Investigating the Effects of Custom Made Orthotics on Brain Forms: A Pilot Study

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A thesis submitted in partial fulfillment of the requirements for the degree in Master of Science  
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INVESTIGATING THE EFFECTS OF CUSTOM MADE ORTHOTICS ON BRAIN FORMS: A PILOT STUDY

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by

Lindsay Carey

Graduate Program in Health and Rehabilitation Sciences

A thesis submitted in partial fulfillment of the requirements for the degree of Master of Science

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Abstract

OBJECTIVES: To determine (1) the feasibility of this novel approach and technique of recording brain activity, wirelessly and continuously, during human gait, and (2) if custom made orthotics will alter the brain activity patterns recorded. METHODS: Gait trials were performed on 16 participants walking with and without orthotic devices in their shoes while simultaneously collecting EEG data through the Emotiv wireless neuroheadset. RESULTS: The Emotiv neuroheadset was capable of detecting changes in brain activity between the two gait trials. The differences in brain activity identified between conditions were not statistically significant. CONCLUSION: The findings suggest the Emotiv EEG device is sensitive enough to detect changes in brain activation patterns during human gait. Further research is required before definite conclusions can be made about this novel device, or about what effects, if any, orthotics have on brain activation patterns during gait.

Keywords: human gait, neuroimaging, electroencephalogram (EEG), Emotiv EPOC neuroheadset, orthotic devices
Co-Authorship Statement

This thesis is the primary work of Lindsay Carey. The creation of the research topic and design of the intervention was created by Dr. Colin Dombroski through collaboration with Dr. Jeff Holmes and Dr. Andrew Johnson. The moulding and fabrication of the orthotic devices were completed by Dr. Dombroski and SoleScience Laboratories. The process of data collection was facilitated by both Dr. Dombroski and Lindsay Carey. The raw EEG data produced from the Emotiv neuroheadset was processed and analyzed through the use of EEG lab within MAT lab by James Desjardins from Brock University. I wrote the original draft of this thesis in its entirety, including the interpretation of the statistical results (with the assistance of James Desjardins). The drafts were then sent to Dr. Dombroski and Dr. Holmes for their comments and suggestions for critical revision of this thesis.
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I would like to acknowledge James Desjardins for his expertise and contributions involving the processing and analysis of the EEG data.

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I would like to express my gratitude towards Dr. Lisa Fischer, for her incredible support and advice during the last two years.

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List of Abbreviations

99mTc- HM-PAO - Technetium-99m-hexamethyl-propyleneamine oxime

AL - anterolateral system

b1 – first section of baseline continuous gait

b2 – second section of baseline continuous gait

BCI – brain computer interface

BOLD - blood oxygen level-dependent signal

C.D. – primary investigator Colin Dombroski

CNS – central nervous system

CPG - central pattern generator

CT – computed tomography scan

DCML - dorsal-column medial-lemniscus system

EEG – electroencephalography

EMG – electromyogram

ERP – event related potentials

fMRI – functional magnetic resonance imaging

fNIRS – functional near infrared spectroscopy

g1 – gait initiation

g2 – turning events during gait

g3 – gait termination
H2O-15 – oxygen-15 labelled water

‘high’ – smaller amplitude rhythmic oscillations (resembling stationary EEG)

Hz – hertz

ICA - independent components analysis

‘low’ – larger amplitude rhythmic oscillations (resembling movement artifacts)

MRI – magnetic resonance imaging

NIRS – near infrared spectroscopy

o1 – first section of orthotic continuous gait

o2 – second section of orthotic continuous gait

PET – positron emission tomography scan

PMC – primary motor cortex

PSC – primary somatosensory cortex

PSD – power spectral density

rCBF – regional cerebral blood flow

SMA – supplementary motor area

SPECT – single photon emission computed tomography

TMS – transcranial magnetic stimulation
Chapter 1

1 Introduction

Human gait is a complex task that requires the coordination of multiple systems in the body, most importantly, the musculoskeletal and nervous systems. The bulk of the current knowledge pertaining to the neural mechanism of walking has been derived from studies on the nervous system of animals (Whittle, 2007). It is, however, unclear to what extent these findings may be applied to the control of human gait. Historically, human gait has been considered an automatic process involving little or no higher cortical input (Wang et al., 2009). It has been suggested that human gait is controlled by central pattern generators (CPG’s) in the spinal cord that produce the repetitive oscillation movements of the lower limbs. Armstrong (1988), however, suggested that supraspinal inputs play a major role not only in initiating locomotion, but also in adapting the locomotor pattern to environmental and motivational conditions. Through advancements in technology, recent studies have demonstrated that human gait involves much more than just CPG’s and the spinal cord. For example, it is speculated that many areas are involved in this complex process, including the cerebral cortex, basal ganglia, cerebellum (Fukuyama et al., 1997), and supraspinal centers in the brainstem (Jahn et al., 2004). Furthermore, there is important information relayed from peripheral sensory receptors, as well as integrated information from the visual, vestibular and auditory senses (Rossignol et al., 2006). Although this research has contributed to a greater comprehension of the complexities involved with gait, an understanding of the neuronal circuits involved in the process of human gait is still largely unknown and requires further investigation.

The brain-scanning techniques that allow researchers to observe the levels of activity in an individual’s brain are known as neuroimaging. Neuroimaging techniques have facilitated a greater understanding of the human brain. Prior to neuroimaging, the only way to learn about the human brain was by cataloguing a set of impairments that a group of patients would exhibit when they suffered damage to a specific brain region (Sadava, Heller, Orias, Purves & Hillis, 2008). Over the years there have been developments in non-invasive brain imaging techniques that have allowed researchers to investigate neural control during walking in humans. These non-invasive imaging studies
have used either functional magnetic resonance imaging (fMRI) or positron emission tomography (PET) scans, both of which have associated limitations. The most important limitation being that a person is incapable of performing walking activities due to the physical restrictions with the technology (ie. individuals must lie down inside a confined space) (Wang et al., 2009).

To overcome this shortcoming, some research has used repetitive foot movements as a surrogate to gait (Miyai et al. 2001). For example, while inside an fMRI machine, a participant would perform alternating foot flexion-extension movements as recording of the brain activity occurs. Similarly, other research has addressed this issue by adopting procedures that allow for the recording of cerebral activity while performing imaginary gait tasks using motor imagery (mental simulation of gait without actual execution, or watching a video of another person walking) and planning (recording just prior to gait initiation, after receiving an external cue to start walking) (Jahn et al., 2004; Wang et al., 2008; Hanakawa, 2006; Iseki et al., 2008). Additionally, these techniques do not take into account the simple task of un-supported upright stance, the effects of gravity on the body, balance control, or the coordination of multiple limbs and joints. The complex task of integrating vestibular, visual and somatosensory signals while moving in the environment is also not considered.

To address these limitations, the usage of near-infrared spectroscopic (NIRS) imaging and single photon emission computed tomography (SPECT) have been investigated as they allow for the acquisition of data during dynamic activities such as gait. NIRS allows for visualization of cortical activities during dynamic movements, by detecting changes in oxygenated hemoglobin, deoxygenated hemoglobin and total hemoglobin within a few centimeters’ depth of the skull surface (Mayai et al., 2011). Mayai et al. (2011) used this NIRS headpiece to investigate neuronal activity while participants completed gait tasks on a treadmill. The obtained NIRS topographic maps were compared to the MRI anatomical images. Similarly, knowing that changes in regional cerebral blood flow (rCBF) reflect changes in underlying neuronal activity, Fukuyama et al. (1997) used SPECT technology to detect changes in neuronal activity during gait. This was achieved by intravenously injecting a radioactively labelled
substance (technetium-99m-hexamethyl-propyleneamine oxime) during locomotion, which is rapidly distributed through the brain in proportion to rCBF, allowing for cerebral activity to be detected. Both NIRS imaging and SPECT technology have allowed for further advancements in the understanding of neural control during human gait. However, SPECT technology involves an invasive intravenous injection and NIRS imaging usually needs to be compared with MRI anatomical imaging afterwards.

In this study, we used a non-invasive, functional brain imaging device (i.e. Emotiv), which has yet to be explored in the realm of neuronal activity in human gait. This device has been tested and has been proven to be a reliable and valid tool when compared to other similar systems (Badcock et al., 2013). It involves a revolutionary wireless neuroheadset, which serves as a portable electroencephalogram (EEG) device. Through electrodes placed on the scalp, an EEG detects changes in electrical potential differences between electrodes, where these differences reflect the electrical activity of the neurons in the brain regions under the electrodes (Sadava et al., 2008). This wireless neuroheadset allows for the collection of real time electrical brain activity while the participant walks in a live setting. This technology from Emotiv™ allows researchers to gain direct information concerning brain activity during human gait. It does not only relay the neuronal activation patterns produced during gait, but also gives insight into the activation patterns that result from things like somatosensory input, visual stimulus, and balance control.

Many structures within the central nervous system contribute to the development of the motor neuron signals that activate muscles. The three main levels of control include the spinal cord, the descending systems of the brainstem, and the motor areas of the cerebral cortex (Enoka, 2002). The spinal cord and brainstem mediate reflexes and automatic behaviours, while the cortical motor areas initiate and control more of the complex voluntary movements (Enoka, 2002). It is the ascending pathways, however, that deliver somatosensory information from the arms, legs and trunk through mechanoreceptors, thermoreceptors, nociceptors and chemoreceptors (Enoka, 2002). The somatosensory input from mechanoreceptors in the skin, muscles and joints, as well as visual input and information from the organs of balance, all play an important role in
human gait (Pinel, 2009). Currently, it is unclear in the literature what effect, if any, that custom made foot orthoses have in the brain. The human neuromuscular system acts as an integrated structure using control mechanisms and feedback systems (Nurse & Nigg, 1999). Although, the response of the feet to shoes or inserts is not well understood, it has been theorized that sensory feedback from the foot plays a role in the kinematic and/or kinetic response and the effects of inserts and orthotics need further investigation (Nurse & Nigg, 1999).

In this study we will be investigating possible changes in brain activation patterns during each trial, using the Emotiv wireless EEG device. To our knowledge, there has been no published research using the Emotiv neuroheadset to record brain activation patterns during human gait. The purpose of the current study is to examine the feasibility of this novel approach and technique of recording brain activity, and to determine if custom made orthotics will alter the brain activity patterns recorded by the Emotiv EEG during human gait.
Chapter 2

2 Literature Review

2.1 Introduction

Human locomotion is a complex task that involves interactions between the brain, spinal cord, peripheral nerves, muscles, bones and joints (Whittle, 2007). The ability to coordinate limb segments is essential in everyday activities, but very little is known about the neural basis of coordination in humans. Understanding how the seemingly simple and automated movements of walking and running are controlled, forms a main challenge for modern neuroscience (Duysens et al., 1998). Most of the existing knowledge about the cerebral control of gait in humans comes from studies that were completed in cats and rodents (Armstrong, 1986; Rossignol et al., 2006). It is unclear whether the organization and knowledge from the animal models can be inferred to human locomotion. Central pattern generators (CPG) have been widely studied in animal models, but human locomotion differs from most mammals in that it is bipedal and assumed to be under more supraspinal control (Jahn et al., 2008). There have been numerous studies conducted using a variety of neuroimaging techniques to investigate the cerebral activity that occurs during different human activities and movements. However, insufficient knowledge has been gained, largely due to the technical inability to perform walking tasks inside the machinery (Hanakawa, 2006). New techniques are being investigated to overcome these problems and allow for better imaging and a greater understanding of the human cerebral cortex.

Locomotion also incorporates visual information from the eyes, somatosensory information from the feet, knees and hip, as well as vestibular information about balance from the inner ears (Pinel, 2009). Although visual and vestibular feedback is essential, it has been proposed that the sensory feedback control system plays a large role in human locomotion (Newmann, 1980; Robbins et al., 1988). New methods have been employed to enhance the amount of sensory feedback, as sensory information is used to adapt motor output appropriately for environmental conditions (Lundy-Ekman, 2013). The sole of the foot is continuously in contact with the physical environment during locomotion. This
provides valuable information about the changing surface characteristics to the central nervous system (Hohne et al., 2012), as well as afferent feedback for timing and the transitions between stance and swing (Van Wezel et al., 1997; Lundy-Ekman, 2013). Foot orthoses are one type of method commonly prescribed to help with foot and ankle problems. Orthoses can be administered for a variety of reasons, including shock absorption, reducing shear forces on the foot, improving sensory feedback, or even improving balance (May & Lockard, 2011). The use of orthotic devices in the shoe will be explored to determine if sensory feedback can be enhanced and therefore alter human gait.

2.2 Anatomy of the Brain and Central Nervous System

The human central nervous system (CNS) consists of the brain, spinal cord and the peripheral nerves. Globally, the brain can be subdivided into four main areas; the cerebrum, the cerebellum, the brain stem and the diencephalon (Nieuwenhuys, Huijzen, & Voogd, 2008). The cerebrum consists of four lobes; the frontal, parietal, occipital and temporal lobes. Included in the brainstem area are the midbrain, pons and medulla oblongata. Forming the center core of the brain, the diencephalon incorporates the epithalamus, thalamus and hypothalamus (Sadava et al., 2008). The human brain contains, at least, one hundred billion neurons. Each neuron represents a complex structure that processes incoming information in many ways, receives inputs from other neurons, and has numerous output connections or synapses (Latash, 2008). Voluntary movement is processed in a top-down hierarchy, from brain to spinal cord to muscle (peripheral nerves), known as efferent output (Lundy-Ekman, 2013). Whereas, afferent input is in the reverse order, with nerve signals originating in the peripheral nerves, travelling through the spinal cord and projecting onto different areas of the brain (Lundy-Ekman, 2013).

2.2.1 Cerebrum

The surface of the two cerebral hemispheres is composed of grey matter which is called the cerebral cortex. This is the part of the brain that is traditionally associated with higher nervous activity; including perceiving and interpreting sensory, motor and
memory information, making conscious decisions, controlling voluntary movements, language, non-verbal communication, intelligence and personality (Latash, 2008; Lundy-Ekman, 2013). Underneath the cerebral cortex is white matter, made up of axons that connect the cell bodies in the cortex with one another and with other areas of the brain (Sadava et al., 2008). As previously mentioned, the cerebrum consists of four lobes, each composed of distinct areas where specific processing takes place.

2.2.1.1 *Frontal Lobe*

The frontal lobe is generally associated with actions involving thinking, planning, central executive function and motor execution. The main regions of the frontal lobe include the Primary Motor Cortex (Brodmann area 4), Secondary Motor Cortex (Brodmann area 6) and Broca’s Area (Brodmann area 44 & 45). The Primary Motor Cortex (PMC) is located in the posterior portion of the frontal lobe, directly anterior to the central sulcus. The PMC controls movement for specific parts of the body, where stimulation of a certain area will result in the movement of that associated body part (Sadava et al., 2008). This region is somatotopically organized, meaning the parts of the body are organized in a specific way (motor homunculus) and are not proportional to their body size (Sadava et al., 2008). For example, parts of the body with fine motor control, like the hands and face, have a disproportionate representation. The PMC works together with other motor areas and several subcortical regions to plan and execute movements. It receives somatosensory information relayed by the thalamus and the primary somatosensory cortex, as well as motor instructions from the motor planning areas (Lundy-Ekman, 2013). It is also the major point of departure for sensorimotor signals from the cerebral cortex (Pinel, 2009).

The Secondary Motor Cortex is comprised of two large areas, the Supplementary Motor Area (SMA) and the Premotor Cortex, and is located just anterior to the PMC. The SMA is important for initiation of movement, orientation of the eyes and head, and planning bimanual and sequential complex movements (Lundy-Ekman, 2013). The Premotor Cortex is involved in more complex patterns of movement than the discrete patterns generated in the PMC (Sadava et al., 2008). It helps control the trunk and girdle muscles (the muscles closest to the body’s main axis), like the shoulders during upper
limb tasks and the hips during walking (Lundy-Ekman, 2013). It is believed that the premotor cortex is also involved in the planning or programming of voluntary movements (Purves, Augustine, & Fitzpatrick, 2001).

A special region located in the posterior-lateral pre-frontal cortex and partly in the premotor area, called Broca’s Area, provides the neural circuitry for word formation (Sadava et al., 2008). This area is responsible for planning the movements of the mouth during speech. Also, motor patterns for expressing individual words or even short phrases are initiated and executed here (Sadava et al., 2008; Lundy-Ekman, 2013). Additionally, Broca’s area works in tight conjunction with Wernicke’s language comprehension area that is located in the temporal region of the cerebrum (Sadava et al., 2008).

2.2.1.2 Parietal Lobe

The parietal lobe is associated with somatosensory perception, integrating sensory information from numerous parts of the body and the integration of visual and somatospatial information. This lobe is located posterior to the frontal lobe and superior to the occipital lobe. The major regions of the parietal lobe consist of the Primary Somatosensory Cortex (Brodmann areas 3, 1 & 2), Secondary Somatosensory Cortex (Brodmann areas 40 & 43), Somatosensory Association Cortex (Brodmann areas 5 & 7) and the Parieto-Occipitotemporal Association Area. The Primary Somatosensory Cortex (PSC) is located in the anterior portion of the parietal lobe, just behind the central sulcus. The PSC is the most prominent region in the parietal lobe and is the main sensory receptive area for the sense of touch and proprioception. Touch and pressure information by neurons that are relayed from the body through the thalamus are received by the PSC. These neurons can identify the location of the stimuli as well as discriminate among various shapes, sizes and textures of objects (Sadava et al., 2008; Lundy-Ekman, 2013). The PSC is also somatotopically organized, so it is organized according to a map of the body surface (somatosensory homunculus), where the lips and hands are the most prominent. The somatosensory information ascends from each side of the body to the PSC through two major pathways; the dorsal-column medial-lemniscus system and the anterolateral system (Pinel, 2009). The dorsal-column medial-lemniscus pathway tends to
carry information about touch and proprioception, whereas the anterolateral system carries information about pain and temperature.

The Secondary Somatosensory Cortex is located on the lateral edge of the parietal lobe, just superior to the temporal lobe. This area analyzes sensory input from both the thalamus and the PSC and integrates tactile and proprioceptive information obtained from manipulating an object (Lundy-Ekman, 2013). Neurons in this area provide stereognosis by comparing somatosensation from the current object with memories of other objects (Lundy-Ekman, 2013). The Somatosensory Association Cortex (posterior parietal cortex) is situated posterior to the PSC and just superior to the occipital lobe. This cortex is involved in locating objects in space and plays a role in visuomotor coordination. It is a point of convergence between visual and proprioceptive information, to determine where an object is in relation to parts of the body (Sadava et al., 2008). The nervous system must know the original position in space before a body part can be moved and the positions of the external objects that the body will be interacting with. To accomplish the aforementioned task, the posterior parietal cortex receives information from the visual system, auditory system and the somatosensory system, and most of its output goes to the areas of the motor cortex (Pinel, 2009).

The Parieto-Occipitotempral Association Area provides high levels of interpretative meaning for signals from all of the surrounding sensory areas. Areas located in the posterior portion of the parietal lobe, extending into the superior occipital lobe, provide continuous analysis of the spatial coordinates of all parts of the body as well as of the surroundings of the body (Sadava et al., 2008). This area receives visual sensory information from the posterior occipital cortex and simultaneous somatosensory information from the anterior parietal lobe, allowing it to compute the coordinates of the visual, auditory and body surroundings (Sadava et al., 2008).

2.2.1.3 Temporal Lobe

The temporal lobes are located on the lateral sides of the human brain, just above the ears and have a variety of sensory functions. These lobes are involved in auditory perception, the organization of sensory input, language and speech production and
comprehension, as well as memory formation and emotional responses. The Primary and Secondary Auditory Cortices are located in these lobes, where sound waves, pitch and frequencies are organized and processed. Wernicke’s Area, spanning the region between the temporal and parietal lobes, works together with the auditory cortex and has a key role in speech and language comprehension (Lundy-Ekman, 2013). The temporal lobe also houses the structures of the limbic system, which includes the olfactory cortex, amygdala and the hippocampus. The olfactory cortex is responsible for the detection of scents. The amygdala is involved in both emotion and memory and the hippocampus is linked to memory storage and function (Sadava et al., 2008). Lastly, the Parietotemporal Association cortex incorporates spatial coordination for constructing images of one’s own body and for planning movements, as well as integrating information involved in the recognition, identification and naming of objects (Sadava et al., 2008).

2.2.1.4 Occipital Lobe

The last of the four lobes forming the cerebral cortex is the occipital lobe, which is situated in the most posterior portion of the cerebrum. The occipital lobe is the main center for receiving and processing visual information. Located within this lobe is the Primary Visual Cortex, where there is a visual cortex in each hemisphere of the brain. This region is highly specialized, as it has different groups of neurons that separately encode for colour, orientation and motion information (Pinel, 2009). The visual cortex receives information and projections from the retina and is then able to process and interpret these signals. In the anterolateral region of the occipital lobe lays the visual association area which relays visual information from words read into Wernicke’s area, the language comprehension area (Sadava et al., 2008). The association area is essential for making sense of the visual world and translating visual experience into language (Sadava et al., 2008). Facial recognition is another important aspect involving the ability to process visual information and this area is located on the medial underside of both occipital lobes.
2.2.2  *Cerebellum*

The cerebellum is located at the back of the brain, tucked underneath the occipital and temporal lobes of the cerebral cortex and behind the portion of the brainstem called the pons. The cerebellum contains three lobes; the flocculonodular lobe, anterior lobe and posterior lobe. However, areas of the cerebellum are also described corresponding to their functional subdivisions, which include the vestibulocerebellum, spinocerebellum (medial sector) and cerebrocerebellum (lateral sector). The smallest region, the flocculonodular lobe, often referred to as the vestibulocerebellum, is concerned with vestibular reflexes and in balance and spatial orientation (Byrne, 2007). The medial sector of the anterior and posterior lobes comprises the spinocerebellum, which functions mainly in the fine-tuning of body and limb movements as a result of the integration of sensory input. This sector receives proprioception input from the dorsal columns of the spinal cord (mainly the spinocerebellar tract), and its output projects to the deep cerebellar nuclei, as well as the cerebral cortex and brain stem (Byrne, 2007). The lateral sector and also the largest region, the cerebrocerebellum, receives input exclusively from the cerebral cortex via the pontine nuclei and sends output mainly to the venrolateral thalamus (Byrne, 2007). This sector is thought to be involved in the planning and timing of movements as well as cognitive functions of the cerebellum.

The cerebellum plays a large role in motor activity, however, there are many theories as to what these roles actually entail. The cerebellum does not initiate movement, but it contributes to the coordination, precision and accurate timing of movements (Byrne, 2007). It has been described as a timing device that ensures the correct order and timing of individual muscle action, as well as the timing in rapid, smooth progression from one muscle movement to the next (Latash, 2008; Pinel, 2009). This area is believed to play a major role in motor learning, by acquiring and memorizing new skills, but particularly in the learning of sequences of movements, which is a critical factor in the timing and precision of movements (Pinel, 2009). The cerebellum has also been referred to as a coordination device, putting together components of complex multi-joint or multi-limb movements, in addition to controlling balance in complex movements such as walking or running (Latash, 2008). Continuous sensory feedback information is received
by the cerebellum, allowing for it to monitor and make corrective adjustments in the body’s motor activities while they are being executed (Sadava et al., 2008). This feedback allows the cerebellum to compare the actual movements depicted by the peripheral sensory feedback system with the movements that were intended by the motor system (Sadava et al., 2008). If the sensory feedback and intended movements do not compare, then there are instantaneous subconscious corrective signals transmitted back into the motor system which will either increase or decrease the levels of muscle activation.

2.2.3 Brain Stem and Spinal Cord

The brain stem, consisting of the pons, medulla oblongata and midbrain (mesencephalon), is the connection between the cerebrum and the spinal cord. The brain stem controls several important functions in the body including the regulation of cardiac and respiratory functions, attention, arousal and maintaining consciousness (Lundy-Ekman, 2013). One of the brain stem’s main functions is relaying information between the peripheral nerves and spinal cord to the upper parts of the brain. Both motor and sensory neurons travel through the brainstem allowing for the relay of signals between the spinal cord and brain. Motor and sensory information is relayed via the corticospinal tract (motor), posterior column-medial lemniscus pathway (fine touch, vibration, sensation and proprioception), as well as the spinothalamic tract (pain, temperature, crude touch) (Pinel, 2009). Also, distributed in the brain stem are at least three motor centers that send efferent fibers to influence the motor neurons of the spinal cord, consisting of the red nucleus, lateral vestibular nucleus and the reticular formation (Enoka, 2002).

The spinal cord has two main functions; to convey information between neurons innervating peripheral structures and the brain, and to process information (Lundy-Ekman, 2013). The spinal cord receives somatosensory feedback information from peripheral structures, which allows it to generate reflex loops, like movement of a limb away from a painful stimulus. Therefore, the spinal cord contains neuronal networks that can produce reflexes and automatic behaviours independently of input from the brain stem and the cerebral cortex (Enoka, 2002).
2.3 Central Pattern Generators

For many species, the cyclical patterns needed for walking, respiration, mastication or other rhythmical activities are generated by neural networks (Duysens et al., 1998). However, for locomotion, one usually refers to the term central pattern generator (CPG) to indicate a set of neurons responsible for creating a motor pattern, where this pattern involves the alternating activity in groups of flexors and extensors (Grillner & Wallen, 1985). CPGs are adaptable networks of spinal interneurons that activate lower motor neurons to elicit alternating flexion and extension movements of the hip, knees and ankles, with each limb having a dedicated CPG (Lundy-Ekman, 2013). Human gait has been considered in past decades only as an automatic process involving little or no higher cortical input (Wang et al., 2009). However, the majority of current knowledge pertaining to the neural mechanism of human gait has been obtained from studying the nervous system of animals, in particular cats and rodents. There has been an abundance of data collected on animals, leading to the general assumption that CPGs are the underlying factors in the central control of locomotion. Unfortunately, little is still known about the spinal networks acting like CPGs in humans.

Originally in 1910, Sherrington suggested that walking could be produced entirely by a series of reflexes. However, further studies have provided evidence for neural networks in the spinal cord, now termed central pattern generators, that are capable of governing locomotion (Grillner & Wallen, 1985). It was in 1911 that Thomas Brown discovered the existence of what he called ‘half centers’ in the spinal cord. These half centers were mutually inhibiting mechanisms which ensured that in one limb the flexor motoneurons are excited first, while the extensor motoneurons are inhibited (Brown, 1911). The pattern was then reversed, so that the limb could go through the flexion-extension cycle. This work was further explored in the nervous system of cats and rodents. When limbs of a decerebrate cat were placed on a treadmill, movement of the treadmill at a constant speed induced locomotor-like stepping in the limbs of the cat (Latash, 2008). It can be concluded from this type of observation that CPGs for locomotion exist in the spinal cord of mammals and produce locomotor-like activity in response to both descending stimulation and peripheral input. Grillner (1981) also
confirmed the existence of spinal networks with the capacity of generating basic locomotor rhythms in the absence of any supraspinal or sensory input to the spinal cord in a variety of lower mammals.

Following the confirmation of the existence of central pattern generators in the spinal cord, the theory of higher cortical input was then tested. It was discovered by Rossignol (1996) that, after the transection of their spinal cords, most cats were not able to generate locomotor movements. This insinuated that commands for the initiation of locomotor activity must be given at some level in the central nervous system above that of the spinal lesion. Therefore, by varying the level of transection in the spinal cord, it was shown that the regions for initiation of locomotion were located in the brain stem (Rossignol, 1996). Similar findings occurred when a decerebrated cat’s brain stem structures were electrically stimulated. A corresponding descending input to the lumbar cord was elicited, suggesting that the spinal rhythm-generating network is likely activated by higher structures (Nielson, 2003).

The experiments completed in cats and other mammals demonstrated that the spinal cord contributes substantially to the control of complex motor functions such as locomotion (Guertin, 2009). Even though the control of locomotion has been mainly studied in animal models, there has been inferential evidence of the existence of a CPG for locomotion in humans. Most of the testimonies to the existence of locomotor CPGs in humans come from studies completed on individuals with clinically complete spinal cord section and individuals who are gait retraining following a spinal cord injury (Duysens et al., 1998; MacKay-Lyons, 2002). It has been concluded that there is sufficient evidence to suggest that the human spinal cord, with intact sensory inputs, is capable of generating rhythmic motor bursts with assisted leg movements by therapists and a partial body weight bearing support harness (Stewart et al., 1991; Wernig & Muller, 1992; Dietz et al., 1995; Harkema et al., 1997). Observations in these spinal cord injury patients have lead to the belief that spinal locomotor generators are located in the lower thoracic-upper lumbar level of the human spinal cord (Latash, 2008). Through the use of advancing neuroimaging techniques, supraspinal locomotor networks have been identified in humans that include the frontal cortex, basal ganglia, brain stem locomotor nuclei and the
cerebellum. This circuitry regulates the initiation and termination of gait, changing the direction and velocity of gait and spatial orientation and navigation around obstacles (Jahn et al., 2008; Bakker et al., 2007b; Wagner et al., 2008).

2.4 Gait

Human walking is a complex task that requires the coordination of a number of different muscles acting on the hip, knee, ankle and foot in order to advance the body in a desired line of progression (Nielson, 2003; Hsu, Michael, Fisk, & AAOS, 2008). Walking utilizes a repetitive and rhythmical sequence of limb motions to simultaneously move the body forward while also maintaining stability of the body mass (Perry & Burnfield, 2010). Each limb combines the patterns of motion, passive force, and muscular control into a sequence of activity, called the gait cycle, which includes all of the body’s activity from the time one foot strikes the floor until the same foot strikes the floor again (Edelstein & Moroz, 2011; Hsu et al., 2008). There are a number of actions and events that occur in the complex mechanism of gait. This mechanism is further complicated with the introduction of environmental influences (obstacles, people, changing walking surfaces, etc).

The gait cycle is commonly defined as the time interval between two successive occurrences of one of the repetitive events of walking (Whittle, 2007). For instance, the cycle would start with the initial heel contact of the right foot and would end with the heel contact of the same right foot. There are two phases of the gait cycle, the stance phase and the swing phase. However, these phases are not distributed evenly throughout the cycle. It has been determined that the stance phase lasts for about sixty percent of the cycle, and the swing phase lasts for the remaining forty percent (Whittle, 2007; Perry & Burnfield, 2010). Edelstein and Moroz (2011) defined the stance phase as the period during which the reference foot is in contact with the ground. The swing phase was described as the period during which no part of the reference foot is in contact with the ground, or simply, when the foot is in the air (Edelstein & Moroz, 2011). Therefore, stance begins with initial heel contact (heel strike) and swing begins as the foot is lifted from the floor during toe off. Perry and Burnfield (2010) have reported that the two gait cycles could be further broken down into eight components. These include initial contact
(heel contact/strike), loading response, mid-stance, terminal stance, pre-swing (toe off), initial swing, mid-swing and terminal swing. This pattern of gait results from the complex interactions between many neuromuscular and structural elements of the locomotor system (Whittle, 2007). The initiation of gait serves as a transitional process from the balanced upright standing position to the beginning of steady-state walking, whereas the termination of gait requires the deceleration of the forward momentum of the body to return to a stable stance position (Jian et al., 1993; Patla, 2004).

As the body moves forward, one limb serves as a dynamic source of support while the other limb advances itself to the new support site, and then the limbs reverse roles (Perry & Burnfield, 2010). Kiehn (2006) explained the key features involved in the gait process consisted of rhythm, ipsilateral coordination of flexors and extensors across the same or different joints in a limb, and left-right coordination. In an ideal situation, gait appears to be coordinated, efficient and somewhat effortless resulting in the conservation of energy expenditure. Each stride in the gait cycle involves an ever-changing alignment between the body and the supporting foot during stance and limb during swing (Perry & Burnfield, 2010). Therefore, in the absence of disease or trauma, it is important to maintain proper lower limb biomechanics. This will minimize energy expenditure and reduce the stress on the bones, joints and soft tissues of the lower extremities (Hsu et al., 2008). However, in the event of disease or trauma, gait patterns can become less optimal and, in turn, result in excessive amounts of expended energy. This alteration of gait could result in a disruption to the precision, coordination, speed and versatility of the regular pattern, causing an individual to alter the motion of adjacent joints and controlling muscles, thereby increasing the energy cost of walking (Perry & Burnfield, 2010; Hsu et al., 2008).

2.5 Sensory, Motor and Proprioceptive Feedback

During gait, interaction with the environment is multisensory. The senses are classified as exteroceptive and involve the continual integration of sensory input from visual, vestibular and somatosensory receptors by the central nervous system to assess the position and motion of the body (McGlone et al., 2007; Johansson & Magnusson 1991). The CNS uses this sensory feedback information to make corrections to ongoing
movements, future movements, to avoid obstacles and for placement of the feet. Somatosensory feedback is derived from the sensory input of proprioceptors and mechanoreceptors. Van Wezel et al. (2000) stated that somatosensory feedback from muscles, joints and skin is essential in the normal execution of human gait.

The sensory feedback information is integrated into the motor commands at all levels of the CNS. The CNS makes use of this sensory information in two ways: the sensory activity helps internal commands in the driving of output neurons as a part of all normal voluntary movements; and the sensory information may be used to inform the CNS about errors in the execution of a movement (Nielson & Sinkjaer, 2002). The sensory information produced from proprioceptors provide details about limb position and muscle forces and are used to monitor and control limb movements. Sensory feedback, produced by the skin (cutaneous receptors), can provide information about touch, temperature, itch and pain. The CNS relies on sensory input from muscles, joints, and cutaneous receptors in the lower extremities to generate effective motor patterns for human posture and locomotion (Nurse & Nigg, 2001). This is made possible by the constant information it is provided regarding muscle and joint loading, joint kinematics, and pressure distributions on the plantar surface of the foot (Nurse & Nigg, 2001). Nielson and Sinkjaer (2002) suggested that less central input to the motoneurons is necessary when there is sensory feedback present, than if it was absent. Thus, declaring that less central drive is necessary in order to activate the motorneurons with the presence of sensory afferent feedback.

2.5.1 Receptors

Receptors are specialized cells or subcellular structures that change their properties in response to a specific stimulus in order to make information about that particular stimulus available to other neurons within the central nervous system (Latash, 2008). Cutaneous receptors are one of four different groups of proprioceptors found in the body, which involve specialized receptors in the skin that are sensitive to different sensory modalities. Somatosensory information includes afferent signals from mechanoreceptors, thermoreceptors, nociceptors and chemoreceptors. Mechanoreceptors respond to mechanical deformation of the receptor by touch, pressure, stretch or
vibration; thermoreceptors are sensitive to temperature; nociceptors are sensitive to potentially damaging stimuli (pain); and chemoreceptors are sensitive to chemical stimuli (Lundy-Ekman, 2013). The main focus here will be on the mechanoreceptors, of which there are four different types located in the skin; Pacinian Corpuscles, Merkel Disks, Ruffini Endings and Meissner Corpuscles. Pacinian Corpuscles respond to the displacement or mechanical deformation (vibration) of the skin; Merkel Disks respond to gradual skin indentation or vertical pressure on the skin; Ruffini Endings respond to gradual skin stretch and Meissner Corpuscles are sensitive to quickly changing pressure on a small area of skin (Pinel, 2009). These receptors are also located in different layers of the epidermis and dermis. Meissner Corpuscles and Merkel Disks are located close to the skin's surface, whereas Ruffini endings and Pacinian Corpuscles are deep in the dermis (Latash, 2008). These different types of mechanoreceptors can either be slow or fast adapting. Patel et al. (2011) suggested that slowly adapting mechanoreceptors provide information about how the pressures are spatially and sequentially distributed on the skin-surface interaction. Whereas the rapidly adapting receptors provide information about pressure amplitude and changes of pressure exerted on the skin (Patel et al., 2011).

Of the four classical cutaneous modalities of the somatosensory system (touch, temperature, pain and itch), it is the discriminative touch that subserves the perception of pressure, vibration, slip and texture (McGlone et al., 2007). All of which involve critical information to be relayed to the central nervous system throughout human locomotion.

2.5.2 Signal Pathways

The speed of information transmission within the central nervous system is very rapid with signals being transmitted at tens of meters per second (Latash, 2008). Each system, sensory and motor, uses information carried by a number of anatomically distinct pathways, usually containing synaptic relays so that moving information can be processed and integrated (Latash, 2008). A peripheral nerve contains both afferent and efferent axons. The afferent axons carry information from peripheral receptors towards the CNS (sensory) and efferent axons carry information away from the CNS (motor) (Lundy-Ekman, 2013). Primary afferents are the first link between skin, muscles or joints and the spinal column. These sensory signals enter the spinal cord almost entirely through
the sensory (posterior) roots (Guyton & Hall, 2006). From here, information is relayed through the ascending white matter spinal tracts. Peripheral receptors convey sensory signals associated with the sense of touch, pressure, proprioception (stretch of muscles, tension on tendons, position of joints) and vibration, towards the supraspinal structures (Guertin, 2012).

Somatosensory information ascends from each side of the body to the brain by means of two major pathways; the Dorsal-column medial-lemniscus system (DCML), and the anterolateral system (AL). The DCML system tends to carry information about touch and proprioception. The sensory neurons of this system enter the spinal cord via the dorsal root, ascend in the dorsal columns and synapse in the dorsal column nuclei of the medulla (Pinel, 2009). The axons of the dorsal column nuclei cross over to the other side of the brain ascending then to the ventral posterior nucleus of the thalamus, where they will be projected to the primary and secondary somatosensory cortex or the posterior parietal cortex (Pinel, 2009). The AL system tends to carry information about pain and temperature, and is comprised of three different tracts which project to different areas in the thalamus (Pinel, 2009). This sensory feedback is an integral part of the overall motor control system and is critical in modifying CPG generated motor programs in order to facilitate constant adaptations to the environment (MacKay-Lyons, 2002). Pearson (1993) identified three potential roles for somatosensory afferent feedback in the production of rhythmic movements; it reinforces CPG activity, assists with the function of timing, and facilitates phase transitions in rhythmic movements.

2.5.3 Plantar Surface of the Foot

Throughout a healthy gait cycle, the sole of the foot is continuously in contact with the environment. This enables the foot to provide information about changing surface characteristics and serve as a tool for continual feedback to the central nervous system. The foot cushions the musculoskeletal system during impact, supports the body during ground contact, transmits force, keeps the body in balance and serves as a system for sensory input (Mulder & Hulstijn, 1985; Newman, 1980; Robbins et al., 1988). During the gait cycle there are large amounts of afferent input generated both from the skin, in contact with the ground, and from the skin being stretched by the movements of
the limb (Duysens et al., 1995). It would seem that there is an abundance of somatosensory information generated from the plantar mechanoreceptors. However, according to Kennedy and Inglis (2002), there is limited information and knowledge about the characteristics of the mechanoreceptors specific to the foot sole. There have been many studies conducted concerning the cutaneous sensation of the hand or reports from other skin regions that are often used to predict the properties of cutaneous mechanoreceptors in the foot sole. However, attempting to transfer the properties of skin receptors from other body regions to the foot may prove to be inappropriate, as there are differences between glabrous and hairy skin types (Kennedy & Inglis, 2002).

Numerous studies have examined how the foot and sensory system reacts to changes in shoe soles, insoles, or different surfaces. Additionally, effects from vibration and varying pressure exertion have been studied. In 1981, Watanabe and Okubo provided evidence that standing on different surfaces can alter the transmission of afferent signals from the plantar surface of the foot. In a study conducted by Patel et al. (2011) foam surfaces were used to test the relative contributions of the visual, vestibular and somatosensory inputs. Standing on a foam surface was shown to challenge postural control by decreasing the reliability of sensory information from the plantar mechanoreceptors. In order not to involve the other two inputs, participants were instructed to keep their eyes open and make sure their head was facing forwards (Patel et al., 2011). A study was carried out by Nurse and Nigg (1999) to quantify the relationship between the pressure and vibration sensitivity of the sole of the foot with plantar pressure distribution during walking and running. The researchers’ results suggested that sensory feedback from the foot plays a role in the kinematic and/or kinetic response while walking and running. Zehr et al. (1997) completed a study investigating the effects of electrical stimuli on superficial peroneals and tibial nerves during human gait. The superficial peroneals and anterior tibialis nerves were stimulated and the stimuli were reported as being sensed on the plantar surface of the foot. It was concluded that afferents from various skin sites on the foot play a crucial role in the timing of the transition between stance and swing in gait (Zehr et al., 1997). It appears that plantar cutaneous sensation plays an important role in balance and stability, kinematics and kinetics, and the timing of transitions in gait. Perry et al. (2000) also suggested that plantar
mechanoreceptors can provide detailed spatial and temporal information about contact pressures in the foot, and have the potential to provide information that could facilitate the control of compensatory stepping reactions.

It has been proposed by Nurse and Nigg (2001) that reduced feedback from the receptors in the foot may contribute to gait abnormalities. Impaired proprioception obstructs walking because it deprives the individual of knowing the exact position of their knee, hip, ankle and/or foot, and as a result, that individual does not know when it is safe to transfer body weight onto the limb (Perry & Burnfield, 2010). Similarly, Hsu et al. (2008) stated that impaired sensation on the soles of the feet delays the awareness of floor contact and could result in a greater occurrence of falls or other lower extremity biomechanical problems.

2.6 Neuroimaging Techniques

Functional brain imaging is a multidisciplinary research field that encompasses techniques devoted to a better understanding of the human brain through non-invasive imaging of the electrophysiological, hemodynamic, metabolic and neurochemical processes that underlie normal and pathological brain function (Baillet et al., 2001). Neuroimaging refers to a set of brain-scanning techniques that allow for observation of varying levels of activity in an individual’s brain. Prior to neuroimaging, the only way to learn about the human brain was by cataloguing the set of impairments that a group of patients would exhibit when they suffered damage to a given brain region (Sadava et al., 2008). With neuroimaging, came the use of a variety of different scanning and imaging techniques, including; electroencephalography (EEG), functional magnetic resonance imaging (fMRI), positron emission tomography scan (PET), computed tomography scan (CT), single-photon emission computed tomography (SPECT), near-infrared spectroscopy (NIRS), and transcranial magnetic stimulation (TMS). Unfortunately, functional neuroimaging of the brain during human gait poses a practical problem. The investigation of brain activation patterns during locomotion has been limited by the impossibility of actually performing these tasks while inside standard scanners, such as MRI machines or PET/CT scanners (Jahn et al., 2004). Some of these problems have been overcome, as researchers are adopting alternative neuroimaging strategies (Wieser
et al., 2010). These strategies included techniques such as recording cerebral activity during actual gait after radioactive injections (Fukuyama et al., 1997; Hanakawa et al., 1999; Miyai et al., 2001); recording cerebral activity during motor planning of walking or just prior to gait initiation (Yazawa et al. 1997); using tasks that share some cerebral processes with gait, without the actual need to engage in gait, like motor imagery or repetitive foot movements (Malouin et al., 2003; Jahn et al., 2004; Miyai et al., 2001); and even recording cerebral activity in patients with gait disorders.

2.6.1 Electroencephalography (EEG)

EEG is a method of studying the collective electrical behaviour of large groups of neurons, where it detects changes in the electrical potential differences between electrodes, resulting from the flow of current through the extracellular space (Latash, 2008). The EEG electrodes are placed on the skull, commonly over the four major cortical lobes, with reference/grounding electrodes placed behind each ear. The EEG device is able to detect frequencies ranging from 1-30 Hz, with four different wave classifications depending on their frequency. Delta waves detect frequencies from approximately 0.5-4 Hz, theta waves from 4-8 Hz, alpha waves from 8-13 Hz and beta waves from 13-25 Hz (Latash, 2008; Emotiv). The low frequency waves (delta and theta) are commonly only seen during certain phases of sleep. Alpha waves are associated with relaxed wakefulness and beta waves are dominant during intense mental activity (Latash, 2008; Sadava et al., 2008). A distinct advantage of using EEG is its high temporal resolution, which can follow changes in brain activity from millisecond to millisecond, making it far more superior to PET or fMRI in this respect (Irani, 2011). However, EEG has very poor spatial resolution, since the electrodes are simply placed over the skull, they cannot be used to properly identify the exact source of the changing electrical signals that are recorded (Latash, 2008).

Many studies have been conducted using EEG to determine areas of the cerebral cortex involved in the initiation of gait, or during lower limb movements similar to that of gait movements, and even during gait performed on a treadmill. In 1997, Yazawa et al. performed a study using EEG to explore the cortical mechanism underlying gait initiation after two auditory cues (tone bursts at 1000Hz or 2000Hz). They also incorporated the
recording of surface electromyogram (EMG) on the anterior tibialis muscle in order to determine the onset of movement. From this, Yazawa et al. (1997) demonstrated that activity in the supplementary motor area as well as the primary motor cortex was related to gait initiation. Similarly, a study investigating the cortical activity during lower limb movements by means of EEG was used with the assistance of a tilt table. Wieser et al. (2010) collected EEG data while participants were strapped into an Erigo tilt table, which allowed them to perform automated gait-like stepping movements in an upright position for thirty minutes. The Erigo is driven in a sinusoidal function of time referring to the hip angle, with the duration of extension and flexion phases identical for each leg, while moving conversely to each other at a constant speed (Wieser et al., 2010). This study provided further indication that the primary somatosensory cortex, primary motor cortex and the supplementary motor area, all play essential roles in cortical control of human gait. While standing, walking and running on a treadmill, Gwin et al. (2011) used EEG to examine patterns of intra-stride electrocortical dynamics. It was proposed that the electrocortical dynamics, particularly in the sensorimotor cortex, would exhibit intra-stride patterns of activation and deactivation. Clusters of electrocortical sources were spatially localized to the prefrontal cortex, left and right sensorimotor cortex, anterior cingulated cortex and posterior parietal cortex (Gwin et al., 2011). However, Gwin et al. (2011) stated that their findings did not indicate whether the human cortex is actively involved in controlling locomotion via direct pathways, or if it processes sensory afferents that are used to modulate a descending signal to other locomotor regions in the brainstem and spine.

All complex EEG systems used to detect brain activity are sensitive to a set of elements that can negatively influence the accuracy of their measurements (Cernea et al., 2012). Recording of EEG during walking is challenging due to movement artifacts (Bakker et al., 2007b) and most EEG devices actually detect a mixture of skin, muscle and nerve activity instead of just a pure signal generated by the electrical activity of neurons (Van De Velde et al. 1998). Nuwer (1990) reported that not only are there movement artifacts to consider, but also line noise, frequency interference, as well as medications that people are taking and other clinical factors. One limitation to EEG that was expressed after the study conducted by Wieser et al. (2010) was that activity in
subcortical brain structures (activations in the striatum, cerebellum, pons, basal ganglia) could not be considered or recorded with this device.

2.6.2 Magnetic Resonance Imaging (MRI) / Functional Magnetic Resonance Imaging (fMRI)

MRI is based on the property of elements with an odd atomic weight to align the spin axes of their nuclei with a constant external magnetic field (Latash, 2008). A brief electromagnetic pulse can be used to perturb the orientation of the spin axes, so when the pulse is turned off, the nuclei return back to their original orientation defined by the external magnetic field (Latash, 2008). This process involves the release of energy in the form of electromagnetic waves, where the frequency of emitted waves can vary with regards to the different types of atoms. A particular version of MRI, the fMRI, has become popular for studying the brain processes associated with different actions and movements. This method involves comparing MRI measurements obtained before and after performing a task, where the differences in the patterns are expected to reflect task-specific changes in the neuronal activity (Latash, 2008). fMRI can be used to produce activation maps showing which parts of the brain are involved in a particular mental process (Sadava et al., 2008). Irani (2011) states that fMRI is currently considered the ‘gold standard’ for measuring functional brain activation, since it offers a source of safe, non-invasive, functional brain imaging with high spatial resolution. The primary measure used for this technique is the blood oxygen level-dependent (BOLD) signal, which reflects the amount of oxygenated hemoglobin in the total amount of hemoglobin, and accompanies neuronal activation in the brain (Irani, 2011). Unfortunately, studying the cortical involvement in gait control for humans using this technique has been limited, due to the incapability to walk inside the scanning machine. However, it has been suggested that activation maps during movement execution are similar to those observed during the imagery of the same task (Jeannerod, 2001; Munzert et al., 2009).

In 2008, Wang et al. conducted a study requiring the observation of video clips of human walking and the mental imitation of the visualized process while using an fMRI. This method was based on the response of mirror neurons to actions being observed. This involved observation with the intent to imitate and has been associated with regions
involved in the planning and generation of actions. Wang et al. (2008) determined that there was activation in the left supplementary motor area and the dorsal premotor area during the visual task using motor related imagery. Similarly, another study was conducted by Wang et al. (2009) exploring the cortical control of gait-related imagery using fMRI. These gait related imagery tasks were specific to gait initiation, stepping over obstacles and gait termination. From these imagery tasks, Wang et al. (2009) were able to provide evidence that the supplementary motor area was preferentially activated during imaginary gait initiation, and demonstrated that there is cortical control involved in the different phases of human gait. Comparable to the study completed by Wang et al. (2008), Iseki et al. (2008) explored the idea of higher-level cortical gait planning centers in humans using fMRI and imaginary gait situations that were viewed through video clips of differing situations. The video clips consisted of observation of gait movements from the third person perspective, observation of stepping movements, observation of standing posture, virtual walking, and the scrambled versions of the observation of gait and virtual walking. Iseki et al. (2008) established that the virtual walking condition activated the dorsal premotor cortex, supplementary motor area/cingulate motor area and subcortical nuclei. Whereas the observation of gait movements from a third person perspective yielded activation of the supplementary motor area, dorsal premotor cortex, the inferior frontal gyrus and inferior parietal lobe. It was suggested from these results that activation of the supplementary motor area as well as the premotor area occurred during both the observation of gait and the mental imagery of gait.

A completely different approach was taken by Jahn et al. (2004). This group used fMRI during imagined stance and locomotion to identify areas of the brainstem, cerebellum and cerebral cortex involved in locomotion. Participants were asked to perform four different conditions; lying, standing, walking and running, and were then trained to imagine these conditions while lying supine on a floor. After these training periods were complete, the participants were then instructed to imagine themselves performing these conditions while inside the MRI scanner, where each condition was tested multiple times. Jahn et al. (2004) detected separate and distinct activation/deactivation patterns for the three imagined dynamic conditions. Walking imagery was associated with activation patterns in areas involved with visuospatial
navigation, occipital visual areas, supplementary motor area, medial and inferior temporal lobe and in the cerebellum. Whereas deactivation patterns for walking were noted in the vestibular and somatosensory cortices. It was mentioned that during the imagination of stance and locomotion, large BOLD signal increases were not found in the cortical motor areas, despite the fact that complex motor performances were being imagined. Jahn et al. (2004) speculated that the lack of activation in the premotor cortex observed could be due to task difficulty, and that this area would be activated during more difficult locomotor tasks.

Currently, there have been only a few limitations mentioned involving the use of fMRI as a neuroimaging technique. These limitations include methods requiring a high degree of participant cooperation, as the participants must remain motionless during the data collection period. Also, the time resolution of fMRI is poor, due to the fact that to achieve observable differences between conditions, the actions being performed between the measurements need time to produce metabolic changes reflected in the signal (Latash, 2008). Lastly, little standardization leads to interpretation differences between researchers and research groups. Most studies use the BOLD response which reflects the amount of oxygenated hemoglobin in the total amount of hemoglobin (Latash, 2008).

2.6.3 Computed Tomography Scan (CT) & Positron Emission Tomography Scan (PET)

CT scans are based on radiography and involves a series of narrow beams of radiation. An x-ray source is placed on one side of the skull and detectors are placed on the opposite. Both are then rotated in steps around the head, where a series of x-ray transmissions are recorded at each step (Latash, 2008). A PET scan is similar to a CT scan, given that it involves analyzing x-rays in different directions. However the source of x-rays is not from an external emitter but it is a radioactive isotope injected into the circulatory system (Latash, 2008). The radioactive isotope injected could be used to label an analogue of glucose, where it is metabolized in the body. Therefore, the amount of radiation that is released by a cell is proportional to the number of metabolized glucose-analogue molecules it contains, which then generally correlates with the level of cell activity. Another method of imaging includes the use of the oxygen-15 labelled water,
where the H2O-15 emits and creates images based on regional cerebral blood flow (rCBF) within the brain. Therefore, PET scans can measure rCBF and the cerebral metabolic rate of oxygen (Franceschini & Boas, 2004). Changes in rCBF reflect changes in underlying neuronal activity, so to assess changes in rCBF during an activation task, at least two conditional studies (baseline and an activated condition) must be completed (Fukuyama et al., 1997).

A study aimed at assessing the cerebral structures selectively involved in locomotor related tasks through the use of PET scans was executed by Malouin et al. (2003). PET was used in conjunction with a form of motor imagery involving visuo-spatial components. Locomotor tasks included standing, gait initiation, and walking with and without obstacles. These conditions were compared to a rest (control) condition in order to identify the neural structures involved in the imagination of locomotor-related tasks. Each participant underwent eight PET scans within two hours. Each of the scans lasted about one minute and were separated by an inter-scan period of approximately ten minutes. During the imagery conditions, the participants were instructed to imagine, in the first-person perspective, the motor tasks illustrated in the video prior to scanning. Malouin et al. (2003) discovered that there was a common set of activated structures in all simulated conditions, which included the dorsal premotor cortex, prefrontal cortex and inferior parietal lobe. Additional areas involving the pre-supplementary motor area and the leg areas of the motor cortex were activated during conditions that required the imagery of locomotor events.

PET and fMRI have demonstrated that during rhythmic foot or leg movements the primary motor cortex is activated, consistent with expected somatotopy, and that during movement preparation and anticipation the frontal and association areas are activated (Christensen et al., 2000; Dobkin et al., 2004; Heuninckx et al., 2005; Luft et al., 2002; Sahyoun et al., 2004). However, Miyai et al. (2001) stated that although PET and fMRI are powerful tools for the brain mapping of somatosensory, visual, auditory and cognitive tasks as well as motor tasks of hands, they are ill-suited to analyze true locomotor tasks. It has also been suggested that PET and SPECT are limited in their ability to perform continuous or repeated measurements since there are concerns regarding the use of
radioactive isotopes and the half-life of the oxygen-15 labelled water molecule (Irani, 2011).

### 2.6.4 Single-Photon Emission Computed Tomography (SPECT)

SPECT is a nuclear imaging test that allows special gamma ray-detecting cameras to create 3-D pictures of structures inside the body. SPECT is used in combination with technetium-99m-hexamethyl-propyleneamine oxime ($^{99m}$Tc-HM-PAO), which is a gamma-emitting tracer molecule. This technique records cerebral activity during walking by injecting the radioactively labeled HM-PAO during locomotion and recording cerebral activity afterwards with SPECT (Bakker et al., 2007b). When radioactive HM-PAO is injected intravenously during gait, it is rapidly distributed in the brain in proportion to regional cerebral blood flow and is retained in the brain for hours (Bakker et al., 2007b). Therefore, the distribution pattern of HM-PAO at the time of scanning reflects the pattern of cerebral perfusion at the time the radioactive tracer was injected into the bloodstream. This technique enables researchers to study human brain function even when participants are moving freely during task performance, as time for the task performance and time for the image acquisition can be separated (Hanakawa et al., 1999). SPECT imaging is similar to PET scans in its use of radioactive tracer material and detection of gamma rays (Sadava et al., 2008). A major difference, however, is that the tracer used in SPECT stays in the bloodstream rather than being absorbed by surrounding tissues like in PET scans, which limits the images to areas where blood flows (Mayfield Clinic for Brain and Spine).

Fukuyama et al. (1997) performed a study that evaluated the changes in brain activity during voluntary human walking in normal participants, using $^{99m}$Tc-HM-PAO and SPECT imaging. Participants walked along a 20 metre corridor, at a regular pace, for a total of four minutes. The participants were then injected intravenously with HM-PAO and instructed to walk for another four minutes back and forth in the same corridor. Following the second walking period, the participants were then instructed to lay on the scanner for the next thirty minutes. Immediately after the first scanning period was complete, participants were injected again with HM-PAO, in order to collect a resting state, which was fifteen minutes in duration. Fukuyama et al. (1997) discovered that the
medial primary sensorimotor area, supplementary motor cortex, cerebellum, visual cortex, basal ganglia, and a small region of the medial temporal lobe were significantly activated during locomotion. The researchers concluded that higher cortical centers are indeed involved in the mechanism of bipedal gait.

A similar study using SPECT imaging was conducted by Hanakawa et al. (1999) to evaluate the regional cerebral blood flow changes during gait, except gait was performed on a treadmill. This group of researchers were comparing brain activity changes induced by gait on a treadmill in patients with Parkinson’s disease to changes in age-matched controls. Participants were injected intravenously with HM-PAO approximately thirty seconds after gait on the treadmill commenced. This way, the maintenance of walking and not the initiation of walking was evaluated. Participants continued walking for four and a half minutes after the tracer injection, and were then instructed to lay on the scanner bed. This initial scan was performed over a twenty minute period, after which the individual was injected a second time with HM-PAO and evaluated during a rest period for an additional twenty minutes in the scanner. In the age-matched control group, Hanakawa et al. (1999) observed a gait induced increase in brain activity in the medial and lateral premotor areas, primary sensorimotor areas, anterior cingulated cortex, superior parietal cortex, visual cortex, dorsal brainstem, basal ganglia and the cerebellum. Therefore, in addition to the areas reported by Fukuyama et al. (1997), Hanakawa et al. (1999) expanded upon the areas of increased brain activity to include premotor cortex areas, somatosensory association cortex, cingulate areas, as well as areas in the brainstem.

A downfall to this method of neuroimaging involves the use of multiple radioactive injections and the length of imaging, as well as the need to compare cerebral activity between the gait session and the physical resting condition (Bakker et al., 2007b). The comparison between the two different conditions, where the brain activation at rest is subtracted from the gait condition, raises the issue of whether those cerebral changes are in fact related to the process of gait or if other factors are contributing to the observed differences (Bakker et al., 2007b).
2.6.5 Near-Infrared Spectroscopy (NIRS)

NIRS records the transmission and absorption of near infrared light by human tissue. However the skull does not absorb much infrared light, so NIRS can be used to measure the levels of oxygenated, deoxygenated and total hemoglobin that is related to neural activity in the superficial cortical areas of the brain (Bakker et al., 2007b). Essentially, a NIRS system observes dynamic changes in regional cerebral blood flow in real time by measuring the concentration changes in cerebral hemoglobin (Hoshi, 2007). These concentrations of cerebral hemoglobin can be measured within a few centimetres’ depth of the skull surface covering the cerebral cortex (Miyai et al., 2001). Regional brain activation is accompanied by an increase in rCBF, which carries both glucose and oxygen to the area, where oxygen is transported via hemoglobin (Irani, 2011). It is expected that increases in total hemoglobin and oxyhemoglobin, as well as decreases in deoxyhemoglobin will be observed by NIRS in activated areas, as oxygenated and deoxygenated hemoglobin have characteristic optical properties in the visible and near-infrared light range (Hoshi, 2007; Irani, 2011). Functional near-infrared spectroscopy (fNIRS), capitalizes on the changing optical properties of brain tissue and uses light in the near-infrared range of the visible spectrum to measures hemodynamic responses to sensory, motor and other cognitive activity (Hoshi, 2007; Irani, 2011). The optodes of the NIRS system are fixed to the skull, so head movements are allowed during measurements, thereby allowing assessments while individuals walk on a treadmill.

Miyai et al. (2001) executed a study aimed at determining the basic cortical activation patterns during ordinary gait while walking on a treadmill and using NIRS imaging. Before beginning the walking trials on the treadmill, an anatomical MRI scan was used to confirm that the optodes of the NIRS system were appropriately positioned over the cerebral cortex. Participants then performed four different tasks on the treadmill, alternating every thirty seconds, and each task was repeated five times. Treadmill tasks were completed while NIRS simultaneously detected changes in oxygenated, deoxygenated and total hemoglobin concentrations. The NIRS topographic maps were then matched with the MRI anatomical images to determine the correct areas associated with increased activity. Miyai et al. (2001) revealed increased activity in areas
corresponding to the medial portion of the primary sensorimotor region and the supplementary motor areas during bilateral human gait.

Similar testing techniques were carried out by Suzuki et al. (2008), where brain activity patterns were investigated on a treadmill using fNIRS during regular and cued-walking. Suzuki et al. (2008) speculated that some of the gait related changes in hemoglobin concentrations reported by Miyai et al. (2001) may have been within the gait preparation/initiation phase and not actually during the locomotor stage itself. Therefore, the researchers decided to compare between brain activation patterns during regular and cued walking, with pseudo-randomly ordered rest periods. Suzuki et al. (2008) discovered that the verbal instructions to prepare for walking affected regional activations in the prefrontal cortex, the supplementary motor area, the premotor cortex and medial sensorimotor cortex before walking and during the acceleration phase of walking. Suzuki et al. (2008) concluded that preparation of gait shares similar structures in the frontal cortex with gait execution.

In another study conducted by Suzuki et al. (2004), the changes in regional activation in the cerebral cortex were investigate using NIRS imaging while running and walking at different speeds on a treadmill. They evaluated the relative changes in oxygenated and deoxygenated hemoglobin concentration levels through NIRS imaging techniques while walking at three and five kilometres per hour and running at nine kilometres per hour. The NIRS data was compared with anatomical MRI images to locate the corresponding area in the brain that was activated by the locomotor task. Suzuki et al. (2004) observed increasing activation levels in the prefrontal cortex and the premotor cortex as locomotor speed and cadence was increased. The activation levels recorded in the medial supplementary motor cortex appeared to be unchanged or decreased as the locomotor speed increased. Additionally, Suzuki et al. (2004) noticed that immediately after stopping the locomotor tasks at five and nine kilometres per hour, there were drops in the oxygenated hemoglobin levels. It was concluded that multiple motor areas were activated during the periods before reaching constant speeds of walking or running and that the prefrontal and premotor cortices might be involved in controlling locomotion to adapt to the increasing speed during the acceleration phases.
When compared to SPECT, NIRS has several advantages. NIRS does not involve a radioactive tracer, has superior temporal resolution, and allows for the comparing of several different conditions (Bakker et al., 2007b). NIRS temporal resolution is high and completely non-invasive, which allows for long-time continuous measurements in real time and repeating measurements within short intervals (Hoshi, 2007). NIRS also allows for less motion restrictions, so that tasks may be performed with greater ecological validity. Although NIRS has been particularly useful for studying the cortical bases of locomotion control, the limited penetration of infrared light enables it to only assess the responses of the most superficial portions of the cerebral cortex (Bakker et al., 2007b).

2.6.6 Transcranial Magnetic Stimulation (TMS)

TMS is another technique that has been used to examine the neural substrate of gait during actual human walking. This is a non-invasive method used to depolarize or hyperpolarize neurons in the brain. TMS uses electromagnetic induction to generate an electric current across the scalp and skull without physical contact (Sadava et al., 2008). An enclosed coil of wire is held next to the skull and when activated, produces a magnetic field that is oriented orthogonal to the plane of the coil. This magnetic field passes unimpeded through the skin and skull, and induces an oppositely directed current in the brain that activates nearby nerve cells (Guyton & Hall, 2006). If this instrument is used on the primary motor cortex, it can produce muscle activity that can be recorded by an electromyogram (EMG). TMS enables the direct study of intra-stride modulations in corticospinal excitability (Gwin et al., 2011). Studies using TMS have shown that activation of inhibitory circuits in the motor cortex during steady walking disrupted ongoing cortico-muscular interaction and reduced lower limb (plantar and dorsiflexor) activity (Capaday et al., 1999; Petersen et al., 2001). Therefore, TMS has been a useful tool in examining the contribution of the corticospinal tract to the control of gait (Bakker et al., 2007b).

In 2001, Peterson et al. completed a study investigating whether the increased cortical excitability through the use of TMS directly related to the activation of spinal motoneurons during human walking. The involvement of the motor cortex during human walking using TMS and recording the EMG activity in the tibialis anterior and soleus
muscles was tested. A pressure sensitive trigger was placed under the heel of the participant’s shoe and was the trigger used to start the sampling on the computer. This way, after the trigger was activated, the computer would record the EMG activity as well as activate the magnetic stimulator. Peterson et al. (2001) hypothesized that by activating the inhibitory mechanisms effectively within the cortex, there should be a decrease in cortical excitability and thus reduce cortical output, thereby suppressing the EMG activity. It was shown that ongoing EMG activity during locomotion in the tibialis anterior and soleus muscles can be suppressed by weak magnetic stimulation of the motor cortex. From these results, Peterson et al. (2001) inferred that the motor cortex is directly involved in the continuing activation of the lower limb motorneurons during human walking.

To determine the significance of corticospinal input upon the pattern involved in locomotion, Schubert et al. (1997) used TMS on the motor cortex during treadmill walking. Muscle activation during locomotion was recorded using EMG on the tibialis anterior and gastrocnemius muscles. Participants were instructed to walk on the treadmill, where steps were coordinated in time to a metronome set to 100 beats per minute. Stimulation was triggered with reference to the impact of the right heel, which was detected by force transducers located beneath the treadmill belt. The averages of fifteen stimuli, which were introduced randomly at each phase of the gait cycle, were analyzed. This study illustrated the effect that TMS had on short-latency EMG responses when cortical stimuli were delivered during distinct phases of the human gait cycle. Schubert et al. (1997) discovered changes in the amplitude of evoked motor responses in the tibialis anterior and gastrocnemius muscles during locomotion. The researchers concluded that responses in the tibialis anterior muscles could be evoked more easily that in the gastrocnemius muscle in the majority of participants.

Although TMS is often regarded as safe, due to its non-invasive nature and lack of physical contact, an acute risk of TMS is the rare occurrence of induced seizures (Rossi et al., 2009). These seizures could be due to predisposing factors such as medications, brain lesions or genetic susceptibility. Other documented risks include
syncope, minor pains such as headache or local discomfort and other minor temporary cognitive changes (Rossi et al., 2009).

2.6.7 Real Gait vs. Gait Imagery

Given the technical problems and limitations associated with assessing the cerebral bases of true gait control inside scanners, several researchers have chosen to focus their efforts on the more tractable aspects (Bakker et al., 2007b). Some research groups; Miyai et al. (2001), Malouin et al. (2003) and Jahn et al. (2004), have explored motor imagery of gait or imagination of gait as a proxy for the real thing, for example the mental simulation of gait without actual execution. This approach exploits the documented neural and cognitive overlap between movement planning and motor imagery, as imagining a movement relies on neural processes similar to those evoked during actual performance of the same movement (Bakker et al., 2007b).

Miyai et al. (2001) combined fMRI and NIRS to show a degree of overlap between actual and imagined gait. Similarly, Jahn et al. (2004 & 2008) used fMRI to evaluate imagination of gait, revealing there were activations recorded in the supplementary motor area, cingulate motor area, posterior parietal area, the cerebellum and the mesencephalon. Not surprisingly, this group did not observe any significant activity in the primary motor area. Bakker et al. (2007a) concluded that motor imagery of gait is sensitive to the same temporal and spatial constraints as actual walking movements. Hanakawa et al. (2008) evaluated the execution and imagery of sequential finger-tapping movements through the use of an fMRI. They discovered that many frontoparietal and posterior cerebellar areas were activated in both conditions, however, executive regions (primary motor cortex and sensory cortex) were active mainly with actual performance of the movement (Hanakawa et al., 2008). Furthermore, they discussed the regions predominantly involved with the imagery of finger-tapping were regions that were more frontal and posterior based.

Using imagination of gait or gait imagery as a proxy for the real thing has some advantages. During real gait, there is an effect of the visual environment, which may be less confounding with imaginary gait (Hallett & Iseki, 2012). Whereas, motor imagery
allows one to study the cognitive and cerebral properties of movement independently of motor output and sensory feedback. Furthermore, motor imagery can be studied in a recumbent position compatible with techniques such as fMRI and PET (Bakker et al., 2007b; de Lange et al., 2005). As mentioned, imagined locomotion differs from real locomotion in the cognitive (imagery) element, as well as in the absence of correlated sensory input from proprioceptive, vestibular and visual systems (Jahn et al., 2008). However, these inputs are vital in the production of gait in a real life environment, where interactions from a number of elements influence gait patterns and techniques.

2.7 New Neuroheadset Technology

A relatively new device designed initially for gaming, has been introduced to the world of research and is now being utilized as an investigative tool in a variety of different fields. The Emotiv EPOC neuroheadset is a high resolution, multi-channel, wireless portable EEG device enabling a broad range of applications including neurotherapy, biofeedback and brain-computer interface (Emotiv). The headset acquires the neuronal activity in the brain through the use of fourteen electrodes that are placed on the human scalp. The placement of the electrodes are based upon the International 10-20 locations (AF3, F7, F3, FC5, T7, P7, O1, O2, P8, T8, FC6, F4, F8, AF4), along with two reference electrodes that are placed behind each ear (Emotiv). The headset then transmits the encrypted raw brain wave data wirelessly back to a proprietary software package through a USB receiver connected to a computer. Most often in the field of neuroscience, EEG systems are utilized for recording the electrical activity of the brain in order to detect abnormal activity, like in the diagnosis of epilepsy (Cernea et al., 2012). However, this completely non-invasive, wireless, functional brain imaging technique will also allow for the collection of real time brain activity, in the form of brain waves, while participants are walking in a live setting.

The intensities of brain waves recorded from the surface of the scalp can range from 0-200 microvolts and their frequencies can range from once every few seconds to fifty or more per second (Sadava et al., 2008). The character of the waves is dependent upon the degree of activity in the particular area of the cerebral cortex, and there are noticeable changes in wave forms between the states of wakefulness, sleep and coma.
The EEG electrodes that are placed on various areas across the scalp record changes in the electrical potential differences between electrodes over time. EEG waves can be classified as being alpha, beta, delta or theta waves. Alpha waves are rhythmical waves that usually occur at frequencies between 8-13Hz (cycles per second). They are found when you are awake but your eyes are closed, or when you are in a quiet state of relaxation (Sadava et al., 2008). This type of wave form occurs most frequently in the posterior regions of the cerebral cortex, in the occipital region, but can also be recorded from parietal and frontal regions (Emotiv). Beta waves occur at a higher frequency, usually between 13-30 Hz or greater. These waves are predominantly recorded from the parietal and frontal regions of the cerebral cortex. Beta waves are closely linked to motor behaviour or a specific type of mental activity and are generally present during active movements (Sadava et al., 2008; Emotiv). Occasionally, these waves are said to be associated with the muscle contractions that happen in isotonic movements and are suppressed prior to or during movement changes (Baker, 2007). Delta waves occur at frequencies less than 4 Hz, but could have voltages up to four times greater than most other waves (Sadava et al., 2008). Wave forms of this type occur in very deep sleep, in infancy, and also in very serious organic brain diseases (Sadava et al., 2008). Lastly, theta waves occur at frequencies between 4-8 Hz. These waves are normally recorded from the parietal and temporal regions of the brain. Normally, these wave forms appear in young children, or in older children and adults during drowsy, meditative, or emotionally stressed periods (Sadava et al., 2008).

Intended mainly as a device used for gaming, the Emotiv EPOC neuroheadset was designed with an entire spectrum of application fields in mind, ranging from interaction support for disabled patients to artistic expression and even research (Cernea et al., 2012). The software that is incorporated with the headset provides detection functionality for various facial expressions, levels of emotion, cognitive neuro-activities, as well as a research component (Campbell et al., 2010). In 2010, Campbell et al. presented a design, implementing the use of the Emotiv headset with a mobile phone. By using this ‘Neurophone’ system, users could simply think their way through all of their mobile applications, specifically a brain-controlled address book. Scherer et al. (2004) used this headset technology to create a virtual keyboard and essentially created a hands-free input
device. Brain-computer interface (BCI) research has been focused on using an EEG-based system for providing interaction possibilities and movement assistance to physically handicapped individuals or people with motor disorders (Leeb et al., 2007). More recently, in 2010, Ranky and Adamovich introduced the use of the Emotiv EPOC headset as a controlling device for a robotic arm. However, even though this wireless neuroheadset has been employed in a variety of areas thus far, it has yet to be applied and explored in the realm of neuronal activity in human gait.

One of the problems with EEG systems is regarding the placement of the electrodes and their accurate positioning. While the Emotiv neuroheadset implicitly positions the electrodes, the users still need to pay particular attention to reading and following the positioning instructions, as well as ensuring that the electrode have sufficient saline to allow for good scalp contact (Cernea et al., 2012). In a study conducted by Cernea et al. (2012), there was an overall positive consensus regarding headset feedback, involving comfort levels and ability to perform tasks while wearing the headset. However, there were some users who expressed painful sensations or headaches after wearing the headset for prolonged periods of time. These issues were reportedly due to the side pressure that the device applies on the scalp in order for it to remain stationary on the head and not disrupt the EEG measurements (Cernea et al., 2012). One major advantage to this method of functional brain neuroimaging is the wireless aspect of this testing technique. Wireless data capture allows for human gait testing to be conducted in a natural environment, free from attached wires and cords, and walking on a more natural and ordinary surface instead of on a treadmill.

2.8 Use of Orthotics

The foot is constantly subjected to impact forces from contact with the ground and must be able to work with the rest of the body to absorb or dissipate the stresses on the structural components of the foot and lower extremity (Kirby, 2000). Foot orthoses are one type of conservative management tool used to help with foot and ankle problems. Whether by adding custom inserts or through shoe modification, foot orthoses are used to improve gait in the treatment of individuals with arthritis, diabetes, other neuropathies, trauma and congenital deformities (Edelstein & Moroz, 2011). Foot orthoses are devices
that are confined to the foot only and do not encompass the ankle. Therefore, the support primarily covers the plantar surface and benefits the foot upon weight bearing (Hsu et al., 2008). It is widely believed that the most essential element of clothing in any person’s wardrobe is their shoes. This is because shoes perform the vital functions of transferring body weight to the floor during walking and provide support for the feet (Lusardi & Nielson, 2007). The comfort of the orthotic inside a shoe is an important consideration, because a comfortable orthosis is likely to minimize muscular work during walking (Lusardi & Nielson, 2007). According to Nigg et al. (1999), an optimal orthosis would reduce muscle activity, feel comfortable, and improve musculoskeletal and neuromuscular performance in walking.

Foot orthoses are mechanical interventions that are generally designed to affect the individual’s musculoskeletal system. However, the effectiveness of an orthoses in improving an individual’s functional abilities is also impacted by neuromuscular control, cardiovascular and pulmonary function, as well as complex interactions among many personal and environmental factors (May & Lockard, 2011). Orthoses can be administered or prescribed to individuals for a variety of reasons. Some common goals for foot orthoses include providing cushioning to improve shock absorption, providing relief for pressure-sensitive plantar structures and reducing shear forces on the foot (May & Lockard, 2011). Other reasons for prescription would include, but are not limited to, improve sensory feedback, balance or support the joints of the foot, limit or restrict excessive or abnormal movements, correct flexible deformities or accommodate and support a fixed deformity (May & Lockard, 2011). There are two distinct categories of foot orthoses to help with these various issues; functional and accommodative. Functional foot orthoses are an orthopaedic device designed to promote structural integrity of the joints of the foot and lower limb by resisting the ground reaction forces that cause abnormal skeletal motion during gait (Lusardi & Nielson, 2007). The accommodative type of foot orthoses are used to distribute pressures over the plantar surface of the foot and provide more shock absorption, for individuals with fixed deformity or vulnerable neuropathic feet (Lusardi & Nielson, 2007).
In the 1960’s, Merton Root developed neutral impression casting techniques, positive cast modifications and posting techniques. The standards that he established at that time helped enhance the comfort and function of orthotics (Lusardi & Nielson, 2007). An accurate representation of the foot is necessary to create a foot orthosis that can successfully control abnormal and lower-extremity movement and still provide both stability and comfort (Laughton et al., 2002). The fabrication of an orthosis begins with capturing the morphology of the foot by obtaining a negative impression, which is achieved while the foot is in a subtalar neutral position (Laughton et al., 2002). There are multiple strategies that can be employed to take negative impressions of the foot for the purpose of fabricating an orthotic device. Some of these methods include; non-weightbearing plaster casting, partial weightbearing foam impressions, and partial weightbearing and non-weightbearing laser scanning. Historically, the standard method for obtaining negative impressions of the foot was completed using the non-weightbearing plaster casting method (Laughton et al., 2002). This technique provides the clinician with the most control of the foot and its joints, as the patient is lying prone and the clinician uses Plaster of Paris stripes to mould a slipper cast (Hsu et al., 2008). In 1967, Monty Greenawalt patented the use of a simple box filled with crushable foam to obtain a detailed impression of the foot, whereby precise measurements could be made directly from the foam cast (Ball, 2002). The foam impression is usually taken in a partial weightbearing form, where the foot is placed on the flat surface, placed in the correct position by the clinician, and is then pushed into the foam block to create the impression (Hsu et al., 2008). Laser technology has now been incorporated to obtain either a partial weightbearing or non-weightbearing representation of the foot. During the partial weightbearing procedure, the individual is sitting with the foot placed on the glass of the scanner, while maintaining a subtalar neutral position (Laughton et al., 2002). The non-weightbearing scan is taken with the individual in a long sitting position, with the foot over the end of the examination table. The clinician maintains the subtalar neutral position while the scan is completed. In 2002, Laughton et al. performed a study involving the four methods of taking impressions and compared the reliability and validity of the methods involved. Their results suggested that the different methods of obtaining a representation of the foot can produce differences in the measurements of
foot morphologies. This in turn, may affect the comfort, fit and function of the resulting orthosis.

It was reported by Nigg et al. (1999) that the use of orthotics could potentially improve sensory feedback from the plantar surface of the foot. Eils et al. (2004) demonstrated that reduced plantar sensation in a group of healthy adults leads to significant changes in gait kinematics at the ankle, knee, and hip joints and indicates a more cautious ground contact and push-off pattern. Similarly, Meyer et al. (2004) discovered that a reduced sensitivity in the sole of the foot resulted in a relative redistribution of corrective torques from the ankles and trunk to the hip joints. This emphasized the importance of the foot-sole sensation in the control of human balance. One method proposed to combat this issue of reduced plantar sensitivity is the use of an orthotic device to provide enhanced sensory feedback from the sole of the foot. Consequently, Perry et al. (2008) established that a balance-enhanced insole (designed to facilitate foot-sole sensation) improved lateral stability during gait in a group of elderly adults with moderate loss of foot-sole sensation. The idea of a textured insole was also explored by Nurse et al. (2005), where participants were tested while walking with a smooth shoe insert and a textured shoe insert. It was found that reductions in both the soleus and tibialis anterior muscle intensities (through EMG) were identified while participants wore the textured shoe insert. The results from these studies suggest that the provision of enhanced or altered sensory input appear to be of the utmost significance when exploring methods of improving locomotion (Kelleher et al., 2010).

2.9 Summary

Locomotion requires dynamic interaction between peripheral sensors, central pattern generators, and supraspinal locomotion centers, resulting in inputs from descending, peripheral and central pathways (Grillner et al., 2008; Rossignol et al., 2006). Sensory feedback is an integral part of the overall motor control system and is critical in modifying CPG-generated motor programs in order to facilitate constant adaptations to the environment (MacKay-Lyons, 2002). The continuous visual, somatosensory and vestibular feedback information is integrated with motor processing allowing for the locomotor process to operate with ease and gracefulness (Lundy-Ekman, 2013).
Although there are many techniques readily available for the neuroimaging of the brain, the neuroimaging of gait has proven to be especially challenging. This is mainly due to the practical problem of performing these activities while in the standard scanners. However, some of these issues were overcome, as researchers adopted the idea of using motor imagery of gait or imagination of gait as a proxy for the real thing. Few studies have been conducted using measurements from actual human gait, using SPECT technology and NIRS imaging (Fukuyama et al., 1997; Hanakawa et al., 1999; Miyai et al., 2001). The studies completed by Fukuyama et al. (1997) and Hanakawa et al. (1999) both used SPECT technology, however this involved executing the task (walking) prior to the image acquisitioning. Even though the study conducted by Miyai et al. (2001) using NIRS allowed for information to be recorded while walking on a treadmill, these topographic maps still had to be compared to anatomical images from MRI scans. Nevertheless, the studies involving NIRS and SPECT imaging have shown that human gait is associated with widespread cortical brain activity. This activity involves the supplementary motor area, primary somatosensory cortex, the primary motor cortex, premotor cortex, and additional subcortical structures (Fukuyama et al., 1997; Hanakawa et al., 1999; Miyai et al., 2001).

Through the use of the Emotiv EPOC neuroheadset, information about brain activity patterns during human gait can be transmitted back to a computer instantly through a wireless connection. This wireless device allows for brain neuronal activity to be detected in any environment that researchers wish to test in. The Emotiv neuroheadset eliminates the trouble of dragging along wires and cords, or injecting radioactive tracers. It also abolishes the task of comparing the data acquired to anatomical images from MRI’s.

It has been demonstrated that reduced plantar sensation leads to an increase in body sway in standing, an increased variation in foot contact in gait initiation and termination, as well as a modified pressure distribution patterns during walking (Eils et al., 2004; Nurse & Nigg, 2001; Perry et al., 2001). Since orthotic devices are alleged to improve sensory feedback (May & Lockard, 2011), this theory should be tested to discover if brain activity changes or is altered with the use of these devices. Gait and
balance are now both considered functional behaviours resulting from the complex interaction of multiple systems. The realization that neither gait nor balance is a single neural system indicates that clinical gait and balance problems may arise from different physiological derangements (Nutt et al., 2011). Using information collected from studying brain activity and the use of orthotics, some of these rather simple clinical gait problems could be resolved with orthotic devices.
Chapter 3

3 Methodology

3.1 Study Design

This pilot study was designed to explore the feasibility of a new wireless electroencephalogram device (Emotiv neuroheadset) in detecting brain wave activity during human gait. The methods and procedures used within the current study were examined to identify potential modifications needed in the design of future larger scale investigations. In addition, the utility of the Emotiv device in detecting differences in brain wave activity with and without participants wearing orthotics was examined.

3.2 Subject Recruitment and Eligibility Criteria

A convenience sample was recruited to participate in this study from the student body at Western University, London, Ontario, Canada. Individuals were recruited by means of a poster advertisement and by sampling via word of mouth. In order to participate in this study, individuals were required to be between the ages of 18 and 65 years, free of neurological impairment, and not already require the use of orthotics. Participants were excluded from this study if they were unable to attend two testing sessions within the Interdisciplinary Movement Disorders Laboratory, unable to independently walk 40 feet, or were taking medication that may impact the recording of EEG signals. A total of 19 individuals volunteered to participate, of those, two did not meet the eligibility criteria, and one was unable to complete testing secondary to poor signal quality with the electrodes (Figure 1); thus a total of 16 individuals completed this study.

Individuals were excluded from participating due to certain medication use because of the unknown effect the medication may have on gait. Since data are averaged together across all participants, there was potential for medication use to create outliers in the data. The pharmacologics that were excluded, included medications that potentially affect neurological circuitry, for example, those prescribed for anxiety and mood disorders. These types of disorders are often associated with an imbalance of naturally occurring neurotransmitters that help transport messages between nerve cells, such as
serotonin, norepinephrine and in some cases dopamine. There are a number of causes for mood and anxiety disorders, one being due to stress. The medial prefrontal cortex, anterior cingulate, and orbital prefrontal cortex are currently understood to play an important role in relaying information from primary sensory and associated cortices to subcortical structures involved in the stress response (Lopez et al., 1999). Some of these pathways are also involved in the processing of information during gait. Hence, areas of the brain where EEG activity is recorded during gait, may or may not be influenced by these types of medications. Therefore, so as not to create an outlier that could potentially drive the data one way, individuals taking medication were excluded from participating.

The research protocol, recruitment method, and mechanism for obtaining informed consent were approved by the Health Sciences Research Ethics Board, at Western University (Appendix A). All participants provided free and informed written consent.

![Flow Diagram of Participants Screened](image)

**Figure 1:** Flow Diagram of Participants Screened
3.3 Testing Procedures

All testing took place in the Interdisciplinary Movement Disorders Laboratory, located in Elborn College at Western University. Participants were assessed at two different time points throughout the course of the study. During the first visit participants provided informed consent and were assessed/fitted for a pair of custom foot orthotics. Seven to ten days later, participants returned for a second visit. At this time, they received their orthotics and participated in testing that involved having their brain wave activity recorded using an innovative neuroheadset as they walked with and without orthotics. Each walking trial took approximately 30 seconds. Overall, between both visits, a time commitment of approximately thirty minutes was required of the participants.

3.3.1 Testing Session 1 – Informed Consent & Orthoses Moulding

Participants were invited to read the letter of information describing the protocol for the study and to sign a consent form (Appendix B) indicating that they were willing to participate. At this time participants were encouraged to ask any questions and were made aware of their right to withdraw from the study at any time without consequences. Participants were then asked to complete a short demographic questionnaire that required them to provide details regarding their contact information, partial date of birth, sex, height, weight, and leg dominance. Upon completing the brief questionnaire, each participant was fitted for a custom orthotic.

This process began with capturing the morphology of both feet of each participant by obtaining negative impressions (Laughton et al., 2002) using the foam box technique (Greenawalt, 1967). This technique required a practitioner to guide the participants’ feet, one at a time, into a foam tray while maintaining the subtalar joint in a neutral position. Root et al (1977) described the subtalar joint neutral position as a position in which the joint is neither pronated nor supinated. Once in a neutral position, a downward force was applied to the dorsal surface of the foot, and the participant’s foot is pushed into the foam material. The foam box impressions were then labeled with the participant’s unique identifier number and sent to a local laboratory that specialized in fabricating orthotics (SoleScience). All procedures related to capturing the negative foot impressions and
fabricating the custom orthoses were completed by a Canadian Certified Pedorthist (author C.D.).

3.3.1.1 Orthotic Fabrication

All custom made foot orthoses were fabricated by an external orthotic manufacturer (SoleScience). Positive foot moulds were fabricated out of the negative impressions taken by C.D. The following device was made for each patient: Shell material – 16mm depth 3mm SJD 1000; arch fill with 55 durometer soleyflex ground to a zero balance throughout the device; topcovered with 3mm multiform; bottom covered with 1.5 microcell puff.

3.3.2 Testing Session 2 – Headset Calibration and Gait Trials

During the second visit participants received the pair of foot orthotics that were specifically fabricated for them from the negative impressions captured in the first visit. To ensure a comfortable fit, and to help participants acclimatize to the orthotics, participants were instructed to walk around the lab at a self-selected pace prior to the start of testing. Once a good fit was determined, the neuroheadset device was prepared, and calibrated, and participants were asked to complete two gait trials, one in each condition (with and without orthotics).

3.3.2.1 Emotiv Headset Preparation and Calibration

The Emotiv neuroheadset acquires brain wave signals through the use of 14 EEG and two reference sensors/electrodes (Figure 2). To prepare the headset for use, felt pads were moistened with a saline solution and attached to each of the 14 electrode recesses. The Emotiv headset was then switched on, making certain that the built-in battery was charged. Next, the Emotiv Dongle was inserted into the USB port of the research laptop, and, the wireless signal reception was verified by a reported ‘good’ in the Engine Status box. The headset was then gently lowered onto the participants’ head, by pulling apart the headband and lowering the sensor arms on the head from the top down, near the rear of the skull. The placement of the sensors was based upon the International 10-20 locations (AF3, F7, F3, FC5, T7, P7, O1, O2, P8, T8, FC6, F4, F8, AF4), along with two reference
electrodes (CMS, DRL) (Emotiv). Figure 3 provides a schematic of the location of sensors on the scalp of the participant.

**Figure 2:** Emotiv Wireless Neuroheadset

**Figure 3:** Sensor Location Schematic (Emotiv screenshot from research suite)
To facilitate proper placement, the sensors closest to the headset pivot points were placed directly above the ears, as close to the hairline as possible. The two reference electrodes were positioned on the bone just behind each ear lobe, and the headset tilted forward, so that the front-most sensors were symmetrically placed on the forehead, approximately 2-2.5 inches above the eyebrows. As the spatial orientation of the electrodes are fixed, after the placement of the reference electrodes behind the ears and 2-2.5 inches above the eyebrows, all other electrodes align in the proper position (Figure 4). The seating of each sensor was then adjusted until all sensors were identified as making proper contact with the scalp as verified by each sensor contact quality indicator light turning green. For further information regarding headset preparation and calibration refer to the Emotiv website www.emotiv.com and view the quick start guide under the EEG tab.

Figure 4: Headset Placement
3.3.2.2 Gait Trials

Participants completed the gait testing in a single testing session. The testing procedure required participants to walk approximately 20 feet down a flat even walkway, turn, and walk back to the original starting position. At the start of each trial, participants were instructed to walk in a normal fashion (i.e. as they typically would in everyday life), to keep their head looking straight ahead, and to refrain from chewing gum or talking during the trials. To facilitate a forward gaze, testing was completed in a laboratory that presented minimal visual distractions. Each trial was completed in the participant’s normal footwear (i.e. running shoes) and was executed at a self selected pace. In total, each participant completed two walking trials, one with orthotics, and one without.

To control for order bias, the order in which the two trials occurred for each participant were randomized such that half of the participants first completed the gait testing without the orthotics, while the other half first completed the gait testing with the orthotics inserted in their shoes. During each of the trials, the Emotiv wireless neuroheadset simultaneously recorded the participant’s brain activity in real-time and transmitted the raw EEG data to the USB receiver connected to the laptop.

3.4 Sample Size

The sample size of sixteen healthy individuals was chosen because it was comparable to other studies conducted in similar fields of testing (Fukuyama et al., 1997; Hanakawa et al., 1999; Miyai et al., 2001; Wang et al., 2008). This pilot study was not conducted with the intention to have enough statistical power to estimate treatment effects or assess statistical significance (Arain et al., 2010; Leon et al., 2011).

3.5 Outcome Measures

The raw EEG data produced from the Emotiv neuroheadset was the primary outcome measure in this study. The changes in Power Spectral Density (PSD) across the frequency spectrum (alpha, 8-13Hz; delta, 0-4Hz; beta, 13-30Hz; and theta waves, 4-8Hz) were compared between subject’s trials (with and without orthotics) to test the differences in brain wave activation patterns. A grand average of the PSD values for all
channels, which describes how the power of a signal or time series is distributed over the different frequencies, was used to compare between the two trials.

To process and analyze the raw EEG data from the neuroheadset, MATLAB (The MathWorks, Natick, MA, USA) was used along with the EEGLAB module and toolbox (Delorme & Makeig, 2004). Raw data was marked (by eye) in the EEGLAB module and as depicted in Figure 5, the gait trials were separated into sections. Walking sessions, marked as “b1” and “b2” represent the time point at which continuous gait began during the baseline condition. Similarly, the sections marked “o1” and “o2” represent the time points at which continuous gait started during the orthotic walking condition. It was these two sections of continuous walking (b1 & b2) that were averaged to calculate the average PSD for the baseline condition, and the continuous walking (o1 & o2) that were averaged to calculate the average PSD for the orthotic condition. Each continuous walking session lasted for a period of about 6 to 8 seconds. The other areas flagged in Figure 5 signify the portions of the trials that were excluded from the analysis. The label “g1” corresponds to the activity involved in gait initiation, while “g2” captures the artifacts associated with turning events, and “g3” signifies gait termination. The excluded portions of gait were determined for each individual participant by eye, as these patterns of EEG activity were not consistent with the continuous walking portions.
Figure 5: Raw EEG Data from a Single Subject During Both Gait Conditions (where ‘g1’ represents gait initiation, ‘b1’ first section of baseline continuous gait, ‘g2’ represents the turning events, ‘b2’ is the second section of baseline continuous gait, and ‘g3’ is gait termination. Similar events occur after the boundary break, except the conditions ‘o1’ and ‘o2’ represent the continuous walking sections of the orthotic condition.)

The data were then filtered from 1 to 40Hz and walking period time frames were submitted to an extended infomax independent components analysis (ICA) (Figure 6). ICA was used to isolate ocular artifacts (identified “by eye”), which are portrayed by components 8-10 in Figure 6. Furthermore, ICA was used to classify 2 types of activation patterns (determined “by eye”). These activation patterns consisted of: 1 “low”, large amplitude rhythmic oscillations resembling the gyro signal, that were likely associated with movement artifacts related to walking motion; and 2 “high”, smaller amplitude made up of higher frequencies, which resembled more stationary EEG. Components 1-4 (Figure 6) represent activation patterns associated with ‘low’ clusters, whereas components 5-7 and 11-14 (Figure 6) are examples of ‘high’ activation patterns.
Components of each type were then clustered together and subsequent analyses were performed separately for each of the two clusters.

**Figure 6:** EEG Output from Single EEG Channel after Filtering and ICA (where ‘g1’ represents gait initiation, ‘b1’ first section of baseline continuous gait, ‘g2’ represents the turning events, ‘b2’ is the second section of baseline continuous gait, and ‘g3’ is gait termination. Similar events occur after the boundary break, except the conditions ‘o1’ and ‘o2’ represent the continuous walking sections of the orthotic condition. These are all components of one channel, i.e. AF3. All fourteen channels are composed of fourteen components, as they receive information from all other channels as well as there own.)

### 3.6 Statistical Analysis

The Statistical Package for the Social Sciences (SPSS) version 20 (SPSS Inc., Chicago, Illinois) was used to conduct all descriptive analyses. A descriptive table for the demographic characteristics of study participants was provided using means and standard deviations for continuous variables (age, height, weight) and proportions for nominal variables (leg and hand dominance, sex,). Paired t-tests (Dependent group t-test) were conducted using MATLAB to statistically compare the Power Spectral Density values between the two trials. All significant tests were two-sided with alpha set at 0.05. Since all data were collected at one time point, there is no missing or incomplete data.
There were 14 paired t-tests performed on the data, per cluster (high and low), testing the difference between baseline and orthotic condition. These values were averaged across all participants and EEG channels for each cluster. The researchers were aware that by completing multiple t-tests on the data that the chance of performing a Type 1 error was inflated proportional to the number of comparisons being made. A Bonferroni correction could have been applied to control for experiment wide error, to help reduce per comparison Type 1 error and make the estimate of significance more conservative. However, this pilot study was not conducted with the intention to have enough statistical power to estimate effects or assess statistical significance. Rather the t-tests were performed for exploratory reasons.
Chapter 4

4 Results

4.1 Demographic Information

Participant’s demographic characteristics are provided in Table 1. The mean age of participants was 26.4 years (± 4.3), and ratio of males to females was 5 males: 11 females.

Table 1: Participant Demographics

<table>
<thead>
<tr>
<th>Characteristics</th>
<th>Values (n=16)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gender (Male:Female)</td>
<td>5 : 11</td>
</tr>
<tr>
<td>Age (Years)</td>
<td>*26.4 ± 4.3</td>
</tr>
<tr>
<td>Height (Inches)</td>
<td>*67.2 ± 3.5</td>
</tr>
<tr>
<td>Weight (lbs)</td>
<td>*145.1 ± 30.7</td>
</tr>
</tbody>
</table>

* Mean ± Standard Deviation

4.2 Primary Outcome Measure

Raw EEG data produced from the Emotiv neuroheadset was the primary outcome measure of this study. Power Spectral Density (PSD) values were calculated and averaged for every participant for each of the two conditions. The Power Spectral Density of the average referenced, 5 second intervals, following each of the “b1”, “b2”, “o1” & “o2” markers were calculated using the Welch method (in which each 5 second interval was divided into 8, 50% overlapped and hanging windowed segments, of which the resulting periodograms were averaged together) (see Figures 5 & 6 in methods). The resulting spectrograms for the two sessions in each condition were averaged together to generate the final measure for each subject, by condition. The PSD value calculations were based upon the average across all EEG channels.
4.3 EEG Data Results

Data were separated by baseline condition and orthotic condition, as well as by “low” and “high” activation patterns. The “low” cluster activation patterns are associated with movement and environmental artifacts, whereas the “high” cluster patterns are associated more with the stationary EEG during walking. The PSD values recorded in Table 2 are the averaged spectral densities between 1-3 Hz for each participant at baseline and under the orthotic condition, for the “high” cluster activation pattern. The baseline values were derived from the average of “high” “b1” and “b2” values of each participant, and the intervention values from the average of “high” “o1” and “o2” values for each participant. There was one participant, marked with an asterisk in Tables 2 and 3, who was removed from data analysis. This participant was an extreme outlier, one that could potentially drive the data one way, so was therefore excluded from the analysis.
Table 2: Power Spectral Density Data for “high” Cluster (Average between 1-3 Hz)

<table>
<thead>
<tr>
<th>Subject ID</th>
<th>EEG Channel Data (Watts/ Hz)</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Baseline (no orthotic)</td>
<td>Intervention (orthotic)</td>
</tr>
<tr>
<td>1</td>
<td>335.731 ± 516.183</td>
<td>95.154 ± 128.467</td>
</tr>
<tr>
<td>2</td>
<td>28.094 ± 27.744</td>
<td>68.779 ± 93.428</td>
</tr>
<tr>
<td>3</td>
<td>5.303 ± 10.205</td>
<td>2.034 ± 3.630</td>
</tr>
<tr>
<td>4</td>
<td>78.089 ± 53.651</td>
<td>34.560 ± 24.232</td>
</tr>
<tr>
<td>5</td>
<td>194.447 ± 478.830</td>
<td>646.628 ± 1683.798</td>
</tr>
<tr>
<td>6</td>
<td>464.709 ± 364.645</td>
<td>914.205 ± 749.867</td>
</tr>
<tr>
<td>7</td>
<td>587.301 ± 519.092</td>
<td>500.388 ± 439.564</td>
</tr>
<tr>
<td>8</td>
<td>648.222 ± 1721.887</td>
<td>2068.240 ± 6116.052</td>
</tr>
<tr>
<td>9</td>
<td>150.739 ± 205.822</td>
<td>394.021 ± 591.227</td>
</tr>
<tr>
<td>10</td>
<td>53.508 ± 40.823</td>
<td>89.263 ± 72.704</td>
</tr>
<tr>
<td>11</td>
<td>71.076 ± 81.167</td>
<td>50.028 ± 57.683</td>
</tr>
<tr>
<td>*12</td>
<td>4289.545 ± 4177.625</td>
<td>14362.679 ± 14130.577</td>
</tr>
<tr>
<td>13</td>
<td>170.453 ± 167.930</td>
<td>104.733 ± 100.837</td>
</tr>
<tr>
<td>14</td>
<td>387.009 ± 390.924</td>
<td>584.124 ± 577.209</td>
</tr>
<tr>
<td>15</td>
<td>55.133 ± 91.204</td>
<td>38.795 ± 37.743</td>
</tr>
<tr>
<td>16</td>
<td>29.425 ± 34.034</td>
<td>24.646 ± 26.072</td>
</tr>
<tr>
<td>GRAND AVERAGE</td>
<td>217.282 ± 546.475</td>
<td>374.373 ± 1696.386</td>
</tr>
</tbody>
</table>

Mean ± Standard Deviation
* Subject was removed from statistical analysis
From these raw data for the “high” cluster, the grand average PSD values for the orthotic condition as well as the baseline condition are depicted in a Power Spectral Density Graph. In the PSD graph illustrated below in Figure 7, the bold red line represents the grand average PSD for the orthotic condition and the bold black line signifies the grand average for the baseline condition. The dim lines seen in the background correspond to the PSD values for each individual participant in each condition. All of these values were calculated from the average across all EEG channels. Most of the brain activity recorded existed in the regions from 0-5Hz, with little variation occurring above approximately 7 Hz.

![Power Spectral Density Graph](image)

**Figure 7:** Power Spectral Density for “high” Cluster (Units of Watts/Hertz)
(This figure shows where the average power of the signal from the high cluster waves is distributed as a function of frequency for both the orthotic and baseline conditions.)

Similarly, the PSD values recorded in Table 3 are the averaged spectral densities between 1-3 Hz for each participant at baseline and under the orthotic condition, for the “low” cluster activation pattern. The baseline values were derived from the average of “low” “b1” and “b2” values of each participant, and the intervention values from the average of “low” “o1” and “o2” values for each participant.
Table 3: Power Spectral Density Data for “low” Cluster (Average between 1-3 Hz)

<table>
<thead>
<tr>
<th>Subject ID</th>
<th>EEG Channel Data (Watts/ Hz)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Baseline (no orthotic)</td>
</tr>
<tr>
<td><strong>1</strong></td>
<td>97.422 ± 156.519</td>
</tr>
<tr>
<td><strong>2</strong></td>
<td>86.130 ± 84.950</td>
</tr>
<tr>
<td><strong>3</strong></td>
<td>82.721 ± 120.127</td>
</tr>
<tr>
<td><strong>4</strong></td>
<td>351.862 ± 427.659</td>
</tr>
<tr>
<td><strong>5</strong></td>
<td>7639.725 ± 22292.301</td>
</tr>
<tr>
<td><strong>6</strong></td>
<td>2578.164 ± 2146.437</td>
</tr>
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<td><strong>7</strong></td>
<td>4267.011 ± 4279.027</td>
</tr>
<tr>
<td><strong>8</strong></td>
<td>33955.247 ± 56904.198</td>
</tr>
<tr>
<td><strong>9</strong></td>
<td>2427.067 ± 3896.398</td>
</tr>
<tr>
<td><strong>10</strong></td>
<td>310.952 ± 251.371</td>
</tr>
<tr>
<td><strong>11</strong></td>
<td>82.981 ± 147.477</td>
</tr>
<tr>
<td><strong>12</strong></td>
<td>25421.285 ± 28106.926</td>
</tr>
<tr>
<td><strong>13</strong></td>
<td>36.607 ± 32.598</td>
</tr>
<tr>
<td><strong>14</strong></td>
<td>396.791 ± 375.104</td>
</tr>
<tr>
<td><strong>15</strong></td>
<td>114.653 ± 309.919</td>
</tr>
<tr>
<td><strong>16</strong></td>
<td>2734.567 ± 3554.410</td>
</tr>
<tr>
<td><strong>GRAND AVERAGE</strong></td>
<td>3677.460 ± 17482.970</td>
</tr>
</tbody>
</table>

Mean ± Standard Deviation
* Subject was removed from statistical analysis
From these raw data for the “low” cluster, the grand average PSD values for the orthotic condition as well as the baseline condition were depicted in a Power Spectral Density Graph (Figure 8). The bold red line represents the grand average PSD for the orthotic condition and the bold black line signifies the grand average for the baseline condition. The dim lines seen in the background correspond to the PSD values for each individual participant in each condition. All of these values were calculated from the average across all EEG channels. Most of the brain activity recorded in this cluster existed in the regions from 0-3Hz, with little variation occurring above approximately 5 Hz.

Figure 8: Power Spectral Density for “low” Cluster (Units of Watts/Hertz)
(This figure shows where the average power of the signal from the low cluster waves is distributed as a function of frequency for both the orthotic and baseline conditions.)

Paired t-tests were performed to compare baseline walking (no orthotic) to walking with orthotics at each frequency in the spectrogram at each recording site, and for each of the ‘low’ and ‘high’ independent component clusters. There were no statistically significant differences ($p < .05$) found between the two walking conditions in the ‘low’ cluster at any sensor site, across all frequency wavelengths. Similarly, in the
‘high’ cluster, there were no statistically significant differences discovered between the walking conditions at any sensor site, across all frequency wavelengths. It is interesting to note, however, that while the results were not found to be statistically significant, the values in the ‘low’ condition, those due primarily to environmental and movement artifact, were lower during the orthotic trial when compared to the baseline condition. Whereas in the ‘high’ cluster, there was an increase in activity levels discovered in the orthotic walking trial when compared to the baseline condition. A topographical effect map was generated by performing a t-test on the average of the frequency bins between 1Hz and 3Hz at each recording site, and then the resulting $p$ values were plotted. The probability topographic map for the ‘high’ cluster is displayed in Figure 9 below, whereas the probability topographic map for the ‘low’ cluster is shown in Figure 10.

![High Cluster Topographic Map](image)

**Figure 9:** Probability Topographic Map for ‘high’ Cluster
(representation of the $p$ values associated with the differences between the baseline and orthotic condition for the “high” cluster components averaged across all participants and components 5-7 and 11-14 for each channel.)
Figure 10: Probability Topographic Map for ‘low’ Cluster
(representation of the $p$ values associated with the differences between the baseline and orthotic condition for the “low” cluster components averaged across all participants and components 1-4 for each channel.)

These topographic maps represent the $p$ values associated with the differences between the baseline condition and orthotic condition in each cluster. A very light green colour would indicate a statistically significant difference between the two walking conditions. As demonstrated in Figures 9 there are no light green areas shown surrounding the EEG sensors. In Figure 10, although there are areas of green that would indicate significance, the $p$ values for these areas were above .05. These areas appear green because of the many values that may fall in that region on the colour gradient scale. However, there were no statistically significant differences between walking conditions in the low cluster either.

The topographic maps presented in Figure 11 exhibit the signal activity in the “high” cluster after separating the two gait conditions into baseline walking and orthotic walking. The differences in activation pattern are represented by the colour gradient, where a higher and wider range of signal activity is demonstrated in the orthotic walking condition.
Figure 11: Topographic Map for ‘high’ Cluster Depicting Baseline and Orthotic Signal Activity

The topographic maps presented in Figure 12 exhibit the signal activity in the “low” cluster after separating the two gait conditions into baseline walking and orthotic walking. There is more signal activity observed in the baseline condition of the “low” cluster when compared to the “high” cluster, which may be attributed to the movement and environmental artifacts detected. The signal activity appears over a broader range in the orthopedic condition, whereas it is more concentrated in the baseline condition.
4.4 Feasibility Results

A Pilot Study is a version of the main study that is run in miniature to test whether the components of the main study can all work together (Arain et al., 2010). It can serve as a guide in the design and implementation of a larger scale study, where components of the study deemed infeasible or unsatisfactory should be modified or removed altogether for the subsequent trial (Leon et al., 2011).

4.4.1 Recruitment

The poster advertisement utilized in this study for participant recruitment was very successful. Participants were able to view a brief summary of the study, as well as the inclusion and exclusion criteria. This was effective as a self screening tool, as the participant could pre-determine if they fit the listed criteria or not. If specific pathological groups or age groups are being targeted for recruitment, the advertisement would need to be altered and focused more towards these groups. The advertisement would also need to be located in areas where these individuals could view them.
4.4.2 Methods

Informed consent and orthotic moulding completed during the participants first visit, was accomplished in a timely and efficient manner. This visit only required approximately 10 minutes of the participants time and was carried out by a Canadian Certified Pedorthist.

The headset calibration procedure was completed using all mandatory steps outlined by Emotiv, which were all essential in acquiring good signal contact and transmission. Initially, there were a number of control factors that needed to be predetermined before trials could commence. These included, walking at a self-selected pace, instructing participants to refrain from gum chewing or talking during trials, as well as keeping their head facing forwards. This allowed for participant trials to be as consistent as possible and to remove any outside factors that could influence the data collection process. Also, to control for order bias, the order in which the two trials occurred for each participant were randomized such that half of the participants first completed the gait testing without the orthotics, while the other half first completed the gait testing with the orthotics inserted in their shoes.

Both trials were conducted in the same testing timeframe, eliminating the need for the participant to return for an additional day of testing. This study required very minimal time commitment from the participants, so as not to disrupt their daily schedules. With that being said, the time allotted for the gait trials needs to be extended, to capture a longer section of continuous gait for analysis. In the future, this may add ten minutes to the overall time requirement of the study participants.

4.4.3 Novel Technique of Data Capture and Intervention

The Emotiv wireless neuroheadset has been previously tested and proven to be a reliable and valid tool (Badcock et al., 2013). The set-up and recording of brain activity was straightforward, with easy to use software, and simple instructions to follow. This user-friendly device allows for minimal training or teaching before being able to use in a research setting. The results of this study have shown it is able to detect changes in brain activation patterns due to the applied intervention. The intervention, custom made
orthotics, was used as a tool to provoke changes in brain wave activation patterns that could be detected by the wireless EEG device. Although statistically significant results were not found, there were changes in brain activation observed due to the intervention, so this system need not be altered.

4.4.4 Outcome Measures

The raw EEG data produced from the Emotiv neuroheadset was processed and analyzed in MATLAB, along with the EEGLAB module and toolbox. Raw data was marked by eye, and separated into different sections, including gait initiation, gait 1, turning event, gait 2, and gait termination. This method, although satisfactory, may not have been the most efficient way of parsing the data. The data were filtered, isolating artifacts and types of activation patterns (low and high clusters). This proved to be beneficial in eliminating environmental and movement artifacts from the brain activation patterns, enabling the researchers to focus on the activation patterns due to the actual movements of gait. For future trials, to eliminate possible human error in marking the raw EEG data by eye, the use of auditory tones and event related potentials, ERP’s, would be helpful in labelling the sections where continuous gait occurs.
Chapter 5

5 Discussion

The purpose of this pilot study was to examine the feasibility of a novel technique of recording brain activity and establish if custom made orthotics alter the brain activity patterns recorded by the Emotiv EEG device during human gait. This study was not conducted to provide a meaningful effect size estimate for subsequent studies, nor was it intended to confirm or validate the efficacy of orthotic devices and their possible effects on the brain (Leon et al., 2011). This study was a necessary initial step in examining the use of the wireless Emotiv EEG neuroheadset during human gait, as well as investigating the testing procedures, methods, and intervention. Changes in brain wave activity between wearing and not wearing orthotic devices in the shoe were detected. Although the differences detected between the two conditions were not statistically significant, these results demonstrate the neuroheadset equipment is sensitive enough to detect change under the intervention condition.

The topographical maps depicted in Figures 11 and 12, related to the high clusters during baseline and orthotic trials and the low clusters during both trials, respectively. The high cluster topographic maps, associated with the stationary EEG during gait, suggest stronger signal activity in the frequency spectrum during the orthotic condition. However, in the low cluster topographic maps, linked with movement and environmental artifacts, there is less movement artifacts (signal activity) in the orthotic condition. This suggests, that with the orthotic device in the shoe, gait has become more efficient in eliminating movement artifact and increasing brain activity.

The topographic maps (Figure 11) indicated stronger signal activity in areas of the brain through the orthotic gait trial when compared to the baseline trial. The areas of concentration in the orthopedic condition appeared to span between the frontal lobe and the parietal lobe. However, in the topographic map depicting signal activity during the baseline condition, there was no real change or fluctuation in signal activity. Generally, the concentrated area spanning the frontal lobe is where thinking, planning and motor function occurs. Similarly, the areas in the parietal lobe are primarily involved in
somatosensory perception. These are the primary areas expected to be activated during human gait.

To our knowledge, there is no published research using the Emotiv neuroheadset to record brain activation patterns during human gait which could be compared to our results. However, the areas in the topographic maps which indicated stronger signal activity were similar to results found in other studies involving functional brain imaging. In an EEG study conducted by Wieser et al. (2010), they concluded that the primary somatosensory cortex, primary motor cortex and the supplementary motor area, all play essential roles in cortical control of human gait. Similarly, using SPECT technology, Fukuyama et al. (1997) discovered that the medial primary sensorimotor area, supplementary motor cortex, and a small region of the medial temporal lobe were significantly activated during locomotion. In addition to the areas reported by Fukuyama et al. (1997), Hanakawa et al. (1999) expanded upon the areas of increased brain activity to include premotor cortex areas, somatosensory association cortex, and cingulate areas. Using NIRS imaging, Miyai et al. (2001) as well as Suzuki et al. (2004), observed increased activity in areas corresponding to the medial portion of the primary sensorimotor region, prefrontal cortex and the premotor cortex. Furthermore, Hanakawa et al. (2008) using an fMRI, discovered that executive regions, the primary motor cortex and the sensory cortex, were active mainly during actual performance of the movement. Unfortunately, even though our data is unable to identify the exact cortical region concerned with the increased activity, the general area is consistent with other findings.

It has been proposed that sensory feedback plays a major role in human locomotion (Chen et al., 1995). Consequently, it is possible that the increased surface area contact provided by the custom orthotic device may have altered the amount of sensory feedback from the mechanoreceptors on the plantar surface of the foot. Changes in feedback pressure on the skin, deformation of the skin, vibration and stretching of the skin caused by the orthotic (Guyton & Hall, 2006) may have contributed to the increased activity in the somatosensory areas (parietal lobe) of the brain. Nigg et al. (1999) suggested that an optimal orthosis could improve musculoskeletal and neuromuscular performance during walking. Nigg et al. (1999) speculated this increase in performance
resulted from the increased awareness of where the foot and leg are in space due to this increased sensory feedback.

The detectable difference of average Power Spectral Density between the two conditions existed mainly in the regions from 0-5 Hz, with little variation discovered above approximately 7 Hz. This demonstrated that the neuroheadset was sensitive enough to detect changes in brain activity between the two conditions and able to distinguish between different frequency levels. The wave forms involved in the 0-5 Hz range include both Delta and Theta waves. Delta and Theta wave forms typically occur either in very deep sleep or during drowsy, meditative states in adults (Sadava et al., 2008). We expected to see wave forms in the Beta range (13-30Hz), which are closely linked with motor behaviour and are present during active movements (Sadava et al., 2008). These abnormal findings do not detract from the ability of the neuroheadset to detect changes in brain activity. However, it does warrant further investigation to explore the reasons why this type of brain wave was prominent during a conscious state involving significant amounts of movement. One explanation for findings in the Delta range could simply be due to the novel application of pressure to the plantar aspect of the foot and the effects of custom made foot orthoses, yet unknown. One common disadvantage or challenge associated with using EEG to record brain activity during walking is the amount of movement and biological artifacts detected during the trials. These artifacts negatively influence the accuracy of the EEG measurements due to an increased number of factors involved in the EEG signals. The results presented in this study were filtered, and the biological artifacts (eye blinks and eye movements), as well as movement artifacts (waveforms that mimicked gyro patterns) were removed. Since the data was as independent of biological and movement artifacts as possible, changes detected by the wireless neuroheadset closely represented the true differences between the two testing conditions, thus providing evidence in favour of the feasibility of this novel approach to record functional brain imaging.
5.1 Feasibility Study Strengths

5.1.1 Recruitment

The system used to recruitment individuals to participate in the study, by poster advertisement, was very effective. The recruitment posters were placed in high traffic areas, where numerous individuals had the potential to view them. The posters worked as a self-screening tool, as the inclusion and exclusion criteria were clearly stated on the advertisement.

5.1.2 Methods

The strengths of this study are mainly associated with its design. There were a number of control factors that needed to be stipulated before the gait trials even took place. Initially, the participants were instructed to walk at a self-selected pace throughout each trial, as opposed to timing steps to a metronome. The rationale behind controlling the cycle time is that many of the measureable parameters of gait vary with the speed of walking. Therefore, by using a controlled cycle time, it provides one means of reducing variability (Whittle, 2007). However, since we were not interested in the specific parameters of gait that the speed of walking influences, it was not necessary to implement this technique in our study. Furthermore, Zijlstra et al. (1995) found considerable differences in the gait of normal subjects between their ‘natural’ walking and their ‘constrained’ walking, during which the subject was required to either step in time with a metronome or to step on particular places on the ground.

The recording of EEG during walking is a challenge due to movement artifact and the potential to detect signals from a mixture of skin, muscle and nerve activity (van de Velde et al., 1998). Therefore, in order to control for as many of these circumstances as possible, we instructed the participants to refrain from chewing gum, talking during the gait trials, and moving their heads excessively. Also, in order to minimize visual stimulation experienced by each participant, all trials were conducted in the same laboratory, where the surroundings were identical for all participants.
5.1.3 Novel Technique of Data Capture and Intervention

The intervention involved in this study, the addition of an orthotic device to the shoe, did not have to be tested in a specific order. To control for order bias, the order in which the two trials occurred for each participant was randomized such that half of the participants first completed the gait testing without the orthotics, while the other half first completed the gait testing with the orthotics inserted in their shoes. Furthermore, the neuroimaging technique used in this study was completely non-invasive when compared to other techniques being used (SPECT, NIRS, fMRI, etc.). This study also required very minimal time commitment from the participants so as not to disrupt their daily schedules.

Lastly, the degree of correction in the orthotic device was kept constant among each participant. All orthotic mouldings and fabrications were completed by the same clinician and laboratory. Each device was moulded specifically for each participant in this study and was not just a generic off the shelf insert that may fit differently for each person. However, the degree of correction in each orthotic was kept consistent for each participant. The basic level of correction was used to ensure that the benefit of the orthotic device was kept equal across the study, so some participants were not experiencing a greater correction over others.

5.2 Feasibility Study Limitations

The majority of the limitations associated with this study were identified as the study progressed. Given that this was a pilot project, there were no previous studies to assist with the design that would have helped to eliminate any mistakes or errors that may have previously occurred.

5.2.1 Recruitment

Young, healthy volunteers were required to participate in this study. For future studies, the poster advertisement may need some modification to target different age groups or certain pathological groups that are being studied.
5.2.2 Methods

The first limitation was related to one of our exclusion criterion, the neurological impairments. Upon recruitment the participants were asked if they suffered from any neurological impairment. Before the orthotic moulding took place the clinician again asked the same question. However, some people may have more or less sensitive receptors on their feet and simply be unaware of it. Kekoni et al. (1989) suggested that the sensation levels on the plantar surface of the human foot within a normal population vary substantially. To ensure that this exclusion criterion was tested appropriately, we could have used 2 point discrimination testing or monofilament testing before continuing on with the study.

The small sample size of 16 participants was also a limiting factor in this study. The small sample size was likely a contributing factor in the study being underpowered, and possibly a reason why no statistically significant results were discovered. A larger sample size would contribute to improving the accuracy and precision of the EEG device in detecting changes in brain activity.

Another limitation included the duration of the gait trials. Although the distance of the walkway used for the trials appeared appropriate, the amount of time required to complete the tasks was inadequate when it came to analyzing the raw data. After the initial portion of the trial, which involved verbal cuing and gait initiation, as well as the turning portion of the trial was removed, the amount of gait left to analyze was approximately 6 to 8 seconds. The length of time spent walking needs to be increased so that a larger portion of the trial can be analyzed. With longer blocks of time, the data would likely be far more stable and have less variance across subjects.

5.2.3 Outcome Measures

The raw EEG data produced from the Emotiv neuroheadset was processed and analyzed in MATLAB, along with the EEGLAB module and toolbox. One drawback to this analysis is that there is no way to objectively dissociate the movement artifacts from the cortical EEG. It is likely that a large portion of the signal at the scalp is contaminated with movement artifacts.
Lastly, the independent component analysis (ICA) used to isolate artifacts and types of activation patterns (low and high clusters), can be a very powerful tool for handling complex artifacts in EEG recordings. However, its power increases with the number of channels included in the recording montage. For example, moving up to 32 channels from 16 would improve the performance of the ICA and provide better topographical information for the classifications of components. In this study, there may not have been enough EEG channels recording to obtain all the benefits of this analysis.

5.3 Directions for Future Research

Further investigation regarding the use of custom foot orthoses and brain activity should be pursued. This pilot study was designed to introduce and test the wireless EEG neuroimaging device and to examine the possibility of detecting a change in brain activation through an intervention. Future research should take into account the following:

A larger sample size could be beneficial in improving the precision and accuracy of the EEG device to detect changes in brain activity. It would also increase the power of the study, possibly allowing for a statistically significant difference between the gait conditions to be demonstrated.

Longer walking trials would be useful for the analysis of the raw EEG data. After gait initiation and turning sections were removed from the trials collected in this study, there was only approximately 6 to 8 seconds of data remaining for analysis.

It would be helpful to incorporate some form of auditory event related potentials (ERP’s), to identify the sections of raw data that correspond to appropriate sections of gait. The auditory ERP’s, usually in the form of high frequency tones (1000 Hz or 1200 Hz), could be used as cues in the raw data for processing and analyzing those sections of the trial.

Testing over a longer period of time, initially when orthotics were first received and then again 6 weeks and 12 weeks later, could demonstrate a training effect in the brain. This should help to clarify if the effects and differences observed in this study between
the two gait trials were due to the novelty of the device in the shoe or if the orthotics can actually produce a lasting effect in areas of the brain. This could also create activation patterns in different brain wave frequencies, where the differences between trials occur further along the frequency spectrum as a patient habituates to the new sensation of the device.

If future research provides evidence indicating that there is a statistically significant difference in brain activation patterns with orthotics versus no orthotics, this line of inquiry should be expanded to include individuals with pathologies. The current study was performed using young, healthy participants, but there are certain pathological groups of individuals that are known to have gait and balance issues. It is important to note however, that although the use of orthotic devices will not slow down or prevent the progression of the disease, orthoses may be beneficial in providing enhanced sensory feedback from the cutaneous receptors on the soles of the feet, which have been altered or damaged due to the disease process. By enhancing sensory feedback, individuals may establish a better sense of where their foot is in relation to their surroundings, have a better feel of the ground characteristics on that which they are walking on, and overall experience improvements in their balance and posture such that improvements in quality of life may be realized.

Specifically, three populations that may benefit from the use of orthotics include individuals with diabetes, Parkinson’s disease, and multiple sclerosis. For example, peripheral neuropathy is the most common long-term complication in diabetes and is involved in changes in a diabetic persons gait and posture. Due to sensory neuropathy, diminished proprioceptive thresholds in the joints, most notably the ankle, have been observed (Son et al., 2009). Another obvious clinical sign of sensory-nerve damage is insensitive skin and reduced plantar sensation (Pham et al., 2000). Therefore, an increase in sensory feedback from the plantar receptors in the foot may help to prolong or minimize some of the effects seen in their gait and posture, such as reduced walking speed, reduced stride length, increased plantar pressures and a heightened risk of falls (Hohne, 2012). Secondly, gait and balance issues, as well as freezing of gait, are common features associated with Parkinson’s disease and parkinsonism. If an orthotic
intervention could be provided early enough in the diagnosis, challenging events such as regulating step length, shuffling and dragging of feet, gait timing, and gait initiation (Nutt et al., 2011; Hanakawa et al., 1999) could potentially be improved with additional sensory feedback. Finally, while a complete loss of sensation is rare in multiple sclerosis, a somatosensory deficit is a common feature of the disease (Kelleher et al., 2010). Thus, providing enhanced sensory feedback from the sole of the foot may, prove beneficial to patients within this population.

It may also be of interest to incorporate a 3dimensional kinematic gait analysis system together with the Emotiv neuroheadset device to determine the brain activity patterns associated with specific phases of the gait cycle.

5.4 Conclusion

This study explored the feasibility of a novel approach to brain imaging and the recording of brain activity through a wireless EEG device on sixteen participants. The use of an orthotic intervention enabled researchers to determine if the wireless neuroheadset was capable of detecting differences in brain activity patterns between the two gait trial conditions. The preliminary findings of this pilot study, although not statistically significant, suggest the Emotiv device is sensitive enough to detect changes in brain activation patterns during human gait. Further research is required, with a larger sample, before definite conclusions can be made about the effects, if any, that custom made foot orthoses may have on power spectral density during continuous gait.
References


Byrne, J.H. (1997). *Neuroscience online: An electronic textbook for the neurosciences.* Houston, Texas: Department of Neurobiology and Anatomy at The University of Texas Medical School.


Appendices

Appendix A: Ethics Approval Form

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This is to notify you that The University of Western Ontario Research Ethics Board for Health Sciences Research Involving Human Subjects (HSREB) which is organized and operates according to the Tri-Council Policy Statement: Ethical Conduct of Research Involving Humans and the Health Canada/ICH Good Clinical Practice Practices: Consolidated Guidelines, and the applicable laws and regulations of Ontario has reviewed and granted approval to the above referenced revision(s) or amendment(s) on the approval date noted above. The membership of this REB also complies with the membership requirements for REB's as defined in Division 5 of the Food and Drug Regulations.

The ethics approval for this study shall remain valid until the expiry date noted above assuming timely and acceptable responses to the HSREB's periodic requests for surveillance and monitoring information. If you require an updated approval notice prior to that time you must request it using the University of Western Ontario Updated Approval Request Form.

Members of the HSREB who are named as investigators in research studies, or declare a conflict of interest, do not participate in discussion related to, nor vote on, such studies when they are presented to the HSREB.

The Chair of the HSREB is Dr. Joseph Gilbert. The HSREB is registered with the U.S. Department of Health & Human Services under the IRB registration number IRB 00000040.

This is an official document. Please retain the original in your files.
Appendix B: Letter of Information and Consent Form

Letter of Information

Title of Research: Investigating the Effects of Custom Made Orthotics on Brain Forms

Principal Investigator: Dr. Colin Dombroski, PhD, C Ped C

Introduction

The purpose of this letter is to provide you with the information required for you to make an informed decision about participating in this research.

You are being invited to participate in this research study because you are a healthy individual between the ages of 18 and 65 years old, do not already require the use of orthotics, and are free of any neurological impairment. The purpose of this study is to determine if custom made orthotics will alter the brain activity patterns observed during human gait.

Human walking is a very complex task that requires the coordination of multiple systems in the body, most importantly, the musculoskeletal system and the nervous system. Most of the knowledge about the neural mechanism of walking has been derived from studies on the nervous system of animals. However, it is unclear to what extent these findings can be applied to the control of human gait.

Recent developments in non-invasive brain imaging techniques have allowed us to investigate neural control while walking in humans. However, these non-invasive imaging studies have used functional magnetic resonance imaging (fMRI) or positron emission tomography (PET) scans, which have their limitations. Most importantly, a person is incapable of performing walking activities inside this type of machinery. Some of these shortcomings have been overcome by developing alternative neuroimaging techniques. These allow for the recording of cerebral activity while performing imaginary gait tasks or repetitive foot movements while inside an fMRI.

This research study will also involve a non-invasive, functional brain imaging technique, but it contains a revolutionary wireless neuroheadset, that is in fact a portable EEG device. This headset will allow for the collection of real time electrical brain activity while the participant is walking. This technology from Emotiv™ will allow us to gain direct information concerning brain activity during human gait.
Procedure

If you agree to participate in this study, you will be asked to make two visits to Western University. The first will involve the molding process of the custom orthotic, which will be completed by a Canadian Certified Pedorthist. A week to ten days later, on the second visit, you will receive your orthotics and testing will begin.

The initial set-up will require a fitting and calibration of the Emotiv™ wireless neuroheadset. Once calibration is complete, you will be asked to walk approximately 20 ft down a walkway and then back to the original starting position, at a self selected pace. The second gait trial will involve inserting the orthotics into your running shoes, and then completing the same task of walking 20 ft down the walkway and back. Testing will be conducted in the Interdisciplinary Movement Disorders Lab, located in Elborn College room 1545. There will be a total of 20 participants involved in this study.

In addition to participating in the two gait trials, we will also gather some demographic information, like your gender, partial date of birth, height and weight.

Risks

There are no known or anticipated risks associated with participating in this research study. There will be a Canadian Certified Pedorthist molding and fabricating all of the orthotics, so this should eliminate any potential harm or discomfort involving the orthotic.

Benefits

There are no direct benefits from participating in this study; however the information gathered will provide valuable insight into brain activity during voluntary human walking and may be beneficial to research involving gait pathologies.

Compensation

You will not be compensated for your participation in this research.

Voluntary Participation

Your participation in this research study is completely voluntary. You may refuse to participate, refuse to answer any questions or withdraw from the study at any time with no effect on your future care.
Confidentiality

All data collected from you will remain confidential and accessible only to the investigators of this study. The data collected will be protected by a username and password and will be stored in a secure, locked location in the Fowler Kennedy Sport Medicine Centre. You will be assigned a unique number that will be used for all of your information and data collection, that way any identifying information will not appear in the data analysis or publication. While we will do our best to protect your information, there is a remote chance that your information could be accessed or hacked, even with high levels of security. If this were to occur, we would inform you immediately. Representative of The University of Western Ontario Health Sciences Research Ethics Board may contact you or require access to your study-related records to monitor the conduct of the research.

Further Information

If you require any further information regarding this research project or your participation in the study you may contact the Principle Investigator, Dr. Colin Dombroski at [redacted] or the student researcher, Lindsay Carey at [redacted].

If you have any questions about your rights as a research participant or the conduct of this study, you may contact The Office of Research Ethics at [redacted] or by email: [redacted].

This letter is yours to keep for future reference.

Sincerely,

Dr. Colin Dombroski, PhD, C Ped C
Dr. Jeff Holmes, PhD., OT Reg. (Ont.)
Lindsay Carey, BSc, MSc Candidate
Consent Form

Project Title: Investigating the Effects of Custom Made Orthotics on Brain Forms

Study Investigator’s Name: Dr. Colin Dombroski, PhD, C Ped C

I have read the Letter of Information, have had the nature of the study explained to me and I agree to participate. All questions have been answered to my satisfaction.

Participant’s Name (please print): ________________________________

Participant’s Signature: ________________________________________

Date: ________________________________________________________

Person Obtaining Informed Consent: ______________________________

Signature: ____________________________________________________

Date: _________________________________________________________

Page 4 of 4       Version Date: February 14, 2013       Participant Initials: ________
Appendix C: Demographics and Contact Information

Contact Information and Demographics

Participant (Please Print):

____________________________  __________________________
Last Name  First Name

Home Phone: __________________________
Work Phone/Cell Phone: ________________
Email Address: _______________________

Date of Birth: ________ - _______

Gender: [ ] Male  [ ] Female

Height: [ ] feet  [ ] Inches
Weight: [ ] lbs

Dominant Side - Hand: [ ] Right  [ ] Left
Foot: [ ] Right  [ ] Left

Do you have any problems with your hips/knees/ankles? ________________________________

What is your occupation? ________________________________

Have you taken any medication today? (if yes, please specify) _________________________
Appendix D: Recruitment Poster

Title of Research: Investigating the Effects of Custom Made Orthotics on Brain Forms

Principal Investigator: Dr. Colin Dombrowski

Procedure
We are looking for healthy individuals to participate in this study. We are investigating the effects that custom made orthotics will have on brain activity patterns while performing normal walking gait. New non-invasive, functional brain imaging technology will be used to wirelessly transmit your brain activation patterns in real-time as you walk down a 20 ft walkway.

You will be required to attend two appointments on Western University campus. The first will involve the molding process for your custom made orthotic, completed by a Canadian Certified Pedorthist. A week to ten days later, you will receive the orthotics and the gait analysis will take place.

Inclusion Criteria
- Healthy individuals between the ages of 18-65
- Individuals free of neurological symptoms
- Individuals who are able to attend two appointments on Western University Campus

Exclusion Criteria
- Individuals who already require the use of orthotics
- Minors

Contact
If you have any questions about the study or to volunteer to participate please contact:

Lindsay Carey at [Contact Information]
LINDSAY CAREY  CURRICULUM VITAE

EDUCATION

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