produced contraction, and adjusted stimulation levels accordingly. Using this technique they were able to prolong standing in four users with paraplegia, using FES with response-EMG control, by a factor of 3-10 as compared to standing without response-EMG control [41]. The correct positioning of the EMG electrodes was essential for the success of these techniques, and as a result, Graupe and Kohn did not include these sensors in the final version of the Parastep system as they found the donning to be too much of a burden for the users. Recently, Triolo and Kobetic have begun investigating the use of EMG from partially paralyzed muscles in people with incomplete SCI to detect gait events and trigger FES-assisted stepping [39]. The gait initiation that is produced using EMG triggered FES was more similar to able-bodied gait initiation than either switch triggered or open-loop patterned FES; however, the issues associated with the donning of EMG electrodes remain to be addressed.

1.7.2 Artificial Sensors

Artificial sensors are man-made devices worn externally on the body to provide various types of information. Common artificial sensors include goniometers, foot contact switches, force sensitive resistors, accelerometers, and gyroscopes.

1.7.2.1 Goniometer

A goniometer is a device that measures the relative angle of a joint connecting two body segments [40, 42]. One arm is attached to one limb segment and a second arm is attached to the adjacent limb, with the axis of the goniometer aligned with the joint axis. Goniometers traditionally use a resistance potentiometer to convert changes in rotation to a voltage that is proportional to the angle between the arms. Devices that use strain gages are also available. Goniometers are attractive sensors to provide feedback for walking because they output joint angle and are inexpensive. However, they have not been utilized much outside of the research laboratory because they are fragile, can encumber normal movement during walking, take an excessive amount of time to fit and align, and often slip once in place [40, 43].

1.7.2.2 Foot Contact Switch and Force Sensitive Resistor

When Liberson developed his peroneal nerve stimulator to correct foot drop, he also incorporated an artificial sensor to turn the stimulation on and off [13]. Liberson used one of the most popular artificial sensors: the foot switch. In Liberson's device, the foot contact switch was placed beneath the heel. When the heel was lifted, the switch was opened and stimulation was turned on. When the leg completed the swing phase and contacted the ground again, the switch was closed and stimulation was turned off. This simple method worked well in real time and has been replicated numerous times in various FES systems. In recent years, force sensitive resistors (FSRs) have replaced foot contact switches. FSRs are resistors that change their resistance in proportion to an applied load. As such, FSRs can be integrated into shoe inserts or taped to the sole of the shoe to determine if a limb is loaded. Skelly et al used FSRs while developing their cycle-to-cycle based FES system [44]. A cycle-to-cycle control strategy alters muscle stimulation of future gait cycles based on the quality of the previous gait cycle. As a result, the sensory signals do not need to be processed instantaneously but can be up to one gait cycle behind. With this allowance, Skelly was able to use machine learning methods and two FSRs per leg to detect 86% of gait event times correctly as compared to video event time estimates made by human observers. This showed that FSRs may be able to detect more events in stance than just the beginning and the end of the phase.

Foot switches and FSRs are widely used because they are thin and inexpensive. However foot switches or FSRs and the wire that connects the sensor to the stimulator are susceptible to mechanical failure. Foot switches and FSRs have also been reported as having poor reliability because they are not able to distinguish between weight shifting and stepping [45]. Another major problem with foot switches and FSRs is that they do not provide any information during the swing phase of walking, as the limb is unloaded. Because of this, the sensor would not be useful for controlling muscles that are active during the swing phase.

1.7.2.3 Accelerometer

Accelerometers are devices that measure acceleration. With advances in integrated microelectromechanical systems (MEMS), the size and cost of the accelerometer device has been greatly reduced while ensuring the fabrication of the device is maintained at a high quality and reliability [46].

As a result, there has been increasing interest in using accelerometers to detect gait events as well as to control FES systems. Accelerometers are attractive sensors to use for gait studies because they are low cost, small in size, low power, and have a dynamic range and sensitivity [43, 47]. Accelerometers are essentially force transducers designed to measure the reaction forces associated with a given acceleration [40]. The basic mechanism of an accelerometer is a mass-spring system governed by Hooke's law ($f = kx$, where f is the force applied, k is the spring constant characteristic of the material, and x is the displacement the force creates in the spring), and Newton's second law ($f=ma$, where again f is the force applied, m is the mass, and a is the acceleration of the mass) [47]. When an acceleration is applied to the device, the reaction of the mass will either compress or stretch the spring. The restoring force produced by the spring is equivalent to the force required to accelerate the mass: $f = kx=ma$. Knowing the spring constant, mass of the element, and displacement of the spring, the acceleration can be determined by rearranging the equation: $a = kx/m$. This is the basic principle on which all accelerometers operate.

There are several classes of accelerometers: fluid, reluctive, servo, and magnetic [47]. However, the classes that are more common for use in detecting and controlling human movement include: the strain gauge, the piezoresitive, the capacitive, and the piezoelectric accelerometer. Strain gauge type accelerometers consist of strain gauges bonded to a cantilever beam with a mass at one end [42]. When the base of the cantilever is accelerated, the cantilever beam is deflected due to the inertia of the mass. This causes a strain in the wires of the strain gauge, changing their resistance proportionally to the value of the acceleration. Piezoresistve accelerometers are manufactured from surface micromachined polysilicon [46]. The polysilicon springs are arranged in a Wheatstone configuration. Thus, changes in the electrical resistances of the spring when a force due to acceleration is applied are proportional to the resulting voltage. Capacitive accelerometers use a silicon mass element surrounded by an array of paired capacitors [47]. As the mass reacts to an acceleration, an imbalance is created between opposing capacitors, producing an electrical signal proportional to the applied acceleration. Piezoelectric accelerometers use a piezoelectric element and a seismic mass [46]. The seismic mass causes the piezoelectric element to bend when an acceleration is applied, allowing charge to build up on one side of the accelerometer. This is recorded as a voltage signal proportional to the applied acceleration.

The signal produced by an accelerometer worn on the body is dependent on four main factors: the position where it is placed, the orientation in which it is placed, the posture of the subject wearing the accelerometer, and the activity being performed by the subject [46]. The maximum range of accelerations associated with normal walking is usually found close to the foot and range from around ± 2-5g [47]. The acceleration signals produced by an accelerometer placed close to the foot will differ distinctly from the signals produced by an accelerometer placed on the trunk. As a result, the location of the accelerometer is an important consideration when developing a system to detect gait events or provide feedback to FES walking system. Orientation is another important consideration as typically DC-coupled accelerometers are used in gait studies. DC-coupled accelerometers are sensitive to static acceleration due to gravity as well as inertial components due to movement. These components are combined in the output signal from the accelerometer [47].

One of the first studies to use accelerometry to investigate walking was performed by Liberson in the 1930s. Liberson discovered critical information for normal and pathological gait could be revealed using basic acceleration patterns of the body [47-49]. Then in 1973, Morris demonstrated that accelerometers could be used to provide sufficient information to define the movement of a segment of the body [50]. Morris used five accelerometers mounted on a Perspex platform, attached to the anterior-medial surface of the tibia to determine the angular velocity, direction cosine, translational acceleration, velocity, and position of the shank. In recent years, the use of accelerometers for gait event detection, and FES control has increased mainly due to the improvements in measurement accuracy and reduction in size of the accelerometer. Willemsen et al studied the possibility of using accelerometers as swing phase detectors to replace foot switches for peroneal nerve stimulation

[51]. Using four uni-axial accelerometers attached to the shank and a linear transform, they were able to calculate the equivalent acceleration of the ankle joint. They used the equivalent acceleration and an automatic detection algorithm based on cross correlation calculations to determine when the transition between stance and swing occurred in four able-bodied subjects as well as four hemiplegic subjects. They were also able to achieve similar results using a single accelerometer measuring the radial acceleration of the shank just below the knee in place of the equivalent acceleration. Willemsen's technique worked for the majority of steps analyzed; however, there were errors that occurred. These errors included the controller becoming out of phase, a step being missed completely, a step being detected too early, a heel strike being missed, and a heel strike being detected too early.

Dai et al also investigated the use of an accelerometer to replace foot switches for peroneal nerve stimulation [52]. They utilized an accelerometer attached to the shank as a tilt sensor to determine the orientation of the shank during gait instead of detecting specific gait events. With the orientation of the shank, a threshold detection method could be used to determine when stimulation needed to be turned on. To use an accelerometer as a tilt sensor, Dai low pass filtered the signal to reduce the non-gravitational components. The resulting signal was proportional to the orientation of the sensor with respect to gravity. This sensor was incorporated into the WalkAide to control stimulation of the peroneal nerve to correct foot drop [18].

Locations other than the shank have also been explored as attachment locations for an accelerometer to detect gait events. In addition to a dual-axis accelerometer placed on the shank, Hanlon and Anderson also placed a dual-axis accelerometer on the foot[53]. Both accelerometers where compared to a FSR to see which could accurately detect initial contact. A force plate was used as the criterion measure for initial contact. While all three sensors were found to accurately detect initial contact for slow, normal, and altered walking in real time, the FSR system showed significantly lower error than either accelerometer location. Interestingly, Mansfield and Lyons reported a single-axis accelerometer placed on the lower, dorsal trunk was able to detect initial contact in 4 able-bodied subjects just as, if not more, reliably, than a foot switch [54]. Zijlstra used a tri-axial accelerometer on the dorsal trunk to detect steps and determine an approximation of step length and walking speed [55].

Accelerometers in combination with FSRs have been considered to provide sensory feedback to control FES walking in a hemiplegic individual [56]. Dosen and Popovic used the signals from accelerometers and FSRs in a controller based on machine learning to determine the activation pattern for six muscles. For this simulation kinematic data were collected from six able-bodied subjects using four dual-axis accelerometers and four FSRs placed on the right leg. Using these data, the optimal controller was able to generate the target, healthy-like gait in 90% of all tested gait trials.

1.7.2.4 Gyroscope

Gyroscopes are another type of inertial sensor that has been investigated to provide feedback in gait studies because of their small size, and low power consumption. Rate gyroscopes measure angular velocity. Gyroscopes operate by measuring the Coriolis acceleration that is generated when a rotational angular velocity is applied to an oscillating piezoelectric bimorph [57]. A piezoelectric bimorph is a cantilever consisting of two active layers: a piezoelectric layer and a metal layer. Electrical activation of the piezoelectric layer causes a displacement in the cantilever when one-layer contracts and the other expands. Unlike accelerometers, gyroscopes measure motion without reference to gravity. As well, because two points in the same plane on a rigid body experience the same angular velocity, the signal from a gyroscope is not dependent on its position along a rigid body.

Gyroscopes are attractive sensors for gait event detection and FES control because the integral of their signal gives the angle of the segment to which it is attached. However the gyroscope signal does contain a zero frequency or DC-component that can change the angular measurements even when the gyroscope is not moving. Because of this DC-component, taking the integral of the signal results in a large drift of the signal that increases as time progresses. In order to use the integral of a gyroscope signal to determine segment angle, this drift must be compensated for. Tong and Granat investigated two approaches to solve the problem of drift in order to use gyroscopes for gait event detection [58]. The first approach used FSRs placed beneath the heel and toe of the foot to detect mid-stance. When mid-stance was detected, the shank and thigh were assumed to be vertical and the integrated signal was set to zero for each gait cycle. In the second approach, the signal was high-pass filtered with a 0.3 Hz cut-off frequency to remove the drift. Both approaches were able to remove the drift during walking in a straight path; however, the integrated signals from the gyroscope still contained drift during and after a turn occurred during walking. Cikajlo et al investigated another approach to remove the drift from the angle calculated from a gyroscope signal. Their approach used a Kalman filter and a dual-axis accelerometer to correct the drift in the angle signal [59]. Both the gyroscope and accelerometer were attached to the front of the shank. The integral of the gyroscope signal was used as the primary estimate of shank segment angle. The accelerometer was used as an inclinometer to also estimate the shank segment angle, with the Kalman filter used to estimate and correct the error between the two estimates of shank segment angle. Using this approach they were able to eliminate any drift introduced by integrating the gyroscope signal for straight walking paths. However the performance of the approach for walking with turns was not investigated.

Despite the problems with drift, studies have reported successful use of gyroscope signals to control stimulation during walking and, when combined with FSRs or accelerometers, to accurately detect gait events [45, 57, 60]. Monaghan et al were able to use a uni-axial gyroscope to reliably determine in real time when stimulation of the triceps surae muscle should occur. Their technique involved utilizing the angular velocity signal from the gyroscope to decide when to begin integrating the signal to calculate the angle[60]. The algorithm looked for a zero crossing that went from a positive to a negative value in the angular velocity signal. When this was detected, integration of the angular velocity to determine the angle was initiated. Once a preset angle threshold was reached by this integrated signal, stimulation was triggered and the algorithm terminated the integration of the angular velocity signal. This effectively reset the integration every step, preventing drift. While this technique did work reliably, there were some drawbacks. Because of the zero crossing required to trigger the algorithm, the first stimulation time was always missed. As well, the final step taken sometimes initiated an undesired stimulation burst even when the user wanted to stand still. Finally, because the deactivation time was preset based on time that had elapsed since activation, the technique was not able to adapt to different walking speeds, when the stimulation time may need to be longer or shorter.

Accelerometers and gyroscopes have been placed on various locations along the lower extremity to detect gait events or control stimulation of muscles for FES assisted walking. These locations have included the trunk [54, 55], the thigh [51, 57], the shank [50-52, 57], and the foot [45, 53, 57]. However, no study has evaluated what sensor and what sensor location is most appropriate to control the stimulation of the major muscles required for locomotion. Kotiadis et al recently investigated what the most appropriate sensor worn on the shank was for controlling stimulation of the peroneal nerve to correct foot drop. They found that algorithms that used gyroscope data produced the best results [61]. These results are limited to one sensor location and were only tested in one subject.

1.8 Summary

FES devices have been developed to improve or restore walking in individuals with various neurological disorders. However, most of these devices have not been readily accepted by users. One of the biggest factors limiting their acceptance is a lack of successful feedback control. In order to develop a successful FES walking system, sensors that can provide feedback should be included. Accelerometers, gyroscopes, and FSRs have been examined as potential sensors to be used to provide feedback in FES devices. Studies have shown that these sensors are able to detect gait events and control the stimulation of the peroneal nerve to correct foot drop. However, no study has compared all three of these sensors and their different attachment locations to determine which is the most appropriate to be used in an FES walking system and where they should be attached.

1.9 Overview of Masters Work

The focus of my thesis work was to determine what the most appropriate sensors and sensor locations are to control the activation and deactivation of six major muscles in each leg required for walking. While many sensors have been shown to be capable of determining different gait events and, in some cases, to control FES, no study has compared these sensors to each other to determine which are the most appropriate. The goal of my work was to identify the best set of sensors that can be used to control FES in order to produce a natural looking functional gait in individuals with spinal cord injury or other neurological disorders. As the control is required to work in real time, a simple threshold crossing detection technique was used. Using this technique limits the computational load required by a controller for a FES walking system. The sensors tested included accelerometers and FSRs, as well as angle and angular velocity signals calculated from kinematic data of the lower leg. The accelerometers were placed on four locations along the lower extremity: the trunk, the thigh, the shank, and the foot. The two FSRs were placed in the shoe insole, one beneath the heel and the other beneath the toe. The angle and angular velocity were determined for the thigh, shank and foot segments. The angle and angular velocities were considered as signals that could possibly be generated by a gyroscope placed on the segments.

Chapter 2 of this thesis describes the experiments I conducted to determine which sensors were most appropriate for control of stimulation, the results I obtained and their significance.

Chapter 3 provides a general conclusion of my work as well as future directions for incorporating the identified sensors to provide feedback for the control of FES walking devices.

1.10 Figures

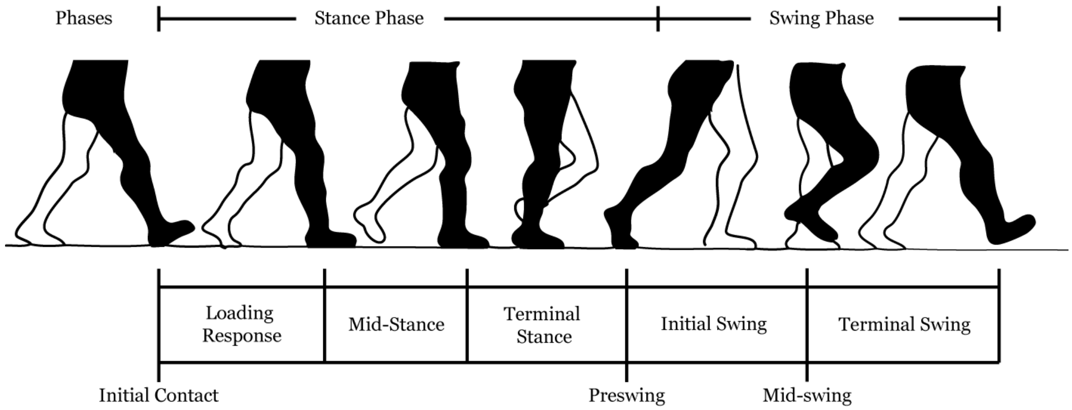

Figure 1.1 – The Gait Cycle

The gait cycle can be divided into two distinct phases: the stance phase and the swing phase. These phases can be further subdivided into initial contact, loading response, mid-stance, terminal stance, pre-swing, initial swing, mid-swing, and terminal swing.

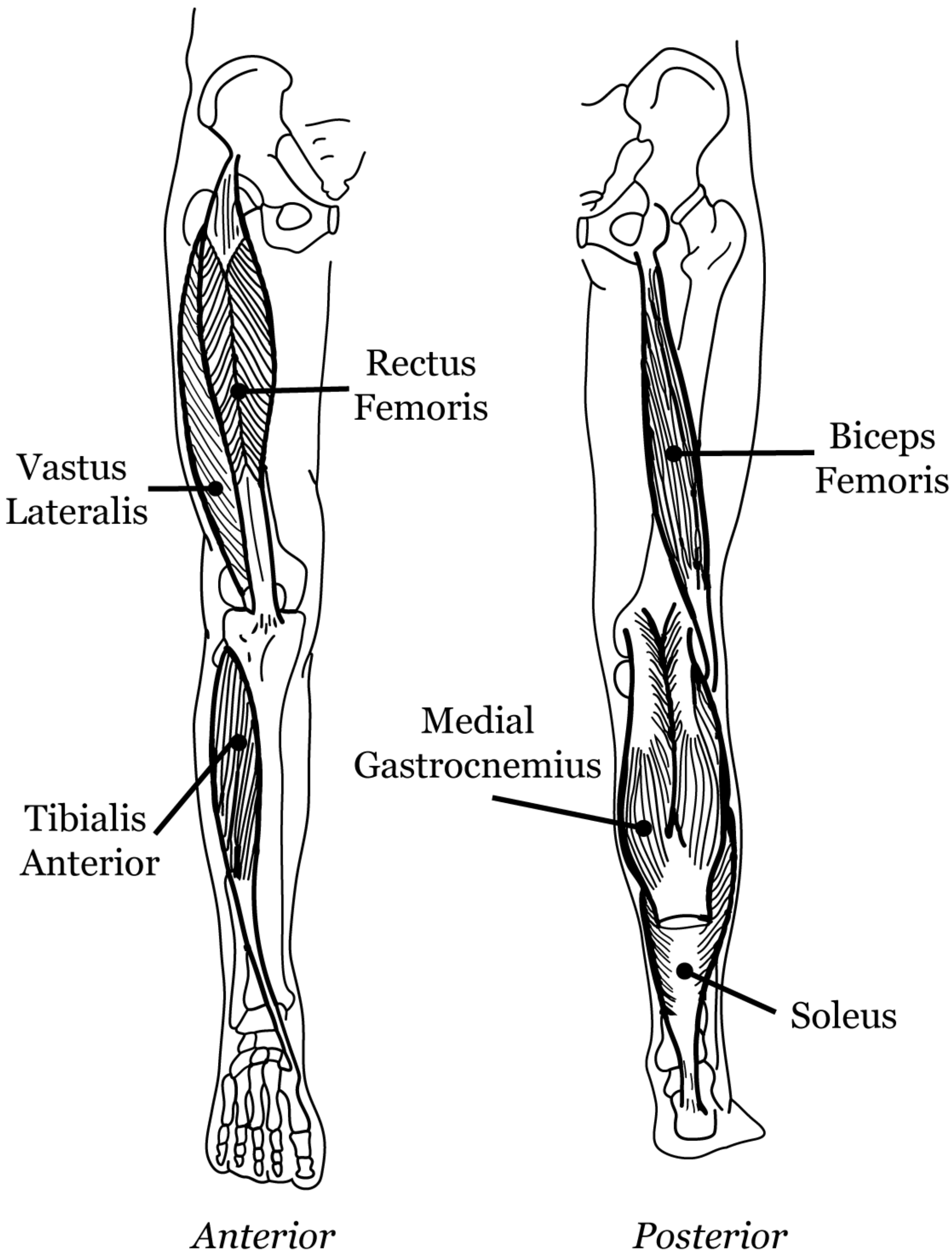

Figure 1.2 – Major Muscles Used for Walking

Six of the major muscles used for walking are the tibialis anterior, soleus, and medial gastrocnemius muscles of the shank, and the vastus lateralis, rectus femoris, and biceps femoris muscles of the thigh.

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Chapter 2 –External Sensors For Determining the Activation and Deactivation Times of Six Major Muscles Used in Walking

2.1 Introduction

Several neurological disorders including spinal cord injury (SCI), stroke, and multiple sclerosis (MS) compromise the ability of afflicted individuals to walk. Many of these disorders damage the central nervous system (CNS) while leaving the peripheral nervous system intact. As a result, it is possible to use functional electrical stimulation (FES) as an interface with the nervous system to restore function to muscles and generate functional walking.

FES is a rehabilitative technology that uses electrical currents applied to peripheral nerves to generate functional and purposeful movements. In 1961 Liberson developed the first FES system to improve walking in individuals who suffered from foot drop due to stroke [1]. Foot drop is a common symptom of stroke, MS, and SCI that is characterized by a lack of dorsiflexion during the swing phase of walking. Since Liberson's study, several FES systems have been developed to generate or improve walking [2-8]. The multi-channel versions of these systems, that provide stimulation to several muscles, have gained limited success with users because of the poor efficiency of the walking they produce and the amount of concentration typically required by the user to prevent falls. One way to improve efficiency of FES systems and limit the amount of required user input is to incorporate automatic feedback control [8]. Simpler, 1- or 2-channel FES systems that are used to correct foot drop, have incorporated feedback control. One such system, the WalkAide (Innovative Neurotronics, Austin TX), uses the orientation of the shank as feedback to control stimulation. This FES system has gained acceptance with users in part because of its ease of use and the decrease in effort required for walking when using the device[3].

In the normal physiological system, stepping is controlled by descending drive from the brain, central pattern generators located in the spinal cord, and input from various sensory receptors within the body. The many sensory systems in the human body provide information about the internal and external environments to the CNS and play a critical role in motor control.

Many of the FES systems for walking could benefit greatly from the incorporation of sensors to provide feedback for determining the appropriate timing of stimulation [2, 9]. While there have been some attempts to incorporate sensors to help control FES systems[10-21], many systems in use still rely on push-button control which requires a conscious effort from the user to control standing and walking[4].

Sensors are essential for feedback. There are two categories of sensors that can be used to provide feedback to control FES walking systems: natural sensors and artificial sensors. Natural sensors are sensory neurons already contained within the body. Using natural sensors to control FES requires significant computational load and can require invasive techniques, which may limit the applicability and acceptance of the system [4, 21, 22]. Artificial sensors are man-made devices worn externally on the body to provide various types of information. Many artificial sensors are low cost, low power, and small in size, making them attractive for providing feedback in FES walking systems. Foot switches such as force sensitive resistors (FSRs) [1, 16], accelerometers [13, 23, 24], and gyroscopes [20, 25-27] have been successfully used to provide feedback during walking.

FSRs are resistors that can be incorporated into a shoe insole. The resistive properties of the FSR change with varying loads producing a voltage signal proportional to the applied load. FSRs are one of the most widely used artificial sensors in FES walking systems.

Accelerometers are artificial inertial sensors that measure acceleration. Because of the low cost, miniature size, and reliability of these sensors, there has been increasing interest in using accelerometers to provide feedback in FES devices [13, 18].

Gyroscopes are another type of artificial inertial sensor that has been investigated to provide feedback in gait studies because of their small size and low power consumption [20, 26, 27]. Gyroscopes measure angular velocity. Their signal can be integrated to determine the orientation of the sensor. If this sensor is attached to a body segment, the integrated signal can be used to determine the segment orientation angle.

While many studies have investigated the applicability of these various sensors individually to provide feedback during gait, no study has investigated which of these sensors is the most appropriate sensor to provide feedback for FES walking systems controlling numerous muscles. A recent study by Kotiadis et al compared accelerometers and gyroscopes to determine the most appropriate sensor worn on the shank for controlling the stimulation of the peroneal nerve to correct foot drop. They found that control algorithms that used gyroscope data produced the best results [28]. Nonetheless, these results are limited to one sensor location and were only tested in one subject.

While many sensors have been shown to be capable of determining different gait events and, in some cases, to control FES, no study has compared these sensors to each other to determine which are the most appropriate. The aim of this study was to determine the most appropriate set of sensors for controlling the activation and deactivation of six major muscles used during walking. The signals produced by artificial sensors are dependent on the location of where the sensor is attached. As such, this study also investigated various sensor locations to determine the most appropriate attachment location. The goal was to identify the best set of sensor signals that can be used to control the activation and deactivation of several muscles using FES, in order to improve walking in individuals with neurological disorders such as SCI, stroke, or MS.

2.2 Methods

2.2.1 Requirements for a good sensor

There are several requirements that must be met in order to consider a sensor signal as appropriate for providing feedback to control the stimulation of one or all of the 6 muscles. The sensor must provide a signal that can detect when to turn stimulation on and off accurately and reliably. This detection should be possible using a simple threshold crossing technique to limit the computational load required. This sensor signal must work both for the altered walking pattern that results due to mobility impairments, as well as for a more normalized walking pattern, similar to able bodied walking, that would result when stimulation is applied to improve walking. The sensor must work well over a range of speeds to prevent restricting the speed of walking produced by an FES system incorporating this sensor. In order to find which of the commonly used sensors for gait event detection meet these requirements, several sensor signals from different attachment locations were tested for walking in both able-bodied (AB) and mobility-impaired (MI) individuals.

2.2.2 Participants

Data were acquired from 5 AB individuals (3 male, 2 female) and 5 MI individuals (4 male, 1 female). All of the MI subjects had chronic neurological disorders that affected their walking: 2 subjects had incomplete SCI, 2 had suffered a stroke, and 1 had a peripheral nerve injury caused by radiation. Typically FES is only applicable to individuals that suffer central nervous system injuries or diseases. However, in this instance the peripheral nerve injury was caused by a focal demyelination caused by radiation. As a result, the nerves below the level of the injury remained intact and were able to be stimulated using FES.

2.2.3 Walking Conditions

All subjects were asked to walk a 6 meter, straight path in view of 8 Vicon motion capture cameras (Oxford Metrics Group, Oxford England). After calibration, the typical accuracy of the motion capture system was less than 1 mm. The right leg of the AB subjects and the more affected leg of the MI subjects were instrumented. Markers were placed on the skin or clothing above the bony prominences of the lower extremity including: the greater trochanter of the hip joint, the lateral epicondyle of the knee joint, the lateral malleolus of the ankle joint, the calcaneal tuberosity of the heel, and the top of the foot over the 2nd metatarsal joint. The AB subjects were asked to walk at 4 different cadences to the beat of a metronome: very slow (40 beats/min), and slow (60 beats/min) cadences to mimic the walking of individuals with mobility impairments; as well as at normal (88 beats/min) and fast (120 beats/min) cadences. The MI subjects were only asked to walk at one, self-selected, comfortable speed. However, as all mobility impaired subjects were users of the WalkAide (Innovative Neurotronics, Austin TX), they were asked to walk both with and without the stimulation system used to correct foot drop. Several walking trials were recorded for all subjects with the various conditions.

2.2.4 Muscle Activity

Electromyography (EMG) was used to detect the muscle activity of 6 major muscles during walking. The activity of three lower leg muscles was measured. The tibialis anterior (TA) muscle was selected to represent ankle dorsiflexion. The soleus (Sol) and the medial gastrocnemius (MG) muscles were selected to represent ankle plantar flexion. In addition, the MG also acts as a knee flexor. The activity of three upper leg muscles was also measured. The vastus lateralis (VL) muscle was selected as a knee extensor along with the rectus femoris (RF) muscle, which is a hip flexor, as well as a knee extensor. The biceps femoris (BF) muscle represents knee flexion and hip extension. EMG electrode placement followed established techniques[29, 30]. Figure 2.2 is a schematic showing the locations of the electrodes for the six muscles.

2.2.5 Sensors

Because of the limited space on the lower extremity, and to prevent the experimental setup from altering the walking pattern, only accelerometers and force sensitive resistors (FSR) were used. In addition, segment angle and angular velocity were calculated from the motion capture data for each segment in an attempt to replicate the signal of a gyroscope attached to each segment.

2.2.5.1 Accelerometers

Tri-axial accelerometers (ADXL 320, Analog Devices, Norwood MA, USA) were placed on four locations along the instrumented leg: 1) over the anterior superior iliac spine (ASIS) to measure the motion of the trunk; 2) on the thigh, between the knee and hip but closer to the hip; 3) on the shank, between the knee and the ankle just below the knee; and 4) on the foot, midway between the heel and the toe (Figure 2.3). While tri-axial accelerometers were used, measurements were only recorded from two axes as accelerations due to walking mainly occur in the sagittal plane. The accelerations were recorded for movements in the distal-proximal (up-down) and anterior-posterior (forward-backward) directions. All accelerometers were orientated to measure acceleration in the radial direction, along the length of the segment, or tangential direction, perpendicular to the long axis of the segment, as shown in Figure 2.3. Accelerations tangential to the leg were labeled trunk At, thigh At, shank At, and foot At, and accelerations along the radial direction of the segment were labeled trunk Ar, thigh Ar, shank Ar, and foot Ar. All the accelerometers were calibrated for each subject using a static calibration method [31]. A linear calibration was performed based on a comparison between the output of the stationary accelerometer to the known constant acceleration due to gravity ($g = 9.81\ m/s^2$).

2.2.5.2 Force Sensitive Resistors (FSR)

FSRs were used to determine the loading of the leg and to distinguish steps. Two FSRs were taped to the insole of the shoe of the instrumented leg; one was placed beneath the heel and the other beneath the medial toe.

2.2.5.3 Segment angle and angular velocity

The segment angles of the three segments of the lower extremity were calculated from motion capture data. The angle of a line segment connecting the hip joint (greater trochanter) marker and the knee joint (lateral epicondyle) marker was used to describe the angle of the thigh segment (Figure 2.4). The angle of the line segment connecting the knee joint marker and the ankle joint (lateral mallelous) marker was used to indicate the angle of the shank (Figure 2.4). The angle of the line segment of connecting the toe (2nd metatarsal) marker and the heel (calcaneal tuberosity) marker was used as the angle of the foot (Figure 2.4). All angles were determined with respect to the horizontal plane. Angular velocities were calculated by differentiating the segment angle determined from motion capture data.

2.2.6 Data Collection

Motion capture and analog data (from the accelerometers, FSRs, and EMGs) were collected synchronously through the Vicon motion capture system using the Vicon Workstation software and an analog-to-digital converter patch panel. All analog signals were sampled at a rate of 1200 Hz. The motion capture data were collected at 120 Hz. The motion capture data were subsequently digitized and reconstructed using Vicon Workstation. The analog and motion capture data were then imported into Matlab (The MathWorks, Natick, MA, USA) for further processing and analysis. Once in the Matlab environment, all data (both analog and motion capture) were low pass filtered using a digital first order Butterworth filter with a 3 Hz cutoff frequency and down-sampled to 60 Hz. Prior to filtering and down sampling, the EMG signals were full-wave rectified. The angular velocity signals were additionally filtered using a first order Butterworth filter with a cutoff frequency of 5 Hz to remove the noise introduced by differentiating the angle signal to determine angular velocity. The signal was filtered both in the forward and reverse direction to remove any time delay introduced by filtering. The data were then split into steps using the signal from the FSRs to distinguish between steps. The heel FSR signal was used to delineate individual steps for all but one subject. For one MI subject, the toe FSR signal was used for trials in which the subject walked without WalkAide stimulation because initial ground contact was made by the toe instead of the heel. In both instances, the beginning of a step was distinguished when the signal crossed a manually set threshold. Steps that were out of view of the motion capture cameras were discarded.

2.2.7 Muscle Activation and Deactivation

Muscle activation and deactivation times were determined using the AB EMG data. For each of the six muscles, a threshold was set based on the minimum of the signal plus 1-2 standard deviations. When the EMG signal crossed this threshold with a positive slope, the time it crossed (in terms of percent of the step cycle) was considered the activation time. When the threshold was crossed with a negative slope it was considered the deactivation time. The detected activation and deactivation times for each step were manually checked to ensure appropriate detection and compared to accepted values in the literature [29]. One activation and one deactivation time were determined for each step.

Because the MI subjects were not able to voluntarily activate all the muscles required for efficient walking and often had an altered activation pattern, the activation and deactivation times were calculated based on their cadence and the AB activation and deactivation times. The AB EMG activation and deactivation times for each muscle were plotted against the cadence at which the AB subjects walked. The data points where then fitted with a linear regression curve (Figure 2.8), one curve for activation and one curve for deactivation for each muscle. If the slope of the curve for a particular muscle in terms of activation (or deactivation) was found to be significantly different from zero, the equation of the line was used to determine the expected activation (or deactivation) for that muscle for the MI subjects. As such, the expected activation (or deactivation) times for the MI subjects were calculated based on their cadence and the equation of the regression curve. If, however, the slope of the regression curve was not significantly different from zero, the average activation (or deactivation) time across all four speeds was used as the expected activation (or deactivation) time for the MI subjects.

2.2.8 Error Calculation

With 6 muscles, each having 1 activation and 1 deactivation time, there were 12 event times to be determined in order to control all 6 muscles during walking. With 4 dual-axis accelerometers, 2 FSRs, 3 segment angles, and 3 segment angular velocities, there were 16 possible sensor signals that could be used to control the stimulation of the 6 muscles. In order to evaluate which of these signals would be most appropriate to determine when to turn stimulation of a muscle on or off, a measure of error for each sensor signal was established. This was done using a threshold intersection technique and root mean squared (RMS) error. For each of the 16 sensor signals, a series of 10 equally spaced thresholds were calculated. Figure 2.5 shows the thresholds for the shank angle of an AB subject. These thresholds all fall within the range of that signal. For the AB subjects, steps at all 4 speeds were evaluated together to ensure that any threshold and sensor signal selected would be applicable for a range of walking speeds. For the MI subjects, 2 separate conditions were considered: with WalkAide stimulation and without WalkAide stimulation. For each threshold, RMS error was calculated based on the difference between the time at which the threshold intersected the sensor signal and the EMG established activation or deactivation time. Intersections with a threshold could either involve a positive slope of the signal or a negative slope of the signal. The RMS error was calculated based on

$$RMS = \sqrt{\frac{\sum_{i=1}^{n}(T_i - t_i)^2}{n}}$$

where n was the number of steps (for AB this included steps at all 4 walking speeds), T_i was the intersection of the threshold with the signal for step i (shown as triangles on Figure 2.5 for one of the thresholds), and t_i was the activation or deactivation time based on EMG for the same step i. The resulting error value was in terms of the percent of the step cycle. The threshold with the smallest error was used as the most appropriate threshold for that signal. For the shank angle signal in Figure 2.5, the solid bold line shows the best threshold. Its error was recorded as the lowest error for that particular signal in predicting the evaluated muscle activation or deactivation. This process was repeated for all 12 event times and all 16 signals, as well as for the 10 subjects, including both stimulation conditions for the MI subjects. All error values are in terms of percent of the step cycle.

2.2.8.1 Eliminated Steps

When calculating the RMS error for some thresholds, steps had to be eliminated. A step was eliminated if the signal crossed the threshold more than 4 times in the same direction as the signal. Such a signal was deemed to be too noisy. Also, steps were eliminated if the signal did not intersect the threshold at all. If steps were eliminated for a given threshold, a note was made that indicated the number of steps eliminated as well as the reason the step was eliminated.

2.2.8.2 Acceptable Error

In order to determine the approximate error that was acceptable, the standard deviation of the EMG determined activation and deactivation times across the four speeds was calculated. This value was determined for all 12 event times. The standard deviation was selected as an approximation of the acceptable error because it indicates the noise associated with the EMG determined activation and deactivation times. As long as the RMS error for the activation and deactivation times predicted using the sensor signals were within the noise of the EMG determined activation and deactivation times, the error was considered to be tolerable and would not negatively affect the gait produced by stimulation based on the timing events.

2.2.9 Ranking

All the sensor signals were ranked based on the error value for their best threshold. An individual ranking was performed for each of the 12 event times as well as a global ranking for all the event times combined. For each AB subject, the 16 sensor signals were ranked from 1 to 16 based on the error value rounded to the nearest percent. The sensor signal ranked first had the lowest error value, and the sensor signal ranked 16th had the highest error value. These individual subject rankings were then combined to determine an overall ranking for all the AB subjects. Sensor signals with greater than 2.5% steps skipped were eliminated from this overall ranking. The top 5 sensor signals were determined for each of the 12 gait events and represented the best sensors for determining a particular gait event across all 4 speeds for able-bodied walking. To ensure these sensor signals would work for the altered walking pattern of mobility-impaired subjects, the top 5 sensor signals for each gait event were further evaluated using the MI data. The sensor signals were then re-ranked based on the combined error of the MI and AB data. The global ranking was determined by combining the ranking of the sensor signals for all 12 activation and deactivation times.

2.2.10 Statistical Analysis

The error values of each sensor signal determining the 12 activation and deactivation times were compared to each other using a one-way analysis of variance (ANOVA). As well, error values for MI subjects walking with and without WA stimulation were compared using a one-way ANOVA, as were error values for AB subjects versus MI subjects. Differences were considered to be significant for p ≤ to 0.05. Tukey's Honestly Significant Difference post-hoc analysis was used when significant differences were observed. All error values are presented as mean ± standard deviation.

2.3 Results

An example of a subset of data collected from an AB subject walking at the slow walking speed with a cadence of 0.5 Hz is shown in Figure 2.6. The data were filtered and down sampled, and the EMG signal was rectified. The heel FSR signal was used to separate the recordings into individual steps prior to analysis.

A total of 183 steps were analyzed. Of these, 68 were AB walking and 115 MI walking. The AB steps were divided into 4 groups based on the walking speed: 19 steps at the very slow cadence of 0.3 Hz, 19 steps at the slow cadence of 0.5 Hz, 15 steps at the normal cadence of 0.7 Hz, and 15 steps at the fast cadence of 1 Hz. The MI steps were split between two groups: 58 steps were walking without using the WalkAide stimulator and 57 steps were walking using the WalkAide stimulator to correct foot drop.

2.3.1 Muscle Activation and Deactivation Times

The EMGs of the 6 muscle groups for the AB subjects walking at the 4 different speeds are shown in Figure 2.7. As walking speed increased, the amount of time spent in stance phase decreased and the swing phase accounted for a greater percentage of the walking cycle. This resulted in a variation in the time of muscle activity with walking speed. The variation was predominantly evident in the TA EMG activation time, but was also visible to some degree in the EMG activation and deactivation times of the other muscles. Linear regression curves were fit to the EMG timing versus cadence data for each gait event. An example of trend lines for the activation and deactivation of the BF muscle is shown in Figure 2.8. For this muscle, across this limited range of speeds, the activation varied with walking cadence while the deactivation did not. This trend was similar for the other two thigh muscles. The linear regression lines fit to the activation times for the VL and RF muscles had slopes that were significantly different from zero, while the lines for the deactivation times for both muscles had slopes that were not significantly different from zero. As a result, to determine the MI subjects' activation and deactivation times for these 3 thigh muscles the equation of the linear regression line was used for activation while the average deactivation time for all AB steps was used as the deactivation time. For the muscles of the shank, both the TA and Sol muscles had linear regression lines with slopes significantly different from zero for both activation and deactivation. Thus the equations of these lines were used to calculate the activation and deactivation times for the MI subjects. For the MG muscle, the linear regression lines for both activation and deactivation did not have slopes that were significantly different from zero. For this muscle the average activation and deactivation times for all the AB steps were used as the activation and deactivation times for the MI subjects. This is summarized in Table 2.1.

2.3.2 Approximate Error

In order to determine an approximate error that would be acceptable for the sensor signal, the 'jitter' in the EMG activation and deactivation times was calculated as the overall standard deviation across the 4 speeds. These values are summarized in the last column of Table 2.1 for each of the 12 muscle activation and deactivation times. The standard deviation or acceptable error was between 2 to 8 % of the step cycle for all 12 of the EMG activation and deactivation times.

2.3.3 Sensor Signals

Figure 2.9 shows the 16 sensor signals for a step cycle. The signals represent the average for all the AB steps at each of the 4 speeds. The signal for the slowest walking speed is shown in black with increasing speed in lighter shades of grey. As the speed increased, the shape of the accelerometer signals changed drastically. At the very slow and slow walking speeds, the acceleration signal was dominated by the static acceleration due to gravity. As the walking speed increased, the acceleration signal became dominated by the dynamic components of acceleration due to the translational and rotational acceleration of the segment. As a result the shape of the acceleration signals changed with speed. The acceleration signals for the accelerometers sensitive to radial accelerations were relatively flat for the very slow walking speed due to a lack of dynamic acceleration and a dominance of the static acceleration of gravity. The shape of the sensor signals for the all segment angles and the FSRs remained fairly constant across the four walking speeds. The shape of the angular velocity signal was also fairly constant but with a large change in amplitude with speed.

2.3.4 Eliminated Steps

Table 2.2 summarizes the percentage of eliminated steps for each of the 16 sensor signals in terms of the 12 event times. Very few steps were eliminated because of a lack of intersection with the threshold. This was expected because of the manner in which thresholds were selected that ensured they remained within the range of the given sensor signal. The vast majority of steps were eliminated because of a noisy sensor signal. The thigh accelerometers had the most eliminated steps: 16% of the steps were eliminated for the thigh At sensor signal, and 18% for the thigh Ar sensor signal. The shank and trunk accelerometers had 6-10% of their steps eliminated. The sensor signals with the least number of eliminated steps were the angle signals. The thigh, shank, and foot angle signals all had less than 0.2% of their total steps eliminated.

2.3.5 Error Values

Figure 2.10 shows the error values of the 16 different sensor signals calculated for determining the deactivation time of the RF muscle. Similar plots were obtained for the activation of this muscle, as well as the activation and deactivation of the 5 other muscles. The average error of each sensor signal is shown for the AB subjects walking at 4 different speeds, the MI subjects walking without stimulation, and the MI subjects walking with stimulation from the WalkAide to correct foot drop. The error values for the MI subjects walking with and without stimulation from the WalkAide were not significantly different for any of the 12 activation and deactivation times (one-way ANOVA, $p > 0.05$). For deactivation of the RF muscle, the errors for the AB subjects were significantly different than the errors for the MI subjects (one-way ANOVA, $p \leq 0.05$). This was also true for deactivation of the TA, Sol, MG, and VL muscles, as well as the activation of the TA and BF muscle. Based on the error value alone, the heel FSR, toe FSR, trunk At accelerometer, thigh angle, and shank angle appeared to be the most appropriate sensors for determining when to turn stimulation of the RF muscle off. However, for the deactivation of the RF muscle, the heel FSR and trunk At accelerometers had 4% and 5% eliminated steps respectively, while the toe FSR, thigh angle, and shank angle had no eliminated steps. As a result, the heel FSR and trunk At accelerometer were eliminated from the ranking of the most appropriate sensor signals to control the deactivation of the RF muscle. This is summarized in Table 2.3b, which along with Table 2.3a also shows the top ranked sensors for the 11 other event times.

2.3.6 Best Sensors for Determining Muscle Activation and Deactivation Times

To meet the requirements to be an appropriate sensor signal, the calculated RMS error for the sensor signal should be near or below the standard deviation of the EMG determined activation or deactivation time. The signal should not contain a large flat section near the threshold. As well, there should be no eliminated steps for that sensor signal.

The best sensor signal for activation of the TA muscle was the foot angle. The toe FSR and foot angular velocity signals also performed well; however, only the foot angle signal had a combined error (3.1 ± 1.3%), calculated from the AB and MI subjects data, lower than the approximated acceptable EMG activation error of 3.2%. The toe FSR and foot angular velocity did have low combined errors with values of 3.5 ± 0.9% and 4.0 ± 2.0%, respectively. The best sensor signals for deactivation of this muscle were the thigh angle (combined error = 5.6 ± 1.6%), toe FSR (combined error = 5.9 ± 1.7), and shank angle (6.4 ± 1.8). All had combined errors lower then the approximate acceptable error (7.4%) for this event time. For TA activation and deactivation there was no significant difference between the error values for the top 10 and top 13 ranked sensors, respectively, however many of the top 10 ranked sensors were not considered as appropriate sensors because of the number of eliminated steps.

For the Sol muscle, the best sensor for activation was the shank angle (combined error = 4.4 ± 2.0). The thigh angle (combined error = 5.6 ± 3.3) and trunk Ar accelerometer (combined error = 5.5 ± 2.5) were also ranked highly. However, they did not have combined errors smaller then the acceptable error of 5.1% for this event time. For the deactivation of the Sol muscle, both the foot angle (combined error = 3.1 ± 1.2) and toe FSR sensor (combined error = 3.6 ± 2.0) signals had combined errors lower than the acceptable error of 4.6%. For activation and deactivation of the Sol muscle, there was no significant difference between the error values for the top 12 and top 11 sensors, respectively.

The MG muscle had two sensors with combined errors less than the acceptable error for determining activation (5.3%). These were the shank angle (combined error = 4.8 ± 2.5%) and thigh angle (5.2 ± 3.8%). The shank angle (combined error = 4.5 ± 2.1%) also had a combined error lower than the acceptable error (4.5%) for the deactivation of this muscle, as did the foot angle (combined error = 2.9 ± 1.6%). For MG activation and deactivation there was no significant difference between the error values for the top 11 and top 12 ranked sensors respectively.

For VL the best sensor signals for activation of the muscle were the shank angle (combined error = 2.7 ± 1.5%) and foot angle (combined error = 3.4 ± 2.3%). Both had combined errors that were lower than the acceptable error (3.4%) for this event time. The shank angle as well as the thigh angle were the most appropriate to determine the deactivation of this muscle. For VL activation and deactivation there was no significant difference between the error values for the top 11 and top 14 ranked sensors respectively.

The top three sensor signals for determining the activation of the RF muscle were the shank angle (combined error = 3.4 ± 2.1%), the foot angle (combined error = 4.0 ± 2.9%), and the thigh angular velocity (combined error = 3.8 ± 1.9%). All three had combined errors within the range of acceptable error (4.5%). The shank angle (combined error = 5.6 ± 2.5%), was also one of the top three sensors to determine the deactivation of this muscle. It had a combined error value lower than the acceptable error (8.0%) for RF deactivation as did the thigh angle (combined error = 5.4 ± 2.5%), and toe FSR signals(combined error = 7.8 ± 2.9%). For the activation and deactivation of the RF muscle, there was no significant difference between the error values for the top 13 and top 15 ranked sensors respectively.

For the BF muscle, none of the top sensors had combined error values smaller then the approximate acceptable error of 2.1%. The top three sensor signals were the foot angular velocity (combined error = 4.3 ± 0.7%), the shank angle (combined error = 4.4 ± 1.4%), and the thigh angular velocity (combined error = 4.0 ± 2.0%). All had combined error values less than 5% of the step cycle which while not less than the acceptable error may still be low enough not to negatively affect walking. For the deactivation of this muscle, the thigh angle (combined error = 4.7 ± 2.2%) was the only sensor signal with a combined error smaller than the acceptable error (5.7%) for this event time. However, the shank angle's combined error value (of 5.9 ± 2.2%) was very close to the approximate acceptable error. For activation and deactivation of the BF muscle, the error values for the top 10 and top 13 ranked sensors respectively, were not significantly different.

When the ranking for all 12 gait events was combined to determine one sensor signal that would perform the best for providing feedback to control all 6 muscles, the sensor signal that ranked the highest was the shank angle. Figure 2.11 shows a box plot that summarizes the error values for 7 sensor signals that had less the 2.5% of their steps skipped. Each box represents the error values for all 12 event times. The lower line of the box represents the 25th percentile while the top line is the 75th percentile. The line in the middle of the box represents the median value. The error bars of each box show the minimum and maximum error values for that sensor signal. The boxes are ordered from the lowest median value to the highest. The shank angle had the least spread in error values as well as the smallest range of errors, and the lowest median error value.

2.4 Discussion

The overall goal of this study was to determine the most appropriate set of external sensors that would provide reliable feedback to control FES for walking. Of the 16 sensor signals studied, the segment orientation angles performed the best. In particular, the shank segment angle was the most appropriate sensor to control all 6 muscles if the use of a single sensor signal is desired.

Often, footswitches such as FSRs are used to provide feedback to FES walking systems [1, 16]. These sensors work very well for detecting events that occur during the stance phase of walking; however, they provide no information during the swing phase. We found similar results in this study. The toe and heel FSRs performed poorly for muscle activations and deactivations that occurred during the swing phase of walking, but the toe FSR was in the top three sensors for many of the muscle activation and deactivation times that occurred during stance. The position of the sensor also limited the success of its signal. The heel FSR sensor signal did not provide a useable signal for one of the MI subjects because he walked on his toes. While it has been shown that loading information is vital to control walking [32], on its own, the load signal from the FSR did not perform well enough to control the 6 muscles investigated. If loading information is desired, FSRs could be combined with segment orientation angles to provide control signals for FES walking.

Accelerometers are increasingly being investigated as attractive sensors to provide feedback for FES because of their small size, low cost, and low power consumption [15, 31]. Accelerometers have been used successfully to determine gait events [13, 24, 33]. The techniques employed to used accelerometers successfully can require complex processing of the signals often combining the signals of several sensors. Using simpler techniques to reduce computational load can cause limitations on walking speed. In this study we aimed to identify appropriate sensors that would work across several speeds and with minimal computational load. For these requirements, the signals from accelerometers placed on the trunk, thigh, shank, and foot did not perform well. One of the main challenges with the accelerometer signals was the noise within the signal even after filtering. This noise could be due to skin motion artifacts, especially in light of the fact that the thigh accelerometers had the largest noise. The femur of the thigh is buried deep beneath layers of skin, muscle and fat. This increases the likelihood of skin artifacts for accelerometers placed on the skin above the femur [37]. Another challenge with the accelerometer signals meeting our requirements was the large fluctuations in the shape of the sensor signals that occurred with walking speed. This can be seen clearly in Figure 2.9 (f) showing the change in shank At signal with cadence. The change in accelerometer signals across this range of speed is due to the components of acceleration that make up the signal. At slower walking speeds, the acceleration of the leg is dominated by the static acceleration of gravity. As walking speeds increase, the dynamic components of acceleration due to the translational and rotational acceleration of the leg begin to dominate, changing the shape of the signal. This change in signal with speed can limit the utility of an FES walking system using the sensor signal to accommodate several walking speeds without increasing the computational load.

The segment orientation angle signals performed the best for determining the activation and deactivation times of the 6 muscles using threshold intersection. This was due to the consistency of the shape of the sensor signal over several walking speeds. At least 1 of the 3 segment angle signals was within the range of acceptable error for all but 1 muscle activation and deactivation time. The only exception was the activation time for the BF muscle. However, even for this muscle the shank angle had a combined average error for the AB and MI subjects of only 4.4%. Although higher than the 2.1% acceptable error, this error is most likely small enough to not cause adverse effects if the shank angle were used as a feedback signal for FES-assisted walking. For the slowest walking speed examined, this 4.4% error would only be an error of 132 ms.

The error values for the MI subjects were, in most instances, smaller than the error values for the AB subjects. This may be because the AB subjects walked over a range of speeds, while the MI subjects only walked at one speed. As a result, the best threshold for the AB subjects had to work well for all 4 speeds, while the threshold for the MI subjects only had to work well for one speed.

While there was no significant difference between the error values for many of the top ranked sensors for all 12 activation and deactivation times, many of the error values for sensor signals were misleading because of the large number of eliminated steps. When eliminated steps were taken into account, the list of top ranked sensors shrunk considerably. The orientation angles remained consistently at the top with small errors and very few eliminated steps. As the sample size was small, it was difficult to find a significant difference between all error values.

There was also no significant difference between the error values for subjects walking with and without stimulation from the WalkAide. This may be due to residual effects of the stimulation that carryover even after the user is no longer using the stimulator. Trials without stimulation of the WalkAide were performed prior to trials with the WalkAide; however, all MI subjects were currently using the WalkAide in their daily life.

In some instances the MI subjects walked with a significantly longer stance phase than would be expected for an AB subject walking at the same cadence. As a result, some of the errors for the MI subjects may be due to real delays in the transition from stance to swing. However, because the aim is to restore a more efficient gait, resembling AB subjects, determining a triggering point that could predict when to turn on stimulation even with the altered walking pattern of a MI subject should allow a more natural activation of the muscles of an MI subject during walking.

Among the segment orientation angles, the shank segment angle was the best overall sensor to provide feedback for the control of all 6 muscles investigated. The shank segment angle was the best sensor signal for 4 of the 12 activation and deactivation times and in the top three for 10 of the 12 of these activation and deactivation times. Therefore, the shank segment angle can be used for controlling the stimulation of the TA, Sol, MG, VL, RF, and BF muscles using only threshold intersections. The shank segment angle has been reported in the literature as being a favorable signal to control TA activation for foot drop stimulators [3, 28]. This result is repeated here as well as extended to 5 other muscles used during walking. The findings in this study suggest that in addition to being an excellent signal to control muscles of the shank, the shank segment angle is even a top sensor for controlling the activation and deactivation of muscles of the thigh.

In order to utilize the shank segment angle signal in an FES device, it has to be easily obtained from a sensor. In 1990 Willemsen proposed a method to use pairs of 2 uni-axial accelerometers for a real-time calculation of leg angles[34]. This method relied on assuming rigid-body dynamics and simple hinge joints for the leg, and was used to calculate relative joint angles. Absolute segment angles could only be determined during the stance phase of walking which limits the applicability of this method. Gyroscopes are one of the most investigated sensors to provide segment angles because the angular velocity signal they yield can be integrated to determine the segment angle of the segment to which it is attached. The main challenge in using integrated gyroscope signals is the drift error that results from the DC offset in the un-integrated gyroscope signal. This DC offset varies randomly and is thus difficult to account for. Several techniques have been investigated to remove the drift error in the integrated gyroscope in real time so it can be used as a sensor for FES applications. The integrated signal has been reset to a known angle for each step cycle using footswitches or accelerometers to determine when to reset [20, 26]. This method works well, but requires additional sensors to be used, and in the case of the FSR, it limits the user to having to wear shoes. Kalman filters have also been investigated to remove the drift for the integrated gyroscope signal [35, 36]. The Kalman filter method uses an accelerometer to estimate the tilt of the segment and compares this estimate to the segment angle calculated from the integration of the gyroscope signal in order to remove the drift. While this works better than using an integrated gyroscope signal on its own, there are still errors in the calculated segment angle. Dejnabadi et al developed a method that estimated the drift in the integrated gyroscope signal at unknown times using an interpolation technique based on piecewise cubic hermite interpolation applied on the drift signal at known intervals [37, 38]. This drift is then subtracted from the integrated gyroscope signal to yield the absolute shank angle at all times. This method has a faster response time, no phase delay, and less computational load than using the Kalman filter method; however, it is limited to post processing and uses a biaxial accelerometer in addition to the gyroscope.

2.5 Conclusion

To summarize, this study demonstrates that the shank segment angle is the best overall sensor signal to determine the activation and deactivation times of 6 muscles if only one sensor signal is desired. If additional sensor signals are acceptable, the segment angles of the thigh, shank, and foot can be used to determine the activation and deactivation times of the TA, Sol, MG, VL, RF, and BF muscles with less than 6% error. The results indicate that segment angles are the most appropriate sensor signals to provide feedback to an FES walking system controlling these 6 muscles.

2.6 Figures and Tables

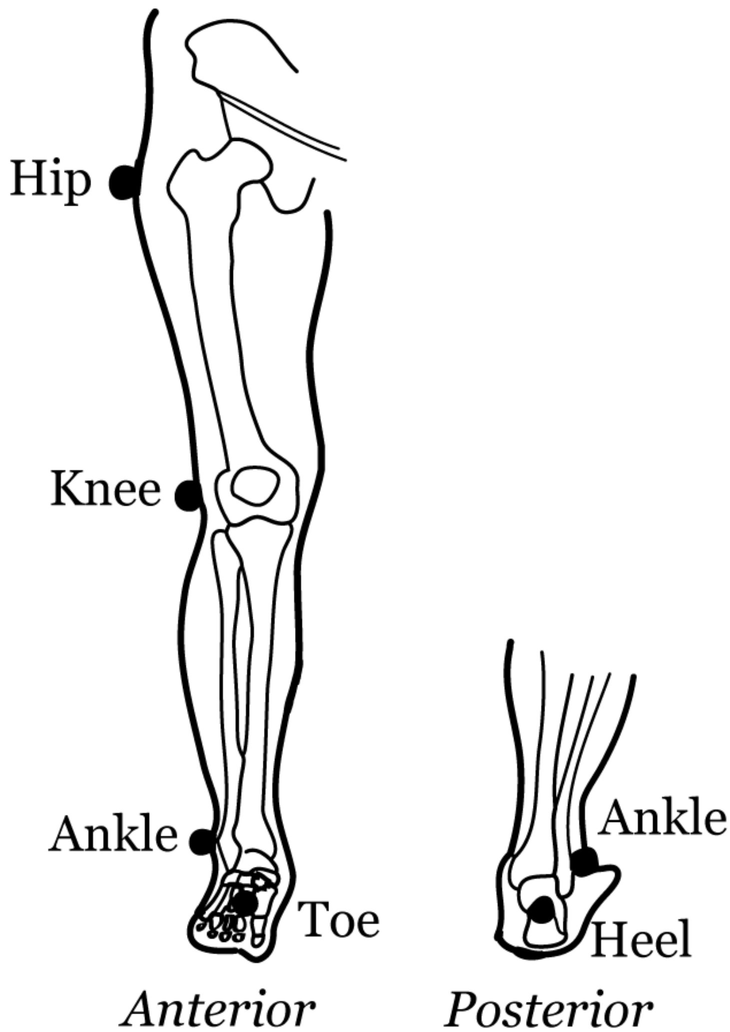

Figure 2.1 – Motion Capture Marker Locations

Markers were placed on the lower extremity to measure the motion of the leg while the subjects walked in view of a Vicon motion capture system (Oxford Metrics Group, Oxford England). The hip marker was placed over the greater trochanter, the knee marker was placed on the skin over the lateral epicondyle, the ankle marker was placed on the skin over the later malleolus, the heel marker was placed on the shoe over the calcaneal tuberosity, and the toe marker was placed on the shoe over the 2nd metatarsal.

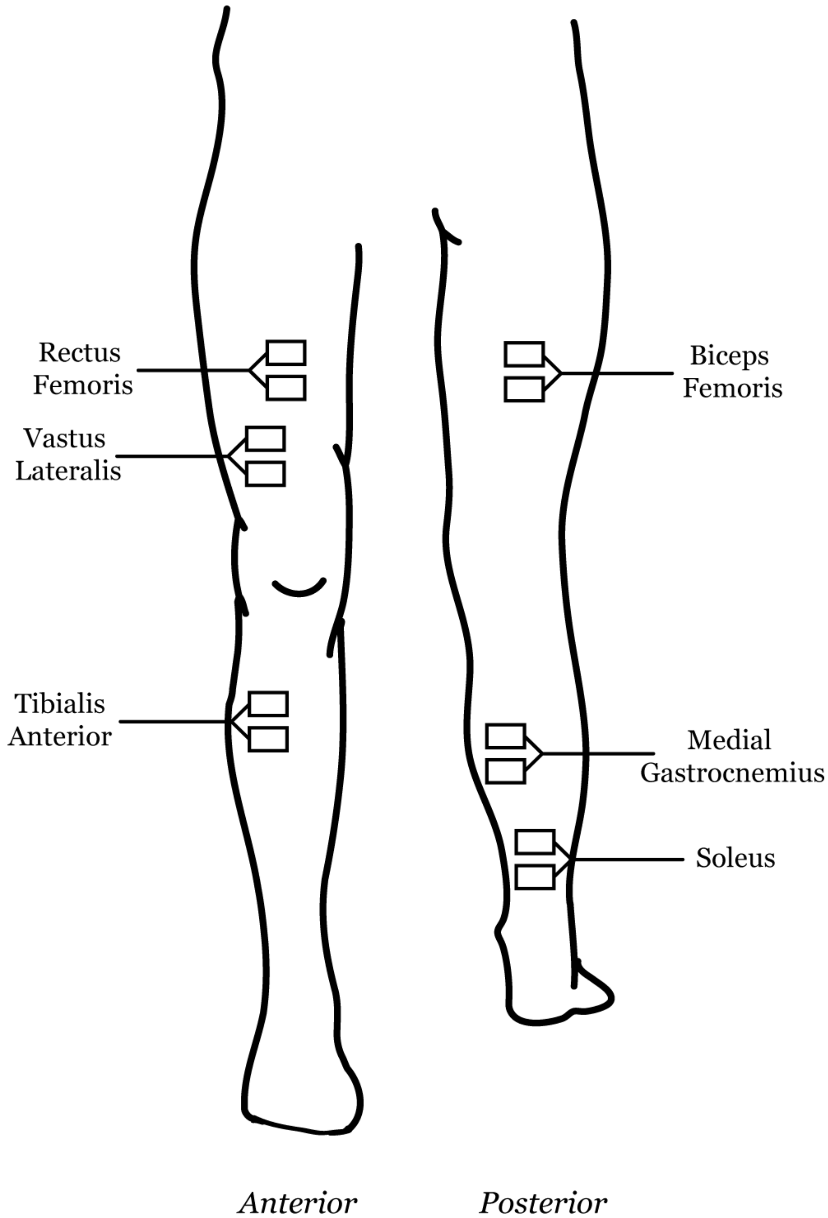

Figure 2.2 – EMG Electrode Placement

EMG electrodes were placed on the skin as shown to measure the activity of the tibialis anterior, soleus, medial gastrocnemius, vastus lateralis, rectus femoris, and biceps femoris muscles during walking.

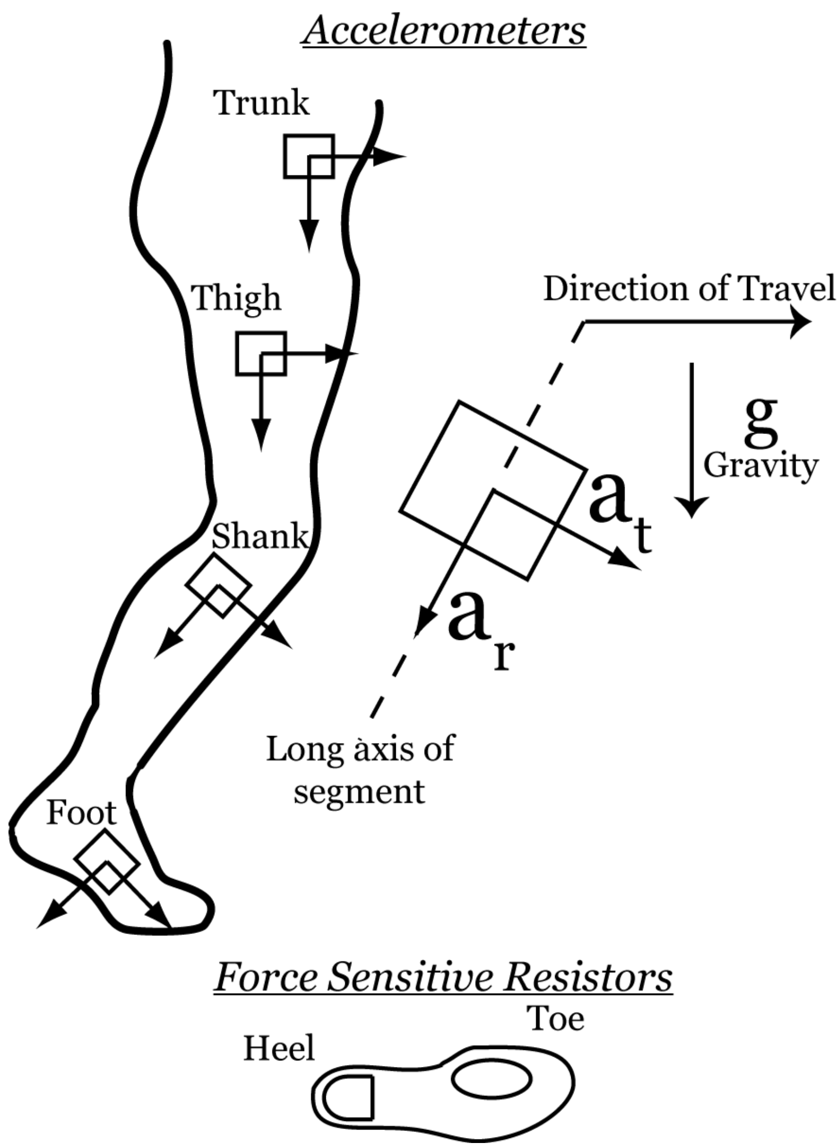

Figure 2.3 – Sensor Locations

Accelerometers were placed on 4 locations along the lower extremity: the trunk, the thigh, the shank, and the foot. The accelerometers were sensitive to accelerations in the radial (a_r) and tangential (a_t) direction as shown. FSRs were placed in the heel and toe of the shoe insole.

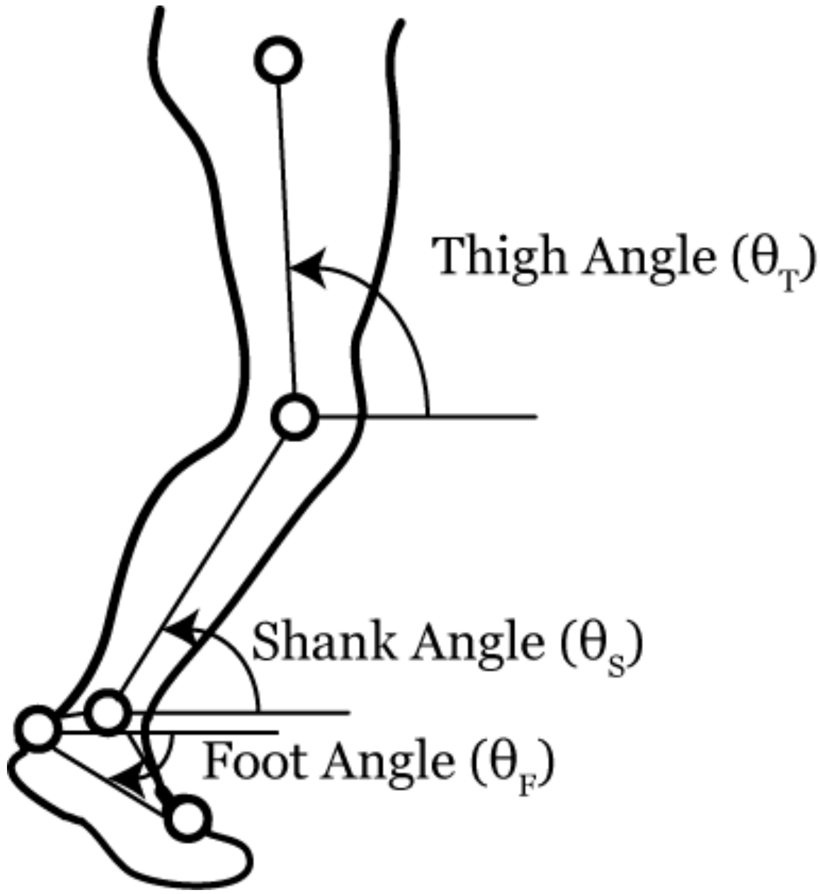

Figure 2.4 – Segment Angles

The segment orientation angles of the 3 segments of the lower extremity were calculated from the motion capture data. The thigh angle was calculated as the angle of the line connecting the hip and knee marker. The shank angle was calculated as the angle of the line connecting the knee and ankle markers. The foot ankle was calculated as the angle of the line connecting the heel and toe marker. All angles were calculated with respect to the horizontal.

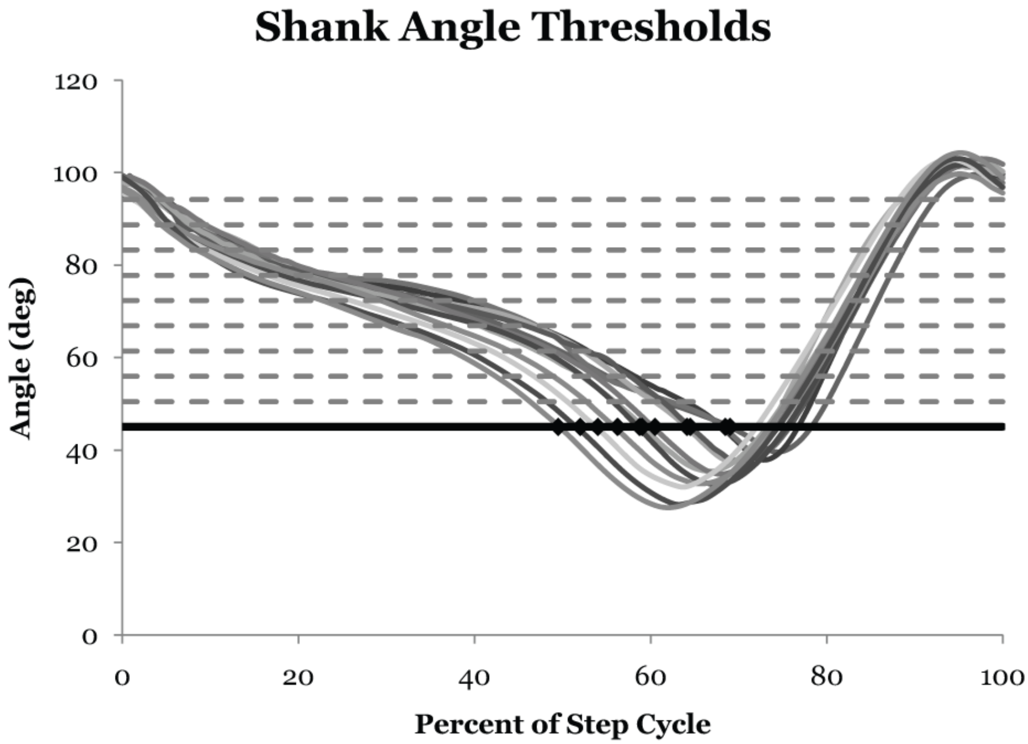

Figure 2.5 – Thresholds Example

Error values based on RMS error are calculated for 10 equally spaced thresholds that fall within the range of the sensor signal. The RMS error is based on the difference between the time at which the threshold intersects each step's signal (shown by the black diamonds) and the EMG established activation or deactivation time. The best threshold, shown by the solid bold line, is the one with the smallest error value.

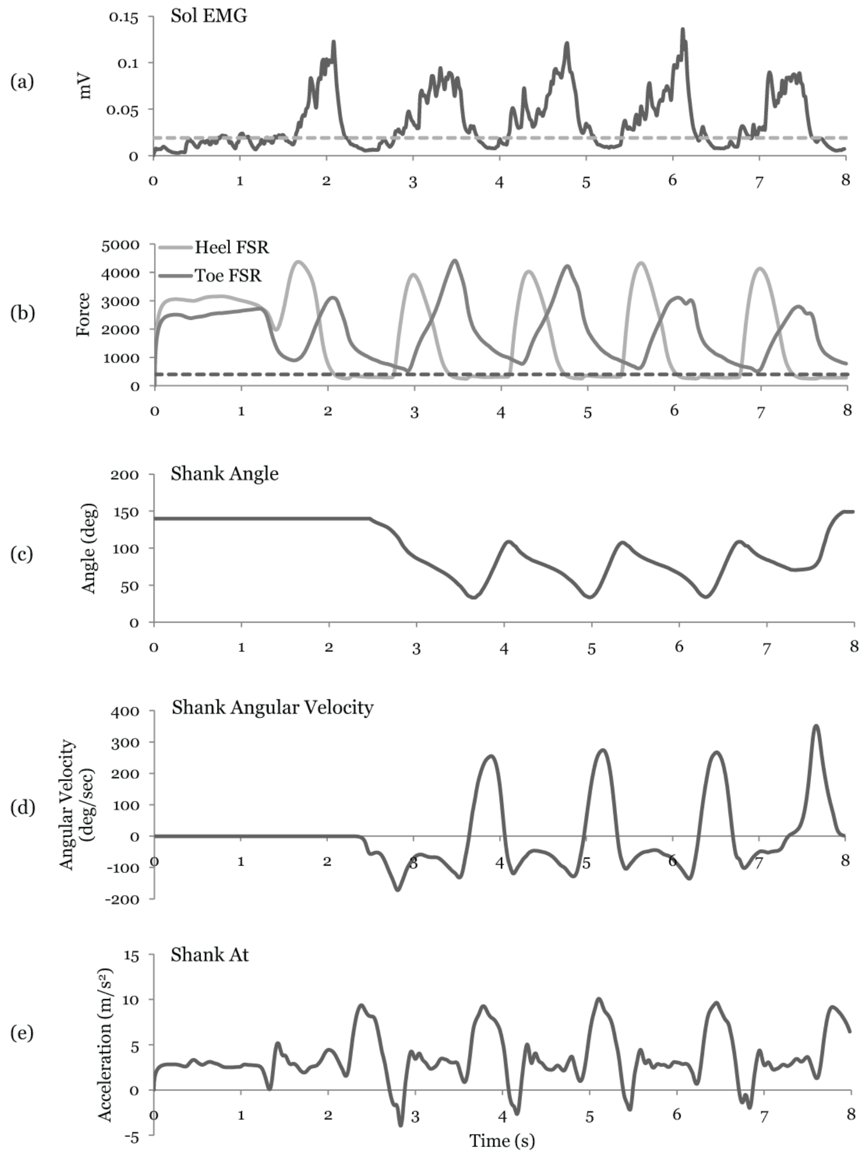

Figure 2.6 – Select Unprocessed Data

Data shown for an AB subject walking at the slow walking speed (cadence 0.5 Hz) has been filtered with a 1st order Butterworth filter with a cutoff frequency of 3 Hz as well as down-sampled to 60 Hz. The soleus EMG (a) has been full wave rectified. The dashed line shows the threshold used to determine the activation and deactivation time of this muscle. The FSR signals for the heel and toe (b) are used to split the data into steps. Intersection of the dashed line threshold with the heel signal indicates a new step. Position data calculated from motion capture data shows the shank angle (c). These data were integrated to determine the angular velocity of the shank (d). The signal for the accelerometer placed on the shank measuring accelerations tangential to the long axis of the leg is shown in (e).

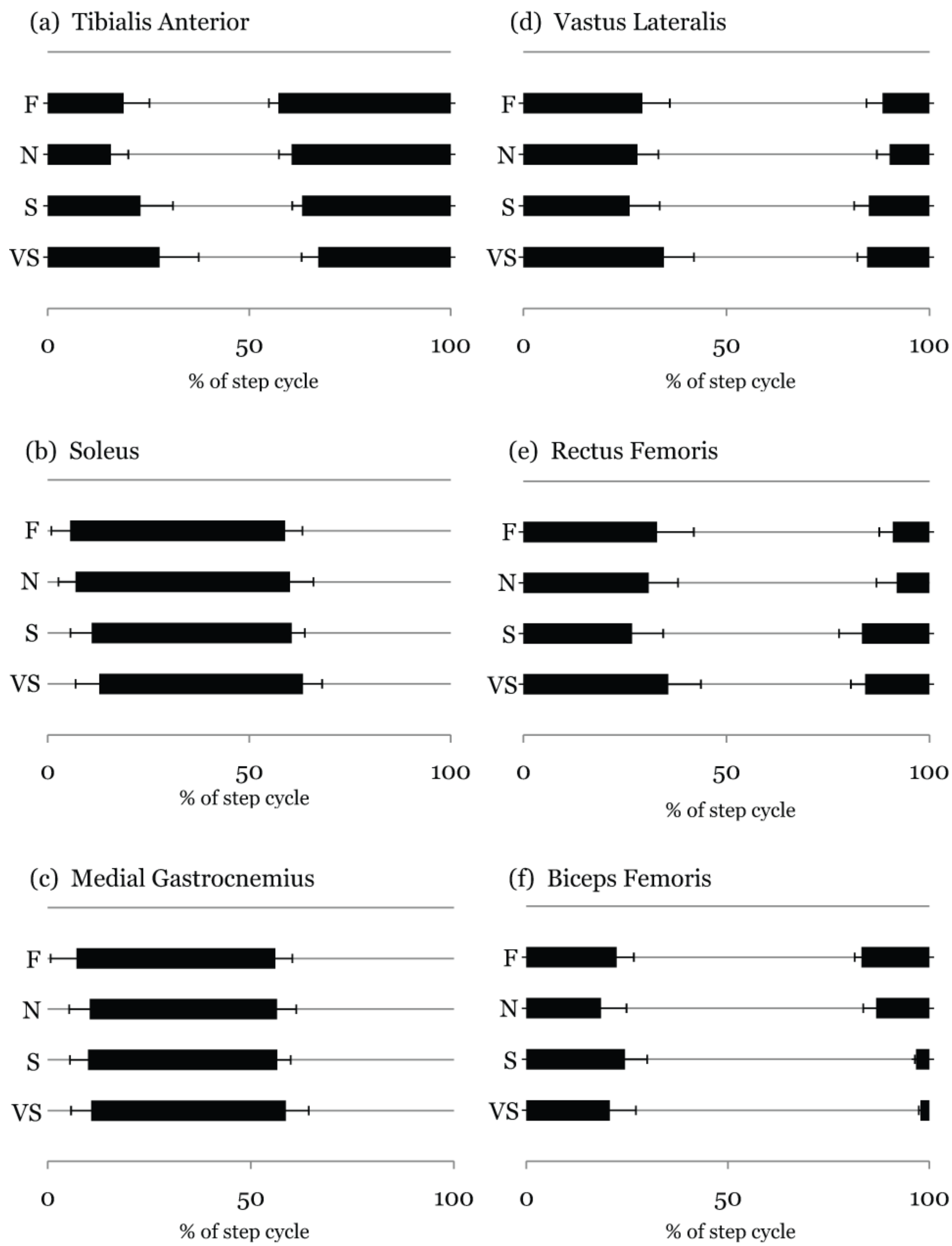

Figure 2.7 – AB Average EMG Activation and Deactivation Times

The average EMG activation and deactivation times for the 5 AB subjects walking at four walking speeds (very slow (VS), slow (S), normal (N), and fast (F)) are shown with standard deviation for the tibialis anterior (a), soleus (b), medial gastrocnemius (c), vastus lateralis (d), rectus femoris (e), and biceps femoris (f) muscles.

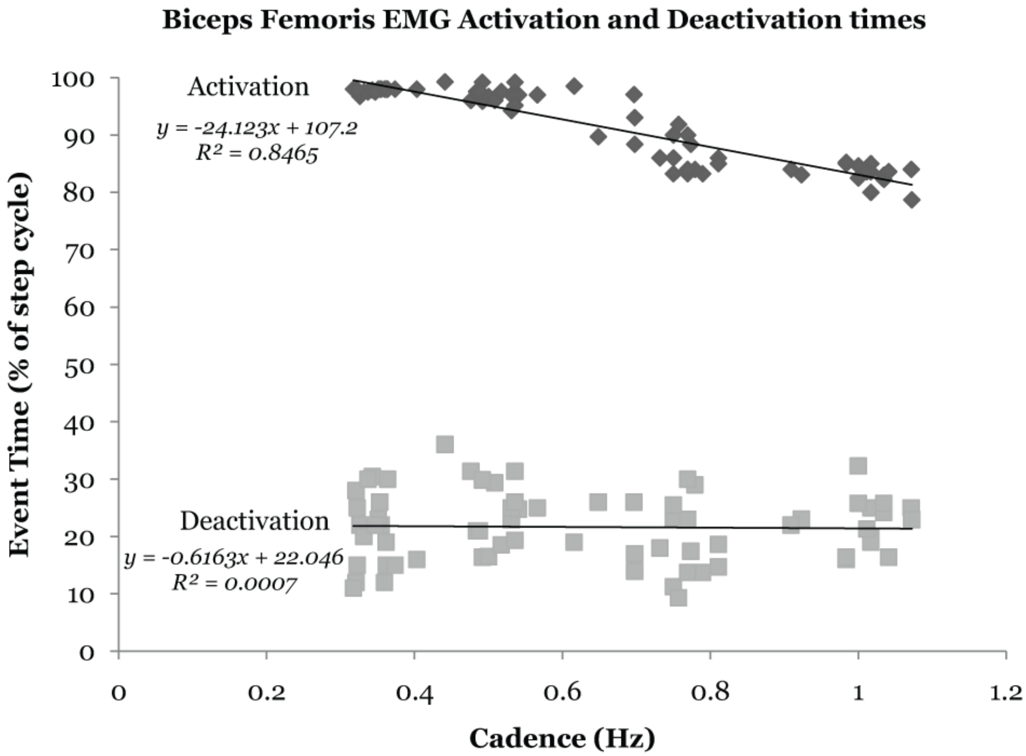

Figure 2.8 – BF EMG Activation and Deactivation Times vs. Cadence

The AB EMG activation and deactivation times for the BF muscle are plotted against the cadence at which the AB subjects walked. The activation and deactivation times for the MI subjects were calculated based on the trends in this data. The activation data are fitted with a linear regression curve with a slope that is significantly different than zero. As a result, the equation of this line was used to calculate the activation time of the BF muscle for the MI subjects based on the cadence. The linear regression curve fit to the deactivation data does not have a slope that is significantly different then zero. Thus the average of these data is used to determine the deactivation of the BF muscle for the MI subjects.

Muscle	Event	AB Overall Std Dev	MI	
			Technique Used	Value
TA	*Act*	3.2	*Equation*	-14.3x+71.5
	Deact	7.4	*Equation*	-14.5x+31.1
Sol	*Act*	5.1	*Equation*	-10.7x+16.2
	Deact	4.6	*Equation*	-5.4x+64.4
MG	*Act*	5.3	*Average*	9.7
	Deact	4.5	*Average*	57.1
VL	*Act*	3.4	*Equation*	7.4x+82.2
	Deact	6.7	*Average*	29.7
RF	*Act*	4.5	*Equation*	12.3x+79.4
	Deact	8.0	*Average*	31.7
BF	*Act*	2.1	*Equation*	-24.1x+107.2
	Deact	5.7	*Average*	21.7

Table 2.1 – EMG Data Summary

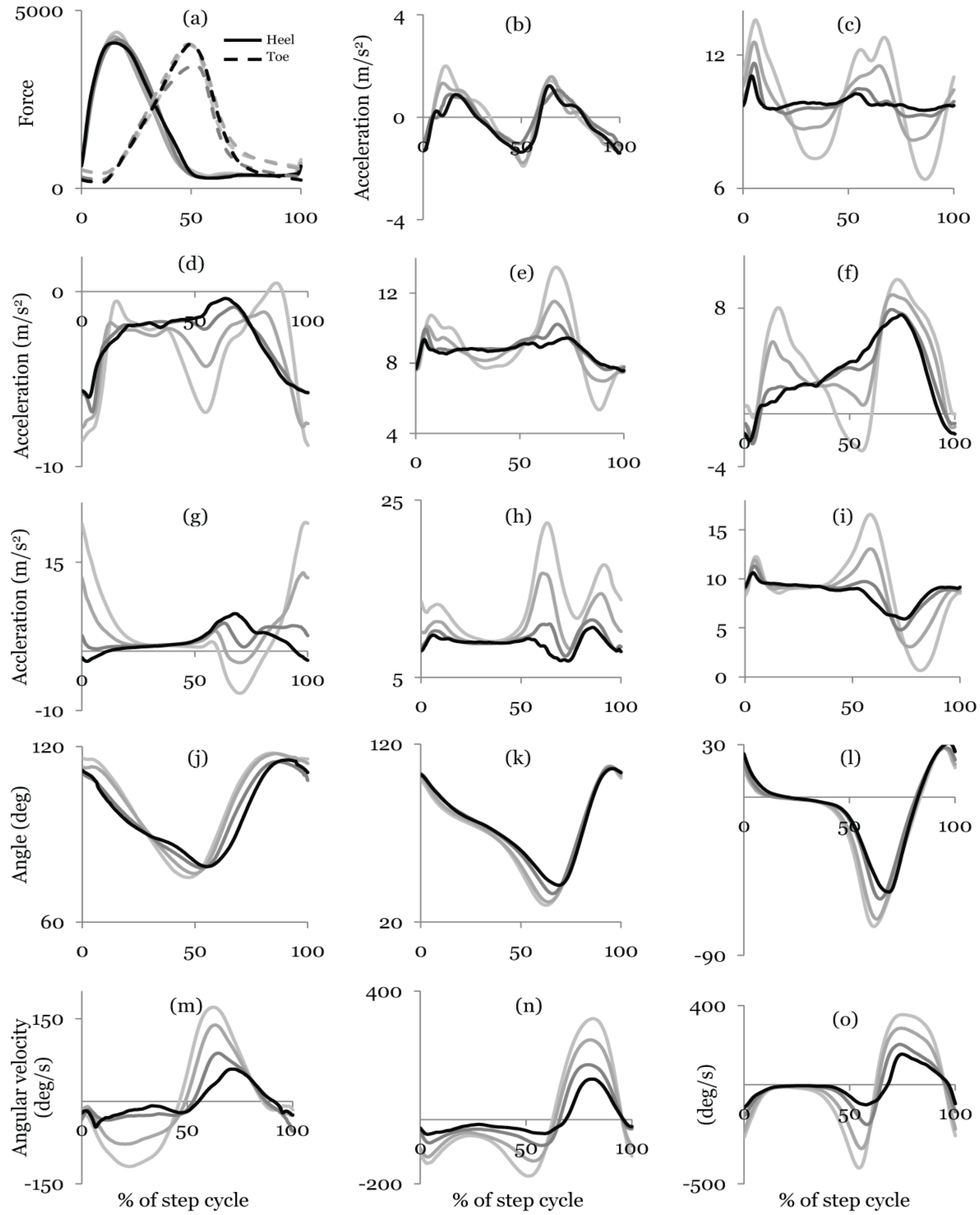

Figure 2.9 – AB Sensor Signals

Average sensor signals for the 5 AB subjects walking at all 4 speeds are shown for the 2 FSRs: heel and toe (a); the accelerometers: trunk At (b), trunk Ar (c), thigh At (d), thigh Ar (e), shank At (f), shank Ar (i), foot At (g), and foot Ar (h); the segment angles: thigh (j), shank (k), and foot (l); and the segment angular velocities: thigh (m), shank (n), and foot (o). The average signal for the very slow walking speed is shown in black, the slow walking speed in dark grey, the normal walking speed in grey, and the fast walking speed in light grey.

Sensor	Percentage of Steps Eliminated												
	TA		Sol		MG		VL		RF		BF		Total
	Act	Deact	Act	Deact	Act	Deact	Act	Deact	Act	Deact	Act	Deact	
Heel FSR	3	1	3	3	4	2	4	4	4	4	3	1	3
Toe FSR	0	0	0	0	0	1	3	0	3	0	3	0	1
Trunk At	6	8	5	6	3	5	5	5	5	5	7	8	6
Trunk Ar	11	14	2	7	2	4	7	15	12	22	6	13	10
Thigh At	10	25	12	4	13	16	15	24	16	19	15	21	16
Thigh Ar	18	23	20	14	12	9	19	21	20	19	19	21	18
Shank At	16	11	9	14	10	16	1	16	1	19	1	10	10
Shank Ar	5	8	5	4	5	7	7	15	7	16	5	7	8
Foot At	1	1	1	1	1	1	0	1	0	1	0	1	1
Foot Ar	4	9	5	4	5	4	0	8	1	6	3	6	5
Thigh Angle	0	0	1	0	1	0	0	0	0	0	0	0	< 0.2
Shank Angle	0	0	1	0	1	0	0	0	0	0	1	1	< 0.2
Foot Angle	0	0	0	0	0	0	0	0	0	0	1	0	< 0.2
Thigh Angular Velocity	2	4	3	1	3	2	1	1	1	1	2	2	2
Shank Angular Velocity	5	5	5	5	5	5	4	5	0	5	1	5	4
Foot Angular Velocity	1	4	1	1	1	4	4	4	4	4	0	2	2

Table 2.2 – Eliminated Steps

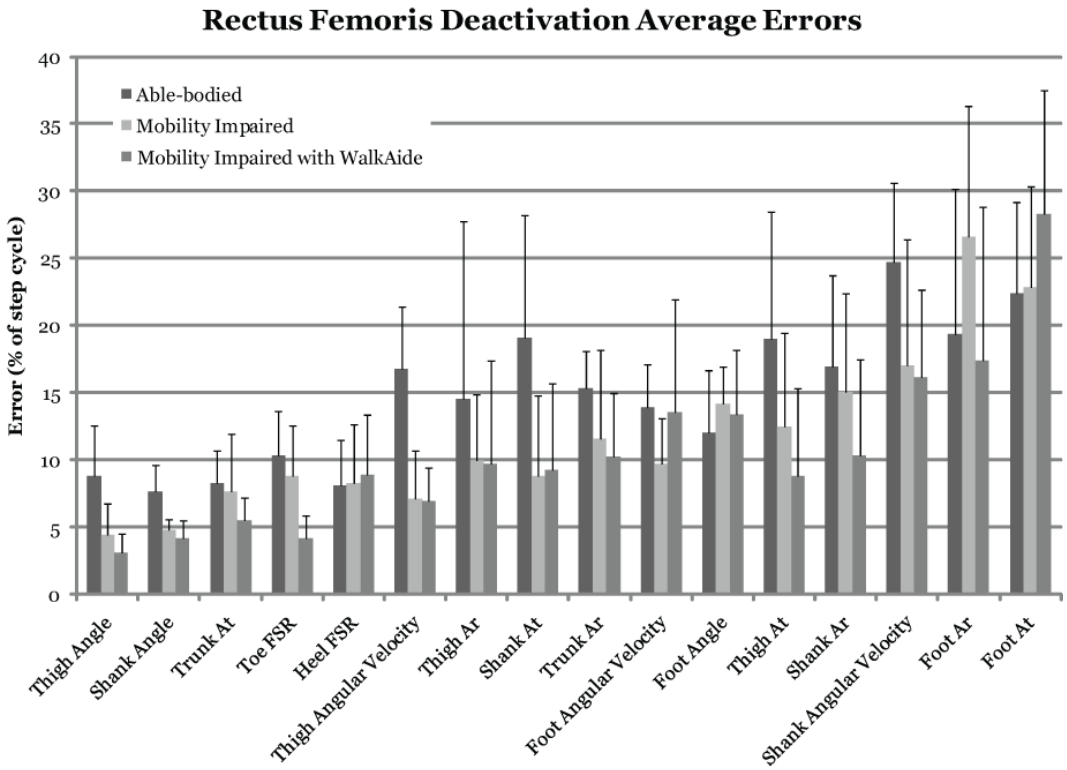

Figure 2.10 – RF Average Deactivation Errors

The error of using a threshold to detect the deactivation time of the RF muscle was calculated for all 16 of the sensor signals. For the 5 AB subject, the average error with standard deviation is shown for steps at all 4 walking speeds. Two conditions were evaluated for the MI subjects: walking without stimulation and walking with the WalkAide stimulator to correct foot-drop. Similar plots were obtained for all 12 activation and deactivation times.

Muscle	Activation					
	Sensor	*Eliminated Steps (%)*	*Combined Average Error*	*AB Average Error*	*MI Average Error*	*MI Average Error with WA*
TA	**Foot Angle**	**0**	**3.1 ± 1.3**	3.9 ± 1.0	2.8 ± 1.2	2.7 ± 1.6
	Toe FSR	0	3.5 ± 0.9	4.8 ± 0.7	3.2 ± 0.7	2.6 ± 1.3
	Foot Angular Velocity	1	4.0 ± 2.0	3.2 ± 1.0	4.7 ± 3.0	4.1 ± 1.9
Sol	**Shank Angle**	**1**	**4.4 ± 2.0**	4.9 ± 1.3	5.4 ± 3.7	3.0 ± 1.0
	Thigh Angle	1	5.6 ± 3.3	8.7 ± 2.0	4.8 ± 5.7	3.2 ± 2.2
	Trunk Ar	2	5.5 ± 2.5	8.3 ± 2.1	3.5 ± 2.2	4.8 ± 3.1
MG	**Shank Angle**	**1**	**4.8 ± 2.5**	5.1 ± 1.3	5.4 ± 4.8	3.9 ± 1.5
	Thigh Angle	**1**	**5.2 ± 3.8**	7.4 ± 1.9	4.8 ± 6.6	3.3 ± 3.0
	Trunk Ar	2	4.5 ± 2.3	7.5 ± 3.2	2.6 ± 1.5	3.3 ± 2.3
VL	**Shank Angle**	**0**	**2.7 ± 1.5**	3.1 ± 1.4	3.3 ± 2.2	1.8 ± 1.0
	Foot Angle	**0**	**3.4 ± 2.3**	3.2 ± 1.4	4.1 ± 2.8	2.9 ± 2.7
	Thigh Angular Velocity	1	3.1 ± 1.4	3.8 ± 0.8	2.7 ± 1.4	2.9 ± 1.9
RF	**Shank Angle**	**0**	**3.4 ± 2.1**	4.5 ± 2.6	3.9 ± 2.8	1.9 ± 1.0
	Foot Angle	**0**	**4.0 ± 2.9**	4.2 ± 2.7	4.6 ± 2.9	3.1 ± 3.0
	Thigh Angular Velocity	**1**	**3.8 ± 1.9**	5.2 ± 1.7	2.9 ± 1.8	3.4 ± 2.1
BF	Foot Angular Velocity	0	4.3 ± 0.7	7.5 ± 0.7	3.0 ± 0.6	2.5 ± 0.7
	Shank Angle	1	4.4 ± 1.4	7.6 ± 0.5	2.9 ± 1.7	2.7 ± 1.9
	Thigh Angular Velocity	2	4.0 ± 2.0	6.7 ± 3.0	3.2 ± 1.8	2.1 ± 1.2

* **Bold** terms indicated sensor signals with a combined error lower than the acceptable error for that muscle activation or deactivation

Table 2.3a – Top Sensor Signals for Activation

Muscle	Deactivation					
	Sensor	*Eliminated Steps (%)*	*Combined Average Error*	*AB Average Error*	*MI Average Error*	*MI Average Error with WA*
TA	**Thigh Angle**	**0**	**5.6 ± 1.6**	10.4 ± 1.7	3.5 ± 2.5	2.8 ± 0.5
	Toe FSR	**0**	**5.9 ± 1.7**	8.1 ± 2.1	5.5 ± 2.3	4.1 ± 0.8
	Shank Angle	**0**	**6.4 ± 1.8**	8.9 ± 1.1	5.8 ± 2.4	4.5 ± 1.8
Sol	**Foot Angle**	**0**	**3.1 ± 1.2**	3.7 ± 1.3	2.9 ± 1.0	2.6 ± 1.3
	Toe FSR	**0**	**3.6 ± 2.0**	3.6 ± 0.9	3.7 ± 1.8	3.5 ± 3.4
	Foot Angular Velocity	1	5.2 ± 3.2	4.3 ± 1.2	6.1 ± 4.7	5.3 ± 3.6
MG	**Foot Angle**	**0**	**2.9 ± 1.6**	3.2 ± 1.3	2.9 ± 1.4	2.6 ± 2.0
	Shank Angle	**0**	**4.5 ± 2.1**	4.3 ± 1.0	5.7 ± 3.5	3.6 ± 1.7
	Thigh Angular Velocity	2	4.7 ± 2.9	4.2 ± 1.0	5.8 ± 5.1	4.2 ± 2.5
VL	**Thigh Angle**	**0**	**4.7 ± 2.7**	7.5 ± 4.6	3.9 ± 2.4	2.6 ± 1.0
	Shank Angle	**0**	**5.8 ± 1.6**	7.7 ± 1.6	5.4 ± 1.8	4.2 ± 1.5
	Toe FSR	0	7.0 ± 2.8	8.9 ± 3.8	8.0 ± 3.2	4.1 ± 1.4
RF	**Thigh Angle**	**0**	**5.4 ± 2.5**	8.8 ± 3.8	4.4 ± 2.3	3.1 ± 1.4
	Shank Angle	**0**	**5.6 ± 1.4**	7.7 ± 2.0	4.8 ± 0.8	4.2 ± 1.4
	Toe FSR	**0**	**7.8 ± 2.9**	10.4 ± 3.3	8.8 ± 3.8	4.2 ± 1.6
BF	**Thigh Angle**	**0**	**4.7 ± 2.2**	6.6 ± 1.6	4.6 ± 4.1	2.9 ± 0.8
	Toe FSR	0	6.4 ± 2.0	8.9 ± 2.0	6.5 ± 2.3	3.7 ± 1.5
	Shank Angle	1	5.9 ± 2.2	6.9 ± 1.7	6.0 ± 2.7	4.9 ± 2.3

* **Bold** terms indicated sensor signals with a combined error lower than the acceptable error for that muscle activation or deactivation

Table 2.3b – Top Sensor Signals for Deactivation

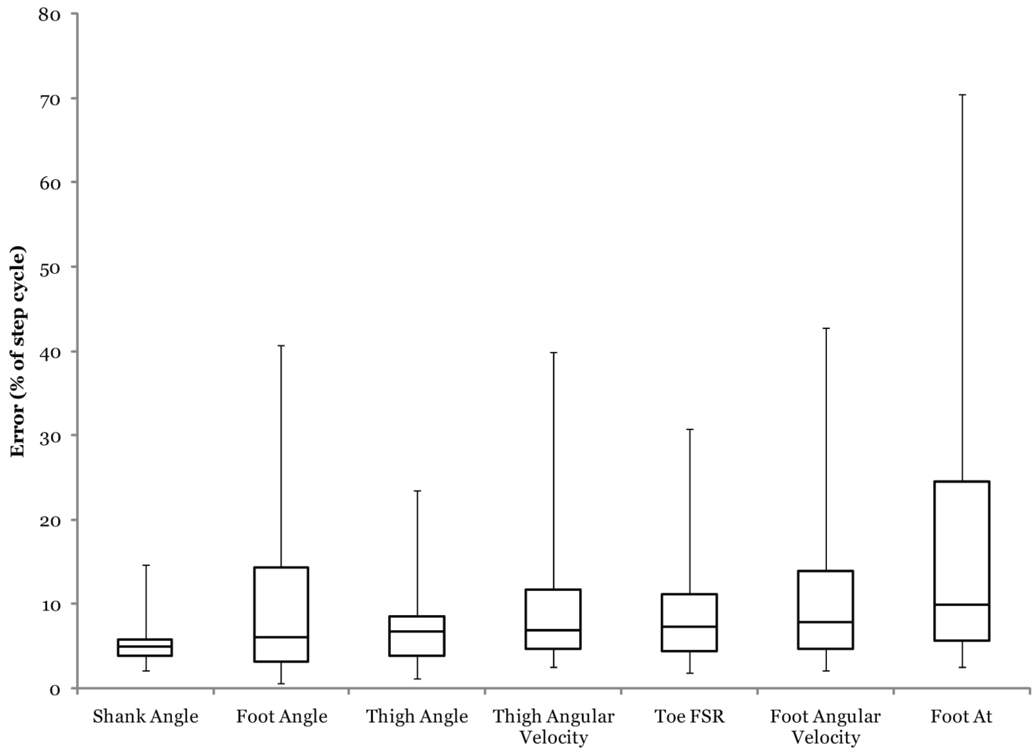

Figure 2.11 – Best Overall Sensor

Box plot of the combined average errors for all 12 activation and deactivation times and both AB and MI subjects. Only sensor signals with less then 2.5% of the steps eliminated are shown. The horizontal lines illustrate the median values while the upper and lower limits of the boxes represent the 75th and 25th percentile respectively. Full ranges of the error values are illustrated by the upper and lower limits of the vertical lines.

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Chapter 3 – General Conclusions and Future Directions

The main goal of my thesis was to determine the most appropriate sensor signals to control a functional electrical stimulation (FES) walking system. To achieve this, I compared the ability of 16 different sensor signals including accelerometers, force sensitive resistors (FSRs), segment orientation angles, and segment angular velocities, to determine the activation and deactivation time of 6 muscles during walking. Activation and deactivation times were determined for the tibialis anterior (TA), soleus (Sol), medial gastrocnemius (MG), vastus lateralis (VL), rectus femoris (RF), and biceps femoris (BF) muscles. Thigh, shank and foot segment angles performed the best for determining the activation and deactivation times of the 6 muscles. All activation and deactivation times were determined within 6% of the step cycle with at least one of the three segment angle signals. The shank segment angle was the best overall sensor to control all 6 muscles with one sensor signal.

3.1 Conclusions

My results indicate that the shank segment angle may be the best sensor signal to provide feedback to an FES system stimulating the TA, Sol, MG, VL, RF, and BF muscles during walking. If the FES system were able to incorporate more than one sensor signal, a combination of thigh, shank, and foot segment angle systems would provide even more accurate activation and deactivation triggering times.

Based on these results, techniques are being investigated to easily determine segment angles using low cost, low power miniature inertial sensors. If successful this would allow the signal to be incorporated in an FES device to improve walking in individuals with mobility impairments due to various neurological disorders such as stroke, spinal cord injury, and multiple sclerosis. Using one miniature sensor for control will help make the FES device minimally cumbersome.

3.2 Future Directions

The results from this thesis indicate that segment angles, especially the shank segment angle, can accurately and reliably determine the activation and deactivation of 6 major muscles used during walking. Some limitations of the current study need to be addressed when planning future investigations:

1. Because of limited space on the leg to attach sensors and electromyography (EMG) electrodes, motion capture data were used to determine the segment angle and angular velocity during walking. The segment angles were found to be the most appropriate sensor signals. It is therefore necessary

to examine the ability of a gyroscope or other sensor to provide accurate angle data in real time during walking. Many studies have shown the ability of using gyroscopes to determine segment angle [1-7]. The methods explored in these studies require additional sensors, a heavy computational load, or post processing. Work to develop a simpler technique to determine segment angle using a gyroscope that would work accurately, reliable, and in real time is currently underway in the Stein lab.

2. The accuracy of the sensor signals was limited by the accuracy of the EMG in detecting the activation and deactivation times of the muscles investigated. Because motion capture was used to generate some of the sensor signals, the walking distance was limited. As a result the EMG signals were not ideal. Future studies could utilize a sensor to determine segment angles over a larger walking distance. A greater number of steps would allow the EMG signals to be normalized.

3. For my thesis, walking was assumed to be a planar motion. As a result, sensor signals were only used to measure movements in the sagittal plane. For normal walking, the majority of the motion of the legs is in the sagittal plane. However, the altered walking patterns of individuals with mobility

impairments can result in some motion in other planes. To improve control in an FES walking system, sensors that measure motion in these other planes could be examined.

4. The thresholds selected for the sensor signals were by no means optimal. Because the thresholds were selected based on dividing the range of each sensor signal into 10 equally spaced thresholds, it could be possible to further optimize the best threshold. This could improve the accuracy of the sensor signals for determining activation and deactivation times. A future study could examine the top sensor signals to determine an optimal threshold for each activation and deactivation time.

5. The sensor signals evaluated in my thesis were generated from walking. In order to use these signals to turn stimulation on, the user must retain some voluntary control of their muscles in order to initiate the step. The sensor signals can then be used to trigger stimulation of muscles to help improve the efficiency of walking. If these sensor signals are to be incorporated in a device to control walking in an individual with complete paraplegia, a sequenced pattern of stimulation may be required to initiate stepping and generate the motion needed to produce the required sensor signals. A future study

would need to evaluate the ability of the segment angle sensor signals to work with stimulation.

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