Comparison of an ankle-foot-orthosis and neuroprosthesis during level and non-level walking for individuals post-stroke

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COMPARISON OF AN ANKLE-FOOT-ORTHOSIS AND NEUROPROSTHESIS DURING LEVEL AND NON-LEVEL WALKING FOR INDIVIDUALS POST-STROKE

by

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A Thesis submitted to the Faculty of the Graduate School,
Marquette University,
in Partial Fulfillment of the Requirements for
the Degree of Master of Science

Milwaukee, Wisconsin

December 2011

ABSTRACT
COMPARISON OF AN ANKLE-FOOT-ORTHOSIS AND NEUROPROSTHESIS
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This study used gait analysis to compare the efficacy of the two foot drop treatments (ankle-foot-orthosis and neuroprosthesis) and to contrast the stimulation control of the two different neuroprosthesis sensors during level and non-level ambulation of post-stroke individuals.

Eight subjects completed two gait analysis sessions, once while using a study-provided articulated AFO and the other while using a WalkAide. After four weeks of acclimation to the device, each subject performed two minute walking trials on a level, inclined and declined treadmill. Kinematic and heart rate data were collected for all sessions. Plantar pressure and WalkAide tilt, heel loading, and stimulation timing data were collected during WalkAide sessions. Temporal parameters, kinematic gait asymmetry (GA), toe clearance, as well as WalkAide stimulation reliability (StR) and stimulation timing, were computed from Vicon and WalkAide data. Wilcoxon signed rank and Friedman tests were conducted to identify significant differences between treatments and amongst treadmill orientations, respectively.

Significantly greater minimum and mean ankle GA during initial swing was observed during AFO versus WalkAide ambulation for all treadmill orientations. Further analysis revealed that the ankle range of motion between affected and unaffected limbs was significantly different for the AFO, but not the WalkAide. Review of functional assessments indicated that these differences in ankle GA during initial swing may be attributed to the rigidity of the AFO, not functional gains with the WalkAide.

Comparison of the two sensor control options yielded mixed results. More optimal StR values were observed during heel sensor-based stimulation by 25% of the post-stroke subjects. Changes in StI on non-level surfaces occurred with the tilt sensor, but not the heel sensor. However, progressively delayed StI timing from declined to inclined walking with the tilt sensor may result in increased ankle plantar flexion during PS, providing beneficial during inclined ambulation.

Improved kinematic gait symmetry and StI changes on inclined surfaces with the WalkAide may be beneficial for community stroke ambulators on inclined surfaces; however, the lower StR and increased stimulating timing variability raise concerns about safety of the WalkAide while using tilt sensor controlled stimulation. The WalkAide heel sensor may provide a consistent, safer substitute.

ACKNOWLEDGEMENTS

Michelle B. Gunther Gallagher, B.S.

Many people have contributed to the success of this project and deserve recognition for their efforts. I could not have begun this work without the support of Drs. Silver-Thorn, Stoeckmann and Starsky, as well as Tom Current, CP, and Craig Peters, DPT, all of whom helped refine the study problem and associated test protocol. I would also like to acknowledge these individuals, and Dr. John McGuire, for providing the opportunity to observe stroke survivors during classroom and clinical visits, providing valuable clinical insight.

This study could not have been conducted without the recruitment of research subjects and therefore I would like to acknowledge Dr. Sheila Schindler-Ivens, Tina Puglisi-Creegan, M.S.,CC-SLP, Monica Diamond, PT, MS, C/NDT, Tom Current, CP, Dr. John McGuire, Dr. Andy Starsky, and Dr. Tina Stoeckmann for their active involvement in subject recruitment.

I would also like to thank Tom Current, CP and Craig Peters, DPT for the many hours of fitting, programming, and adjusting subjects with the AFO's and WalkAide – and for their insight into post stroke gait.

I would also like to thank Dr. Gerald Harris for access to his Motion Analysis Laboratory at the Medical College of Wisconsin for subject testing, and Dr. Jason Long for his assistance with respect to IRB documentation, data acquisition and processing. Additionally I would like to thank Dr. Silver-Thorn, Ryan Inawat, B.S., Caitlin Burke, B.S., and Andrew Kusters for their assistance with data collection and processing. Finally, I would like to thank the Medical College of Wisconsin biostatisticians, especially Dan Eastwood, for their advice regarding the statistical analyses for this study.

I would also like to acknowledge my study subjects for their participation, personal stories, and time and patience during the fitting and gait analysis sessions.

I would also like to thank my friends and family for their support and encouragement, especially my sister, Kristen, who read numerous drafts of this thesis.

I would like to acknowledge the Biomedical Engineering Department and College of Engineering at Marquette University for their financial support for the study-provided AFOs and research materials. Finally, I would like to thank Hanger Prosthetics and Orthotics and Innovative Neurotronics for supplying the WalkAide units used in this study.

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CHAPTER 1: INTRODUCTION AND MOTIVATION

It is estimated that 6,400,000 stroke survivors are alive today, and that approximately 795,000 people experience a new or recurrent stroke each year [1]. A stroke, sometimes referred to as a "brain attack", can have a large impact on a person's ability to function and participate in activities of daily life; stroke is considered the leading cause of serious, long term disability in the United States [1]. One area that is often affected by stroke is a person's mobility. A common gait impairment for stroke survivors is the inability to dorsiflex the ankle during the swing phase of walking. Without compensation, this impairment causes the foot to drag along the floor (i.e. foot drop) during swing. An estimated 20% of stroke survivors suffer from foot drop [2]. This impairment limits mobility, increases instability and increases the risk of tripping or falling [3].

The traditional form of treatment for foot drop is an ankle-foot-orthosis or AFO. The AFO holds the ankle in a neutral position during swing, preventing the toe from dragging along the ground. However, the AFO may decrease the range of motion at the ankle, so gait pathologies may still exist when wearing an AFO. An alternative treatment is a neuroprosthesis that electrically stimulates the common peroneal nerve thereby activating the ankle dorsiflexors during swing. Since the neuroprosthesis potentially increases ankle range of motion, the use of a neuroprosthesis for post-stroke individuals with drop foot is hypothesized to improve gait symmetry and efficacy on both level and inclined/declined surfaces when compared to ambulation with an AFO. Improvement in symmetry and efficacy, as well as the ability to walk on inclined/declined surfaces, enhances a person's mobility and safety inside and outside the home.

Although neuroprostheses have been available for 10 years, insurance reimbursement for neuroprostheses such as the WalkAide is often denied; insurance companies claim that these devices are still considered experimental. Research that quantifies the benefits of the

neuroprosthesis over an AFO in terms of improving safety and facilitating community ambulation may therefore increase the accessibility of these devices.

In addition to insurance reimbursement, this research study may provide insight into design improvements of neuroprostheses during inclined and declined walking. The WalkAide, the neuroprosthesis used in this study, uses an accelerometer to measure the angle of the tibia (i.e. tilt sensor). During initial fitting, the tibial angles that correspond to “toe off” and “heel strike” during level walking are determined. During subsequent ambulation, the WalkAide initiates stimulation to the common peroneal nerve at this “toe off” angle and ceases stimulation when the “heel strike” angle is reached [4]. However, as the fitting is performed during ambulation on a level surface, changes in these swing event identifiers for inclined and declined surfaces are not commonly assessed. In addition to the tilt sensor, clinicians may alternatively program the WalkAide using a heel sensor, where stimulation is based on the force under the heel [4]. The WalkAide can currently be programmed using either the tilt accelerometer or heel sensor. During inclined and declined ambulation, the heel sensor-based programming may be less affected by treadmill orientation. Therefore, it may result in a more accurate detection of the swing phase, the period in which peroneal nerve stimulation is needed to assist in foot clearance.

The goals of this study are to compare the gait of individuals with foot drop due to stroke during level and non-level walking, while using a WalkAide (Innovative Neurotronics; Austin, TX) and an ankle-foot-orthosis (produced by Tom Current CPO, Hanger Prosthetics and Orthotics, Milwaukee, WI) and to evaluate the control sensors for peroneal nerve stimulation currently used for WalkAide programming. Measures of lower extremity joint kinematics (e.g. maximum ankle dorsiflexion during swing) and temporal parameters (e.g. cadence) will be calculated from gait analysis data and used to contrast gait symmetry between two treatments for level and non-level ambulation. Toe clearance will be determined from gait analysis data and used to evaluate treatment efficacy on level and non-level surfaces. WalkAide stimulation reliability and stimulation initiation and termination timing will be calculated using gait analysis

data and WalkAide tilt sensor and heel sensor data will be compared with stimulation data using Matlab; these measures were used for the evaluation of the current control sensors of WalkAide stimulation.

The three primary research hypotheses contrasting treatments and WalkAide sensors are:

1) The use of a neuroprosthesis for post-stroke individuals with drop foot will:

a) improve temporal and kinematic gait symmetry on both level and non-level surfaces compared to an AFO and

b) improve treatment efficacy (toe clearance) on both level and non-level surfaces compared to an AFO.

2) Non-level walking will adversely affect the WalkAide tilt sensor-based stimulation reliability and timing resulting in:

a) missed or extraneous stimulations during ambulation on non-level surfaces and

b) changes in stimulation initiation and termination timing during non-level ambulation.

3) The use of the WalkAide heel sensor versus tilt sensor for stimulation control will:

a) improve stimulation reliability during ambulation on non-level and

b) exhibit more consistent stimulation timing during ambulation on non-level surfaces.

This investigation will provide quantitative data via gait analysis that may enhance the control of neuroprostheses and will help document the efficacy of AFOs and neuroprostheses for treatment of foot drop, potentially improving patient access to these devices.

CHAPTER 2: BACKGROUND AND LITERATURE REVIEW

This section introduces background information relevant to the study motivation, research questions, project methodology and interpretation of the results of this study. Topics include: physiology and statistics on stroke, able-bodied gait, post-stroke gait, treatment of foot drop, previous studies investigating neuroprostheses, gait on non-level surfaces, and gait symmetry.

2.1 INTRODUCTION TO STROKE

2.1.1 Physiology of Stroke

A stroke, also known as a cerebrovascular accident (CVA), occurs when blood flow to the brain is interrupted and brain cells in the immediate area die due to lack of oxygen and nutrients (e.g. ischemia). Ischemia leads to brain cell death (infarction) and a fluid filled cavity (an infarct) replaces the dead cells. A stroke can be caused by either an occlusion of cerebral blood vessels, called an ischemic stroke, or rupture of a cerebral blood vessel, called a hemorrhagic stroke [5]. During a hemorrhagic stroke, the blood irritates the brain tissue, causing swelling. Additionally the blood flowing into the brain collects into a mass called a hematoma. The combination of the swelling and hematoma compresses and displaces brain tissue. For this reason, the initial effects of a hemorrhagic stroke may be more severe. However, hemorrhagic strokes are less frequent than ischemic strokes, with only 20% of strokes being hemorrhagic [5]. Not all cells die immediately. Many surrounding cells are left in a compromised state and so the time between onset and treatment for a stroke is critical in determining the level of disability caused by the stroke [5].

The immediate symptoms of stroke include: sudden onset of numbness or weakness, confusion or trouble speaking or understanding speech, difficulty seeing in one or both eyes, difficulty walking, dizziness, loss of balance or coordination, and/or severe headache with no

known cause [5]. Although many of these symptoms improve after treatment, the location, severity and duration of symptoms before initial treatment determine the patient's long term disabilities. Infarction can result in disabilities such as sensory disturbances, language problems, emotional issues, thinking and memory problems, and paralysis or motor control problems. Since one side of the brain controls the opposite side of the body, neurological complications often arise on the opposite side of the body than where the stroke occurred in the brain [5, 6].

2.1.2 Stroke Incidence

Stroke is the third leading cause of death and the leading cause of disability in the United States [7]. There are approximately 6.4 million stroke survivors in the U.S., and about 795,000 people experience a new or recurrent stroke each year [1]. Approximately 10% of stroke survivors recover almost completely, 25% recover with minor impairments, 40% experience moderate to severe impairments that require special care, 10% require nursing home or long-term facility care, and 15% die shortly after the stroke [7]. Physical, occupational, and speech therapy, as well as long-term care expenses, cost Americans billions of dollars each year. In 2010, Americans paid about $73.7 billion in stroke-related medical costs and disability rehabilitation [8].

Of the disabilities requiring rehabilitation, gait deviations occur in approximately 70% of people following a stroke, with up to 86% of patients admitted to rehabilitation unable to ambulate independently [2, 9]. Even after therapy, only 50% of stroke survivors manage to walk in the community [10] and two-thirds of those do so with limitations (e.g. cannot walk independently in a crowded shopping center) [11].

Many stroke survivors must overcome the challenges of paralysis and motor control complications to walk around the house or in the community. As a result, there are some distinct differences between able-bodied ambulation and post-stroke gait. General gait changes in terms of temporal and stride parameters, as well as energy cost, have been observed.

2.2 ABLE-BODIED GAIT

Before examining the gait complications caused by the occurrence of one or more strokes, it is necessary to review the gait of able-bodied individuals, particularly the phases of the gait cycle, temporal and stride parameters, and lower extremity joint kinematics.

2.2.1 Phases of the Gait Cycle

A gait cycle is defined by the successive recurrence of events; able-bodied gait is typically defined by consecutive heel strikes. The gait cycle can be broadly divided into two main phases, stance and swing. Stance phase is defined as the period between initial foot contact (heel strike) and ipsilateral toe off. For unimpaired ambulators, the stance phase is approximately 62% of the gait cycle. The swing phase is defined as the period between ipsilateral toe off and the ipsilateral heel strike, when the foot is not in contact with the ground; this phase continues for the remaining 38% of the gait cycle. These stance and swing phases can be further divided into 8 functional phases (Figure 1): stance: initial contact, loading response, mid-stance, terminal stance, pre-swing and swing: initial swing, mid-swing, and terminal swing [12, 13].

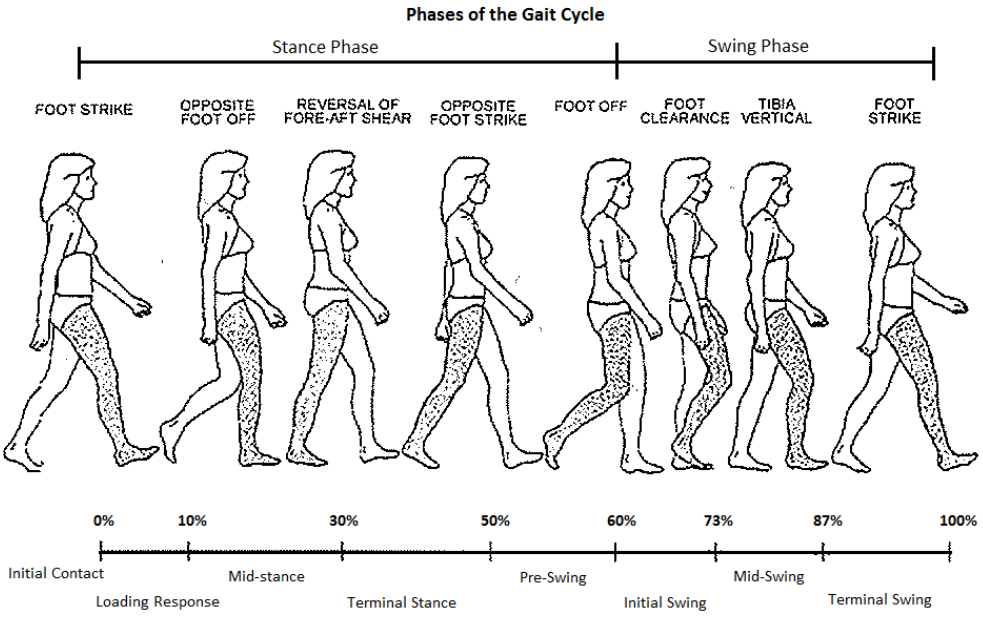

Figure 1: Functional phases of the gait cycle (adapted from [14]).

Initial contact, the first 2 % of the gait cycle, is defined by the instant when the foot first touches the floor [12, 13]. Although short, this phase is important because joint orientations during initial contact have a direct effect on the limb's loading response. The objective of this phase is to position the limb for stance or load-bearing. For unimpaired ambulators, the heel contacts the ground first, creating a rocker in which the heel acts as a fulcrum. The heel rocker continues into the next phase, loading response [12].

Loading response begins with heel strike and continues until the contralateral foot is lifted off the ground (e.g. contralateral toe off); the loading response duration for able-bodied individuals is 2-10% of the gait cycle. This phase is considered the first period of double limb support [12]. During this period, several key events occur to accomplish three main objectives: preservation of progression, shock absorption, and weight bearing stability [12, 13]. As stated previously, when the heel first contacts the floor, it acts as a fulcrum, allowing the foot to rotate at the ankle. The rapid loading of 60% body weight onto the stance limb creates an external plantar flexion torque. The tibialis anterior (dorsiflexor) contracts eccentrically, controlling the foot as it is lowered to the ground, preventing foot slap and prolonging heel support. Both the advancement of the tibia and the prolonged heel support preserve forward progression. Another benefit of the dorsiflexor muscle activity is shock absorption, since some of the body's downward motion is absorbed by the tibialis anterior as it resists the external plantar flexion torque [12]. Knee flexion during this initial loading response provides additional shock absorption, transmitting some of the energy to the eccentrically contracting quadriceps. The quadriceps are also responsible for weight bearing stability, resisting internal rotation of the tibia, the external adduction torque caused by limb loading, and preventing knee buckling (external knee flexion torque) [12].

Mid-stance, which occurs during the first half of single limb support (approximately 10-30% of the gait cycle), begins when the contralateral foot is lifted off the ground and continues until the body is aligned over the ipsilateral forefoot. Momentum from the contralateral limb,

which has just entered swing, and the forward fall of the body results in passive ankle dorsiflexion. This dorsiflexion allows the tibia to rotate over the foot (second rocker), contributing to further forward progression. To control the rate of dorsiflexion and stabilize the ankle, the soleus (plantar flexor) contracts eccentrically [12].

The latter half of single limb support is referred to as terminal stance phase; it occurs from approximately 30- 50% of the gait cycle. For unimpaired ambulators, this phase begins with ipsilateral heel rise and ends when the contralateral foot strikes the ground [12]. During terminal stance, the body advances beyond the supporting foot. The gastrocnemius and soleus muscles contract to stabilize heel rise, allowing the body to roll forward over the forefoot (3^{rd} rocker) [12, 13].

The final phase of stance is the pre-swing phase which begins with contralateral heel strike and ends with ipsilateral toe off (approximately 50-60% of the gait cycle). This is the second interval of double limb support. During this period, weight is transferred to the contralateral limb in preparation for ipsilateral swing. The decreased load on the limb results in rapid plantar flexion (approximately 20°). This plantar flexion causes the tibia to rotate anteriorly; as the toe is stabilized on the ground, knee flexion results. At the end of pre-swing, the tibialis anterior and toe extensor muscles are active in preparation for swing [12].

Initial swing phase (approximately 60-73% of the gait cycle) starts when the foot is lifted off the floor (ipsilateral toe off) and ends when the ipsilateral foot is opposite the contralateral stance foot [12](Figure 2). The main objective of this phase is toe clearance. To achieve toe clearance, the ankle dorsiflexes from its initial 20° plantar flexion to a more neutral position, due to tibialis anterior and toe extensor activity. Toe clearance is achieved by both this ankle dorsiflexion and knee flexion (nearly 60°) [12, 13].

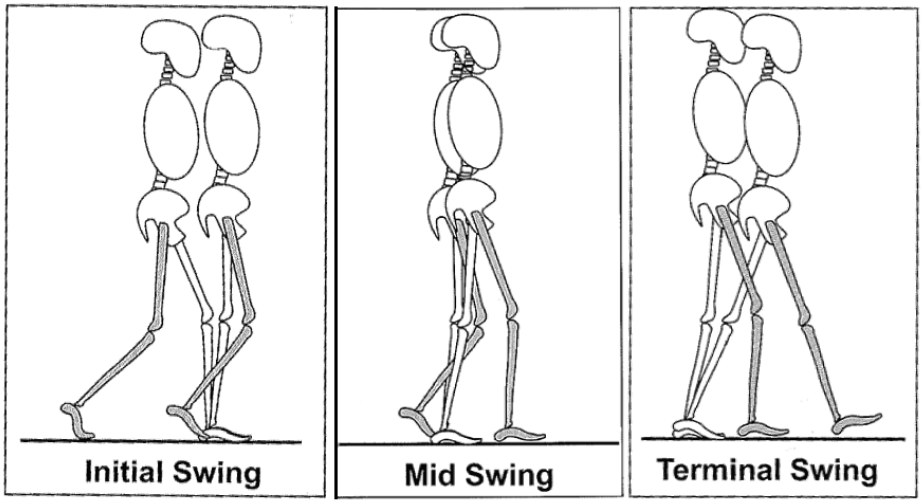

Figure 2: Functional phases of swing, illustrating both limb orientation at both initiation and termination of the respective phase (adapted from [12]).

The second phase of swing (mid-swing; approximately 73-87% of the gait cycle) begins when the ipsilateral swinging foot is next to the contralateral stance foot, and ends when the ipsilateral tibia is vertical (Figure 2). As the tibia approaches a vertical position, the mass of the foot places a greater demand on the ankle, and the activity of the tibialis anterior and extensor hallucis longus increase [12].

The final phase of the gait cycle (approximately 87-100%) is terminal swing. This phase begins when the ipsilateral tibia is vertical and ends when the ipsilateral foot strikes the ground (Figure 2). During terminal swing, the limb prepares for initial contact. Pretibial muscle action, especially tibialis anterior contraction, increases to counteract the inertia of the swinging leg, ensuring that the ankle will be neutrally positioned for subsequent heel contact. The knee extends in preparation for this initial contact [12].

2.2.2 Temporal and Stride Parameters

To evaluate timing of specific events and phase durations in the gait cycle, common temporal and stride parameters are used. The primary temporal parameters are cadence, stance and swing duration, as well as the duration of single limb and double limb support [14]. Cadence is the number of steps taken per unit time (steps/minute). As stated previously, for able-bodied

individuals, stance duration is approximately 62% of the gait cycle, while swing phase accounts for the remaining 38% [12-14]. These percentages and relative phase durations vary between individuals and with velocity [12].

The stance phase consists of two periods of double limb support, separated by a period of single limb support; swing phase occurs during single limb support, as shown in Figure 3. Single limb support defines the period in which only one limb is in contact with the ground; during double limb support, both feet are in contact with the ground. For normal, able-bodied gait, the first 12% of the gait cycle, from ipsilateral initial contact to contralateral toe off (loading response), defines the initial period of double limb support. This is followed by single limb support over the subsequent 38% of the gait cycle until contralateral heel strike (mid- and terminal stance). Late stance phase includes a second period of double limb support (50-62% of the gait cycle), from contralateral heel strike through ipsilateral toe off (pre-swing). The final single limb support period spans the entire swing phase (initial, mid-, and terminal swing), from ipsilateral toe-off through ipsilateral heel strike [12, 14].

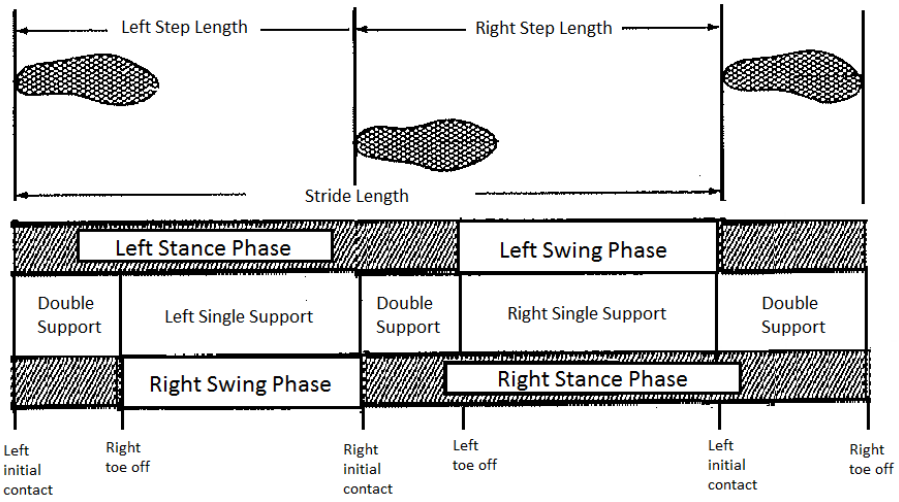

Figure 3: Temporal (stance, swing, single and double support duration) and stride parameters of gait (adapted from [14]).

The most common stride parameters examined during gait are step length, stride length, and velocity. Step length refers to the distance between heel strike of one limb and heel strike of

the contralateral limb, while stride length is the distance between subsequent heel strikes of the same limb (Figure 3). Velocity is defined as the distance traveled per period of time (meters/second) [12, 14].

2.2.3 Kinematic Parameters

Gait kinematics describe the movement of the lower extremities during walking; these kinematic measures are typically presented in terms of lower extremity joint (i.e. hip, knee and ankle) angles in the three respective planes of motion. During gait, the largest ranges of motion are observed in the sagittal plane. Therefore only sagittal plane kinematics are reviewed in this section.

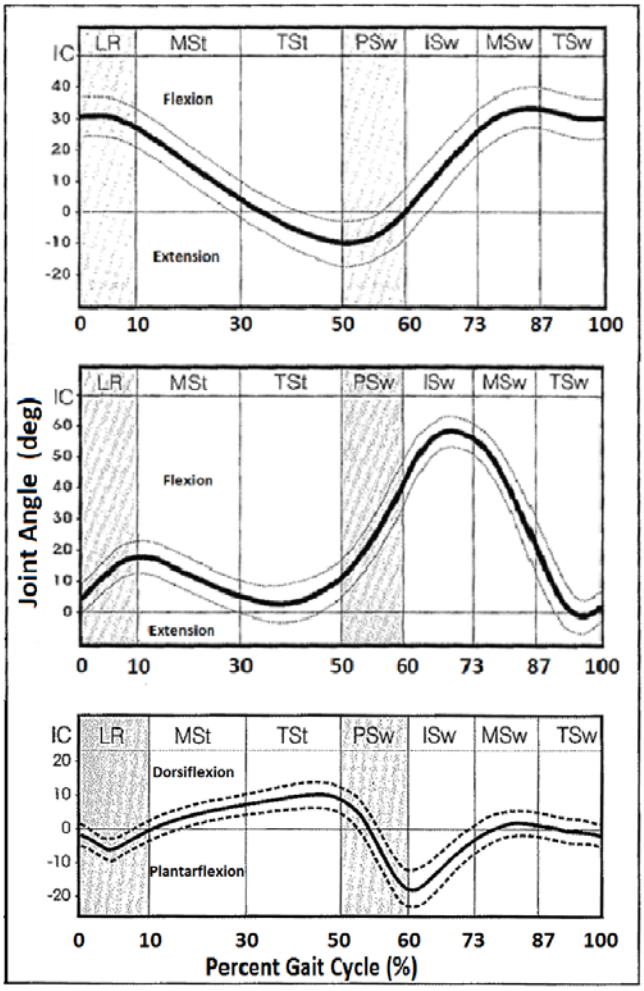

Figure 4: Sagittal plane motion of the hip (top), knee (middle) and ankle (bottom) during level walking for able-bodied subjects (adapted from [12]).

During able-bodied gait, each lower extremity joint displays a characteristic wave-form in the sagittal plane, as seen in Figure 4. Joint transitions occur between periods of flexion and extension throughout the gait cycle [14].

The normal range of motion of the hip during self-selected walking is 40°. At heel strike, the hip is flexed about 30°. The hip extends until the contralateral foot contacts the ground. As body weight is transferred to the contralateral limb during pre-swing, the ipsilateral hip then flexes in preparation for swing. Throughout swing the hip continues to flex, until terminal swing when the ipsilateral hip extensor muscles decelerate the limb in preparation for weight acceptance [12, 14].

As seen in Figure 4, the knee has the largest range of motion of the lower extremity joints (approximately 60°), and displays two periods of flexion. The initial knee flexion period occurs during early stance, with peak knee flexion (20°) occurring at the transition between loading response and mid-stance; this initial knee flexion reflects shock absorption that aids weight acceptance. During single limb support, the knee slowly extends to approximately 5° flexion during terminal stance. Knee flexion then rapidly increases following contralateral heel strike. Maximum knee flexion (approximately 60°) occurs during initial swing, allowing the limb to shorten and facilitate toe clearance. The combination of the inertial forces of the shank/foot and activation of the quadriceps muscles then results in rapid knee extension. Full knee extension is achieved just prior to heel strike [12, 14].

In contrast, sagittal plane motion of the ankle includes four inflection periods which relate to the three ankle rockers; ankle range of motion is approximately 25°. During weight acceptance, the neutral position of the ankle allows the heel contact with the floor. The foot uses the heel as a fulcrum, rotating to achieve foot flat (heel rocker). To provide both shock absorption and deceleration of the tibia, the ankle plantar flexes about 5°. After the forefoot contacts the floor, the tibia rotates about the ankle (ankle rocker), resulting in passive dorsiflexion. Dorsiflexion continues through single limb support (maximum dorsiflexion of approximately

10°). After contralateral heel strike, the center of mass of the body is over the metatarsal heads causing the ipsilateral heel to rise. The foot then rotates over the metatarsal-phalangeal joint (forefoot rocker), transitioning from 10° dorsiflexion to 15° plantar flexion. Finally, during swing the ankle rapidly dorsiflexes to provide foot and toe clearance. At the end of swing, the ankle is in a neutral position in preparation for heel strike [12, 14].

2.2.4 Ankle Muscle Activation

The major muscles controlling ankle motion are the dorsiflexors (tibialis anterior, extensor digitorum longus and extensor hallucis longus) and plantar flexors (soleus and gastrocnemius). Although the tibialis anterior, extensor digitorum longus and extensor hallucis longus all dorsiflex the ankle, the tibialis anterior is the primary dorsiflexor (60% of overall dorsiflexor muscle mass). The dorsiflexors are active during both swing and stance, in a biphasic fashion, with peak activation during initial swing (toe clearance) and loading response (prevent foot slap), as seen in Figure 5. In contrast, the plantar flexors are active during stance only, controlling the forward motion of the tibia via eccentric contraction and providing push off during mid- to terminal stance via concentric contraction [12].

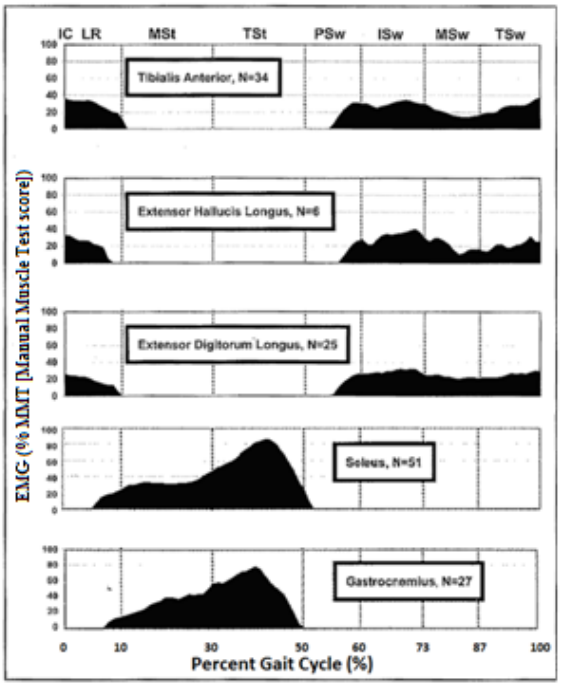

Figure 5: Activation of the ankle muscles during normal walking (adapted from [12]).

2.3 POST-STROKE GAIT

2.3.1 General Changes in Gait

Depending on the location and severity, a stroke may cause hemiparesis which results in the weakness and loss of some function on one side of the body, or hemiplegia which results in total paralysis of an arm, leg, and trunk on one side of the body. Hemiparesis and hemiplegia can affect an individual's gait. There is large inter-individual variability in temporal, kinematic and kinetic parameters of gait for post-stroke individuals [15] due to variations in the site, size, and type of lesion [9], as well as the time elapsed since the stroke [9, 16]. Without treatment, gait pathologies may contribute to reduced physical activity, impaired mental health, falls, fear of falling, frailty, and loss of independence [13].

2.3.2 Temporal and Stride Parameters

Many gait deviations following stroke result from the inability to generate muscle contractions [13], the inability to coordinate muscle contractions [13], and spasticity and joint

contractures that restrict joint mobility [17]. Studies have shown that post-stroke individuals generally have slower walking speeds [17-22] and reduced cadence [18, 21, 22]. In addition to decreases in velocity and cadence, post-stroke individuals have shorter stride and step length [13, 16, 21, 22], spend extended time in double limb support, and exhibit decreased step duration [16, 18, 21, 23] on the affected side. Due to muscle weakness and poor balance of the affected side, the swing time of the paretic (i.e. affected) limb is typically prolonged, with the non-paretic (i.e. unaffected) limb exhibiting prolonged stance duration and single limb support duration with respect to the paretic limb [13]. Post-stroke patients also require more energy to ambulate the same distance when compared to age-matched, healthy subjects [24-25]. The differences in step length, swing time, muscle strength, and range of motion between the affected and unaffected limbs result in an asymmetric gait pattern [16, 20] and increased energy consumption [26].

2.3.3 Gait Kinematics and Foot Drop

In addition to temporal gait parameters, changes in lower extremity joint kinematics are also observed in post-stroke individuals. Post-stroke individuals walking at slow, medium and fast speeds exhibited decreased range of motion of the affected hip, knee and ankle compared to the unaffected limb [23]. Specific changes in joint kinematics were dependent on muscle weakness and/or spasticity and joint contractures specific to the individual. Kinematic changes included: increased/reduced hip flexion at initial contact [27], knee hyperextension during stance [27-28], increased plantar flexion at initial contact (foot flat) [27-29], reduced ankle plantar flexion at toe off [29], reduced dorsiflexion or continuous ankle plantar flexion during swing [28-29], and reduced knee flexion during swing [27-29].

One of the more common changes in joint kinematics during gait for individuals post-stroke is foot drop (the inability to dorsiflex the ankle during the swing phase of walking). Without compensation, this impairment causes the foot to drag along the floor. There are three common compensatory gait patterns for post-stroke individuals with foot drop: 1) steppage gait,

2) vaulting, and/or 3) circumduction. Increasing knee and hip flexion to facilitate toe clearance is effective when there is sufficient muscle strength on the affected side. Insufficient hip and/or knee muscle strength may cause the individual to vault or circumduct the lower limb to assist contralateral foot clearance during swing. These compensatory gait mechanisms result in joint kinematic asymmetry and require more energy for each step. Additionally, these compensatory methods may not consistently yield sufficient toe clearance, and may increase the risk of falling [30, 31]. Therefore, patients with foot drop require some form of treatment to improve foot clearance and decrease their risk of falling.

2.4 TREATMENTS FOR POST-STROKE INDIVIDUALS WITH FOOT DROP

2.4.1 Ankle-Foot-Orthosis

Ankle-foot-orthoses (AFOs) are among the most commonly prescribed lower extremity limb orthoses [31] and are the primary treatment for foot drop [32]. Thermoplastic AFOs are more common than metal AFOs due to decreased mass, lower cost, and increased durability. These AFOs incorporate a three-point loading configuration to encourage dorsiflexion and minimize plantar flexion during swing (Figure 6). This is accomplished with either a solid (fixed ankle) or articulated (ankle motion permitted) AFO.

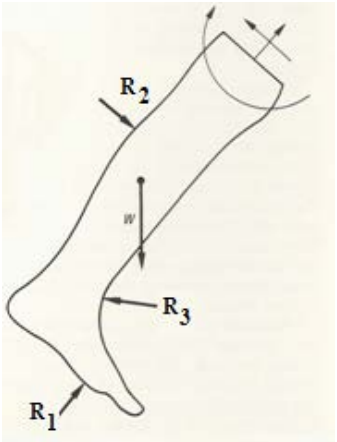

Figure 6: Three-point force system for an AFO [31]
Forces R1 and R2 are provided by the foot shell and proximal strap; force R3is provided by the shoe and/or anterior strap.

An articulated AFO (Figure 7) may also incorporate dorsiflexion assist bands or springs, as well as plantar flexor stops. Although the articulated AFOs permit greater movement of the ankle than a solid AFO, they may still limit the full range of ankle motion [31].

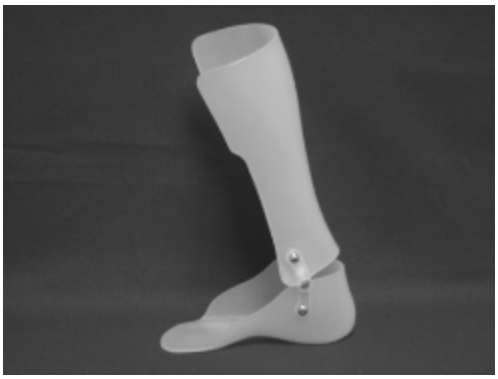

Figure 7: Sample thermoplastic articulated AFO [31].

2.4.2 Neuroprostheses

In 1961, Liberson and colleagues proposed an alternative treatment for foot drop, a Drop Foot Stimulator [33], or neuroprosthesis that stimulates the common peroneal nerve to activate the anterior tibialis and dorsiflex the ankle. This technology has since been further developed and today there are three FDA approved neuroprostheses: the Odstock Dropped Foot Stimulator (Department of Medical Physics and Biomedical Engineering, Salisbury District Hospital, Salisbury, UK), the NESS L300 (Bioness Inc., Valencia, CA), and the WalkAide (Hanger Orthopedic Group/Innovative Neurotronics, Bethesda, MD) [34].

2.4.3 The WalkAide

Both the Odstock Dropped Foot Stimulator and the NESS L300 incorporate a foot switch placed in the shoe to determine the stimulation periods for peroneal nerve stimulation during gait [34]. In contrast, the WalkAide uses an accelerometer to measure the angle of the tibia of the affected limb (e.g. tilt sensor) and define the stimulation timing [4]. During initial fitting, the clinician uses the WalkLink, a WalkAide programming interface, to manually initiate peroneal nerve stimulation as the patient is walking. The manual stimulation allows the patient to walk

with the WalkAide before it has been programmed and allows the clinician to acquire tibial angle data from the accelerometer. (Heel sensor data from the optional WalkAide heel sensor placed in the shoe can be similarly acquired.) This manual stimulation allows the clinician to determine the tibial angles that correspond to "toe off" and "heel strike" during level walking. During subsequent ambulation, the stimulation of the common peroneal nerve is initiated at this "toe off" angle and ceases when the "heel strike" angle is reached [4].

2.5 INVESTIGATION OF POST-STROKE GAIT WITH NEUROPROSTHESES

Prior studies of various neuroprostheses for the treatment of foot drop have shown that these devices improve gait. Improvements are most often quantified in terms of increased walking speed and stride length [32, 35-39], as well as decreased physiological cost index (PCI) which reflects energy consumption and effort [37- 38]. Although walking speed, stride length and energy are important measurements for rehabilitation, they are limited in their ability to identify functional changes, such as increased joint motion, which may facilitate additional activities of daily living (e.g. stair ascent). Therefore studies have also investigated changes in functional ambulation scale scores [40], gait asymmetry index (i.e. comparison of affected and unaffected swing times) [38-39], and stride time variability [38-39]. Most studies have examined changes before and during use of the neuroprosthesis, with only two studies comparing gait parameters between the neuroprosthesis and the traditional AFO treatment [39-40].

Scheffer et al. compared functional ambulation scale scores during ambulation with no treatment, neuroprosthetic treatment (Odstock Dropped-Foot Stimulator) and a previously issued articulated AFO (with dorsiflexion assist and plantar flexion stop) for 14 chronic post-stroke individuals [40]. Although significant differences in functional ambulation scores were observed between no treatment versus each treatment, differences in scores between the two treatments were not significant. However, subjects in this study had only two days to acclimate to the neuroprosthesis prior to evaluation. In a similar study, Ring et al. noted that 8 weeks (versus 4

weeks) acclimation was necessary to observe significant differences in stride time, gait symmetry, and swing time variability between neuroprosthesis and AFO treatments; no significant differences in walking speed were observed for these 15 chronic post-stroke subjects, regardless of acclimation period [39]. Given sufficient acclimation, Ring et al. concluded that a neuroprosthesis can enhance balance control and manage foot drop more effectively than an AFO for chronic post-stroke individuals with foot drop [39].

2.6 LIMITATIONS OF PREVIOUS POST-STROKE GAIT STUDIES

2.6.1 Walking Surface

One limitation of prior neuroprosthesis studies is that only level walking trials were conducted, although non-level walking surfaces are routinely encountered during household and community ambulation. Gait kinematics and kinetics for unimpaired ambulators have been found to vary during ambulation on inclined (5^o, 8^o, 10^o) and declined surfaces (5^o, 8^o, 10^o), as shown in Figure 8 [41]. Significant increases in hip flexion at heel strike, knee flexion at heel strike and during early stance, ankle dorsiflexion at heel strike and during stance, and plantar flexion during toe off were observed for inclined walking trials. Significant increases in peak knee flexion and decreases in minimum knee flexion during stance, increases in maximum knee flexion during swing, and increases in ankle dorsiflexion during stance were observed for declined walking trials.

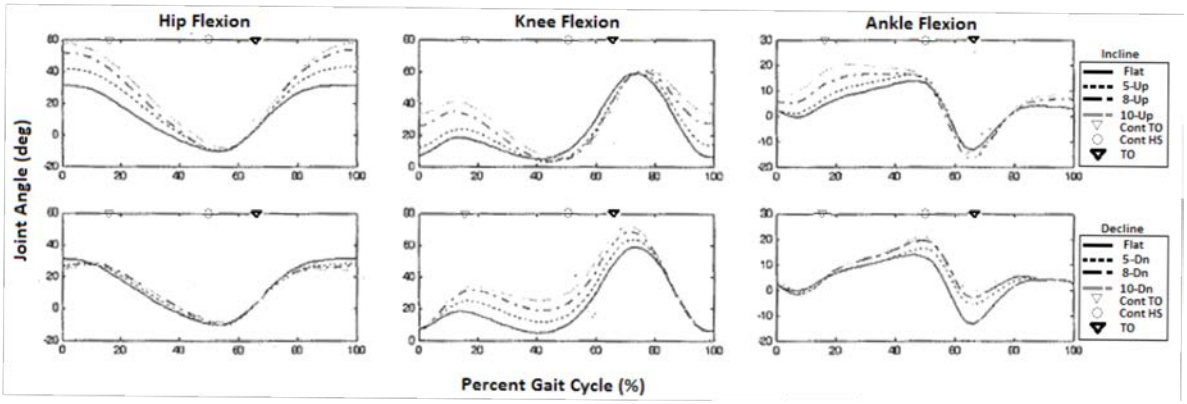

Figure 8: Lower extremity sagittal plane joint kinematics of normal subjects during walking on non-level surfaces (adapted from [41]).

Accommodation to non-level walking surfaces may be problematic for individuals post-stroke with limitations in lower extremity range of motion; these non-level surfaces may be particularly problematic for post-stroke individuals with foot drop who have difficulty with foot clearance and are at risk of falling. For post-stroke individuals using the WalkAide, these non-level surfaces may affect the angle of the tibia at heel strike and toe off, adversely affecting neuroprosthesis stimulation.

2.6.2 Measurement Parameters

Previous studies involving neuroprostheses to treat post-stroke individuals with foot drop have focused on walking speed and PCI as primary functional measures. However, measures such as walking speed provide limited insight into the control of walking. Additionally, walking speed is only one measure of clinical relevance regarding rehabilitation progress. For example, rehabilitative functional goals may also include symmetric weight bearing and improved weight transfer between the affected and unaffected limbs during gait [42]. Further quantification of gait has potential clinical value and may provide direction for future care or therapy.

2.6.2.1 Significance of Gait Symmetry

Quantifying gait symmetry may provide enhanced understanding of rehabilitation progress and functional recovery. Kim et al. have shown that normal gait is symmetric both spatially and temporally [42]. Therefore, if the goal of rehabilitation programs is continual progress toward normal ambulation, then gait symmetry is a more descriptive clinical functional measure than walking speed or PCI. Gait symmetry is also linked to energy expenditure with greater symmetry associated with greater energy efficiency [26]. Patterson et al. have suggested that the increased energy cost during ambulation for post-stroke individuals can be attributed to their asymmetric gait [26]. Gait asymmetry may also reflect balance instability, risk of musculoskeletal injury to the unaffected lower limb, and loss of bone mass in the affected lower limb [43]. Comparison of gait asymmetry may therefore assist in differentiating the efficacy of various treatment methods.

2.6.2.2 Measures of Gait Symmetry

Gait symmetry investigations have focused on temporal and stride parameters. Both a symmetry ratio (SR) and a symmetry index (SI) have been proposed as measures with which to assess symmetry during gait. The SR is defined as:

$$SR = \ symmetry\ ratio\ = \frac{V_{affected}}{V_{unaffected}} \qquad \text{(Eq 1)}$$

where $v_{affected}$ is the value of the respective gait parameter on the affected side and $v_{unaffected}$ is the value of the gait parameter on the unaffected side. Patterson et al. recommended use of the SR for standardization of temporal and stride parameters, such as step length, swing and stance times and double limb support time [43]. However, for kinematic parameters, the SR does not account for sign changes in joint angle (e.g. flexion versus extension). The SR magnitude also fails to account for differences in range of motion that may occur between subjects or joints.

An alternative measure of symmetry, the SI, is defined as:

$$SI = symmetry\ index = \frac{V_{affected} - V_{unaffected}}{0.5\ (V_{affected} + V_{unaffected})} X\ 100 \quad \text{(Eq 2)}$$

where $v_{affected}$ is the value of the respective gait parameter on the affected side and $v_{unaffected}$ is the value of the respective gait parameter on the unaffected side. The SI has also been used to examine temporal and stride parameter symmetry; and to investigate neuroprosthesis treatment of post-stroke subjects with drop foot [38]. However, like the SR, the SI fails to capture sign differences in joint kinematics. While normalized to facilitate potential inter-subject and/or inter-joint comparisons, both the unaffected and affected limb parameters are incorporated into the normalization. As such, the SI may be more appropriate for symmetry investigations of able-bodied individuals. Even for able-bodied subjects, the SI does not provide sufficient resolution to examine small inter-limb differences and cannot differentiate the performance of each limb with large versus modest asymmetries [44].

Other symmetry measures used in gait analysis of able-bodied and post-stroke subjects (see Table 1) include: the non-normalized arithmetic difference between parameters, a logarithmic measure of gait asymmetry (GA), symmetry angle (SA), and two measures that include comparison of constrained versus free joint data (via joint bracing) with able-bodied data [symmetry rejoins of deviation (SROD) and individual rejoins of deviation (IROD)]. None of these measures, however, appear viable for symmetry investigations of treatment options of pathologic populations, kinematic analysis with potential variations in sign, and scaling or normalization to facilitate inter-subject and inter-joint comparison.

Table 1: Review of measures used to quantify gait symmetry

Reference	Symmetry Measure	Population	Parameters Investigated
Brouwer et. al.,2009 [45]	$V_{affected}$-$V_{unaffected}$	Post-stroke	Spatiotemporal joint angles and ground reaction forces
Patterson et. al., 2009 [43]	SR =$V_{affected}$ /$V_{unaffected}$; SI=($V_{affected}$- $V_{unaffected}$)/0.5($V_{affected}$ +$V_{unaffected}$) X 100%; GA=|100 X [ln($V_{affected}$/ $V_{unaffected}$)]|; SA=[(45^0-arctan($V_{affected}$/ $V_{unaffected}$))x 100%]/90	Post-stroke and able-bodied	Spatiotemporal
Robinson et. al., 1987 [46]	SI=($V_{affected}$- $V_{unaffected}$)/0.5($V_{affected}$+ $V_{unaffected}$) X 100%	Able-bodied	Ground reaction forces
Patterson et. al., 2008 [26]	SR = $V_{affected}$/ $V_{unaffected}$	Post-stroke	Spatiotemporal
Herzog et. al. 1989 [47]	SI=($V_{affected}$- $V_{unaffected}$)/0.5($V_{affected}$+ $V_{unaffected}$) X 100%	Able-bodied	Ground reaction forces
Shorter et. al., 2008 [48]	SROD and IROD	Able-bodied	Joint angles

SR=symmetry ratio, SI= symmetry index, GA=gait asymmetry, SA=symmetry angle, SROD=symmetry regions of deviation, IROD=individual regions of deviation.

2.7 SUMMARY

Stroke, the leading cause of long term disability in the United States, affects 6.4 million people in the United States alone, with approximately 795,000 new strokes each year. Stroke can have a large affect on the ambulation of individuals, with gait deviations occurring in approximately 70% of the post-stroke population.

Changes in gait are often caused by the inability to generate muscle contractions, the inability to coordinate muscle contractions, and spasticity and joint contractures that restrict joint mobility. Foot drop is a neurological problem in which the ability to dorsiflex the ankle is restricted or diminished. Two treatments currently used for foot drop are an AFO, a plastic brace which restricts the range of motion around the ankle, and a neuroprosthesis which electronically stimulates the nerve causing ankle dorsiflexion.

Prior research has shown that both AFO and neuroprostheses can improve gait in terms of walking speed, energy consumption, spatiotemporal gait symmetry and functional ambulation scale scores. This research, however, has been limited to ambulation on level surfaces and

investigation of spatiotemporal measures. More thorough analysis of ambulation on level and non-level surfaces, including measures of kinematic symmetry, may further quantify treatment efficacy.

CHAPTER 3: METHODS AND DATA PROCESSING

Due to limitations in prior research involving neuroprostheses to treat foot drop in post stroke individuals, the goal of this research study was to conduct more thorough analyses of post-stroke ambulation, contrasting AFO and neuroprosthesis treatments, on level and non-level surfaces. The specific hypotheses of the study were to determine if 1) use of a neuroprosthesis for post-stroke individuals with foot drop improves temporal and kinematic gait symmetry and treatment efficacy on level and inclined/declined surfaces when compared to an AFO, 2) non-level surfaces cause missed or extraneous stimulations or changes in the stimulation timing while using the WalkAide tilt sensor, and 3) use of heel sensor-based versus tilt sensor-based stimulation leads to improved stimulation reliability and more consistent stimulation timing in neuroprosthetic control for ambulation over non-level surfaces. To achieve these research objectives, gait analysis trials were conducted. Specific measures of interest included: temporal parameters, gait symmetry and asymmetry, and toe clearance (objective 1), as well as stimulation reliability and timing (objective 2 and 3).

This chapter summarizes these gait analysis procedures, including subject selection, experimental groups and treatments, and gait analysis protocol, as well as the data processing procedures for kinematic, gait parameter, WalkAide , and F-scan data. The statistical analysis procedures are also summarized.

3.1 SUBJECT SELECTION

3.1.1 Inclusion and Exclusion Criteria

Eight post-stroke individuals with the ability to actively ambulate within the community were selected from the greater Milwaukee area for participation in this study. This study was approved by the Institutional Review Boards (IRB) of Marquette University and the Medical College of Wisconsin. The following criteria were used to determine participant eligibility:

Inclusion Criteria:

- Hemiplegia due to stroke
- At least 6 months post stroke
- Passive ankle range of motion to neutral position
- No prior use of a neuroprosthesis other than evaluation or trial use
- Capable of walking 30 meters without stopping to rest and without the use of a cane or walker

Exclusion Criteria:

- Lower limb Botox injection within the past 6 months
- History of falls in the last 3 months while walking on level ground
- Pacemaker, defibrillator, or any electrical or metallic implant
- Cognitive disability due to stroke
- Prior history of seizures
- Peripheral nerve disease or Guillian Barre syndrome
- Fractures, dislocations or cancers of the affected lower limb
- Pregnancy

3.1.2 Subject Recruitment

Recruitment took place through fliers distributed at stroke support groups and in clinics[1], referrals by physicians[2] and professors[3], and by word of mouth via research subjects and clinicians[4] directly involved in the study. Referrals and individuals recruited by word of mouth gave written or verbal permission to forward their contact information to research staff. Interested parties were then contacted by phone, screened to determine if he or she met the inclusion criteria, and given a more detailed explanation of the study and study procedures. Informed consent was solicited prior to participation in any research activities.

[1] Department of Physical Medicine and Rehabilitation at Froedtert Memorial Lutheran Hospital, Sacred Heart Rehabilitation Institute at Columbia St. Mary's Hospital, Speech Pathology Department at Marquette University, Therapy Services at the Milwaukee Center for Independence.
[2] John McGuire, MD, Physical Medicine & Rehabilitation, Medical College of Wisconsin.
[3]Tina Stoeckmann, Ph.D. and Sheila Schindler-Ivens, Ph.D., Physical Therapy, Marquette University.
[4] Tom Current ,CPO (Hanger Prosthetics & Orthotics) and Craig Peters, DPT (Innovative Neurotronics)

3.2 EXPERIMENTAL GROUPS AND TREATMENTS

The study involved testing two treatment options for foot drop, a study provided AFO and a WalkAide neuroprosthesis. The treatment order was random, e.g. subjects were randomly assigned to an experimental group (Group 1 was tested with the AFO first; Group 2 was tested with the WalkAide first).

The first research visit for subjects in Group 1 (AFO, then WalkAide) consisted of casting and fitting the AFO by the study orthotist (Tom Current, CPO). After acclimating to the study-provided AFO for four weeks, the subject completed gait analysis in that AFO. Within a week of this gait session, the subject was fitted with the WalkAide by the study physical therapist or PT (Craig Peters, DPT). After four weeks of acclimating to the WalkAide, a second gait analysis session was completed using the WalkAide.

The first research visit for Group 2 (WalkAide, then AFO) subjects was the WalkAide fitting by the PT. During the next four weeks, the subject acclimated to the WalkAide and met with the orthotist for AFO casting. At the end of this period, the subject completed gait analysis with the WalkAide. Within a week of this gait session, the subject visited the orthotist for AFO fitting. After four weeks of acclimating to the study-provided AFO, the subject completed their final gait analysis session in the AFO.

3.2.1 AFO Fabrication and Fitting

The subject was casted for AFO fabrication by the study orthotist. The orthotist wrapped fiberglass casting material around the subject's affected lower leg from just distal to the knee down to the subject's toes. Before the casting material hardened, the leg was positioned such that the ankle was neutral (90^{o}); the ankle was held in this position until the cast set. This cast served as a negative mold of the subject's affected lower leg. The orthotist then filled this negative mold with plaster to produce a positive mold of the subject's lower leg. A thermoplastic (3/16 inch

polypropylene) articulated AFO was vacuum-formed over the positive mold. Ankle joints (Tamarack Flexure Joints, Model #740-M, Becker Orthopedic, Troy, MI) and a plantar flexion stop (TC Ankle Stop, Model #101TCA,Creative Adaptations, Milton, NH) were positioned on the articulated thermoplastic AFO, see Figure 9. The plantar flexion stop was set so that the tibia was vertical while standing. The foot plate was trimmed at the level of the toe sulcus; the proximal AFO trimline was no less than 2 cm inferior to the fibular head and no greater than the level of the fibular head.

After fabrication, the subject met with the orthotist for a second time. During this meeting, the subject ambulated within parallel bars with the study-provided AFO. Adjustments were made to the AFO as needed to improve comfort.

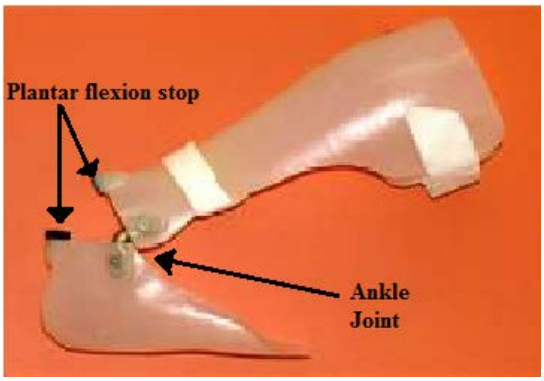

Figure 9: Study-provided AFO.

3.2.2 WalkAide Fitting and Programming

At the first visit for the WalkAide, the PT fit and programmed the WalkAide using the Walklink and WalkAnalysis software. Using a peripheral nerve stimulator (Innovative Neurotronics, Bethesda, MD; Figure 10), the PT stimulated the peroneal nerve at locations on the proximal leg, noting the resultant ankle dorsiflexion. The WalkAide electrodes were positioned over the stimulation site that resulted in 'balanced' eversion and dorsiflexion. The subject then walked within the parallel bars while the PT manually stimulated the peroneal nerve using the WalkLink. The manual stimulation allowed the subject to walk with the WalkAide before it was

programmed, facilitating tibial angle and heel sensor data acquisition from the WalkAide prior to programming. With these data, the PT determined the tibial angle thresholds for stimulation initiation/termination; these threshold angles were adjusted as needed to optimize gait. Each subject was instructed to slowly acclimate to the WalkAide, ramping up from 2 to 8 hours of use per day over a 1 week acclimation period. The subject then returned to the Hanger office for potential adjustments; further WalkAide adjustment visits were scheduled as needed.

Figure 10: Testing of the common peroneal nerve (left) using the peroneal nerve stimulator (right), adapted from [4].

3.2.3 Functional Assessment

To quantify subject functionality and level of impairment, several common clinical assessments [Fugl-Meyer, Modified Ashworth Scale, strength and range of motion (ROM)] were performed. For Fugl-Meyer assessment, subjects were asked to complete five tasks (Appendix A) and were scored based on their level of completion of each task [49]. Subjects' Achilles and patellar tendon reflexes were also assessed. Spasticity of the affected lower leg was evaluated using the five point Modified Ashworth Scale (Table 2).

Manual Muscle Testing (MMT) was completed at the hip, knee and ankle joints to evaluate muscle strength using a five-point scale, see Table 3. Flexion and extension strength were tested at the hip and knee joints. Plantar flexion, dorsiflexion, eversion and inversion strength were tested at the ankle joint.

Table 2: Modified Ashworth Scale [adapted from 50]

Muscle Tone	Score
No increase in muscle tone	0
Slight increase in tone with a catch and release	1
Minimal resistance at the end of the ROM	1+
Slight increase in tone with catch and minimal resistance through range following catch	2
More marked increase in tone through ROM	3
Considerable increase in tone with passive movement difficult	4
Affected part rigid	5

Table 3: Criteria for MMT Grades [adapted from 50]

Muscle Contraction	Against Gravity	Range of motion completed	Resistance added	MMT Grade
No palpable muscle contraction	No	None	None	0
Muscle contraction is palpable	No	None	None	1
Contraction produces movement	No	Partial to Full	None	2
Contraction produces movement	Yes	Partial	None	3-
Contraction produces movement	Yes	Full	None	3
Contraction produces movement	Yes	Full	Minimal	3+
Contraction produces movement	Yes	Full	Minimal to Moderate	4-
Contraction produces movement	Yes	Full	Moderate	4
Contraction produces movement	Yes	Full	Almost Maximum	4+
Contraction produces movement	Yes	Full	Maximum	5

The flexion and extension ROM was assessed at the hip and knee joints. Plantar flexion, dorsiflexion, eversion and inversion ROM of the ankle and subtalar joints were also assessed using a goniometer (see Table 4).

The Modified Ashworth, strength and ROM measurements were evaluated at the end of the subject's first research visit and at the beginning of each gait analysis session.

Table 4: Goniometer Placement for ROM Measurements (adapted from [50])

Joint	Measurement	Center of goniometer landmark	Stationary arm landmark	Moveable arm landmark
Hip	Flexion/ Extension	Greater trochanter	Lateral pelvis midline	Lateral epicondyle of the femur
Knee	Flexion/ Extension	Lateral epicondyle of the femur	Greater trochanter	Lateral malleolus
Ankle	Plantar flexion/ Dorsiflexion	Approximately 1.5 cm inferior to the lateral malleolus	Fibula head	Longitudinal axis of the 5th metatarsal
Ankle	Inversion/ Eversion	Midpoint between medial and lateral malleolus	Tibial crest	2nd metatarsal

3.3 GAIT ANALYSIS PROTOCOL

Gait analysis sessions were conducted at the Motion Analysis Laboratory (MAL) in the Department of Orthopedic Surgery at the Medical College of Wisconsin. This section reviews the test protocol for these sessions including reflective marker placement for kinematic analysis, subject instrumentation for kinematic, plantar pressure, EMG, and physiologic data monitoring, and test protocol during dynamic and static trials. Details regarding the specific equipment and data acquisition are outlined in Table 5.

Table 5: Equipment overview

Data	System	Vendor	Sampling Rate (Hz)
Lower extremity kinematic data	VICON 524 Motion Analysis	VICON Motion Systems Inc.; Lake Forest, CA	120
Tibialis Anterior EMG data	MA300 EMG	Motion Lab Systems, Inc.; Baton Rouge, LA	1800
Physiologic monitoring	S610 Polar heart rate monitor	Polar Electro Inc.; Lake Success, NY	0.02
Plantar/heel pressure data	0.15 mm F-scan insole	F-Scan VersaTek System, Tekscan, Inc.; South Boston, MA	50
Stimulation, heel loading, tibia tilt data	WalkLink	Innovative Neurotronics; Austin, TX	25

3.3.1 Marker Placement for Kinematic Analysis

During each gait session, reflective markers were affixed with double sided adhesive tape on the following anatomical locations (see Figure 11) to facilitate lower extremity kinematic analysis:

- anterior superior iliac spines, ASIS (if palpable, otherwise on the lateral pelvis)
- sacrum (mid-way between the posterior superior iliac spines)
- thigh (in line with the greater trochanter and lateral epicondyle of the femur)
- lateral femoral epicondyle
- tibia
- lateral malleolus (medial malleolus if F-Scan sensors were used)
- second metatarsal head (on shoe)
- calcaneus (on shoe)

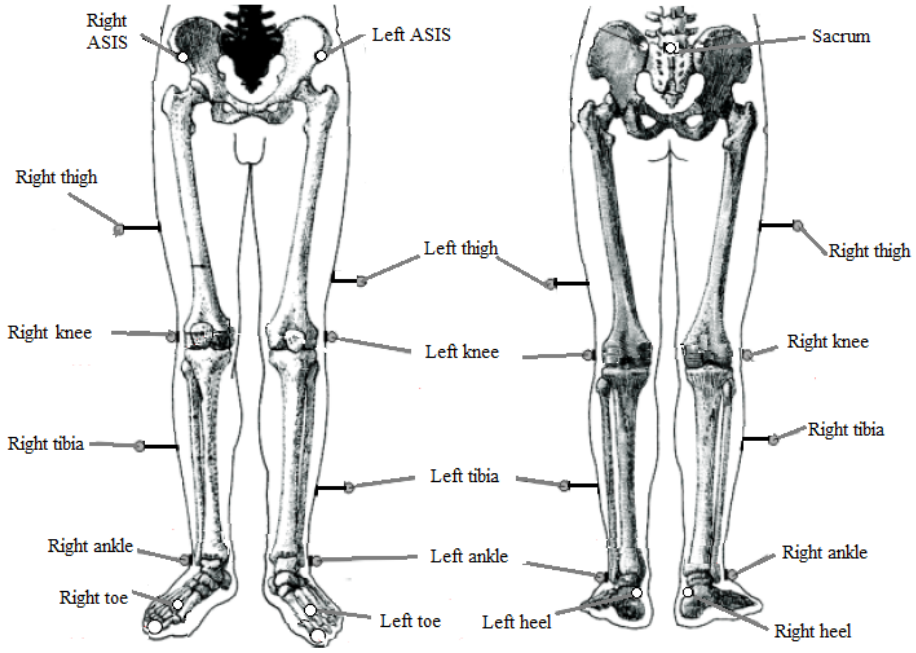

Figure 11: Marker placement for lower limb gait analysis (adapted from [51]). Knee=lateral epicondyle Ankle=lateral malleolus Heel=calcaneus

During AFO testing, the lateral malleolus marker for the affected limb was affixed to the lateral AFO ankle joint. All other markers were placed as described above.

Virtual markers were created during data processing when the anatomical location was obscured by clothes or the body, or when the marker could not be consistently viewed during ambulation. Virtual markers were created for the ASIS and sacrum when body bulk or clothing obscured the marker location; these virtual markers were referenced to markers placed on the lateral pelvis. If the subject wore a knee brace that covered the lateral femoral epicondyle, the virtual knee marker was referenced to a marker triad placed on the thigh. Finally, when F-scan insoles were used to acquire plantar pressure data, a virtual ankle (lateral malleolus) marker was referenced to the medial malleolus. For all subjects and trials, virtual markers were created for the tip of the shoe to assess toe clearance.

3.3.2 Testing Protocol

3.3.2.1 Subject Instrumentation

Strength and joint ROM tests were conducted on both lower limbs before each gait analysis session (see 3.2.3: Functional Assessment). Anthropometric measurements including height, weight, and limb segment dimensions were acquired during the first gait session. Physiologic monitoring consisted of heart rate monitoring (see Table 5); a heart rate monitor was placed beneath the shirt directly over the heart. Kinematic analysis was based on the reflective markers affixed to the respective anatomical locations.

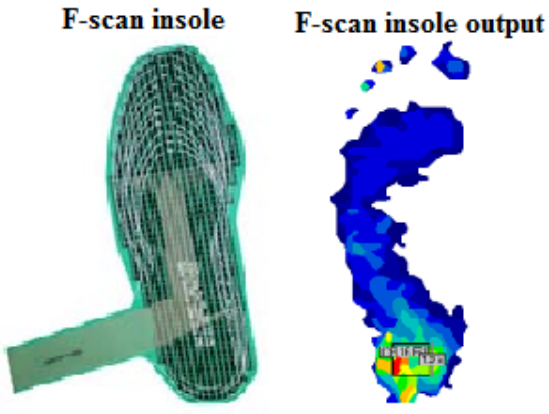

Figure 12: F-scan insole (left) and sample plantar pressure data (right).

During WalkAide gait sessions, the WalkAide heel sensor (Table 5) was placed under the insole of the subject's shoe on the affected side. An F-scan insole (Table 5) was trimmed to the subject's shoe size and placed on top of the shoe insole on the affected side, see Figure 12. After the subject donned their shoes, the WalkAide neuroprosthesis was positioned and research staff manually stimulated the peroneal nerve to identify the tibialis anterior location for electromyographic monitoring.

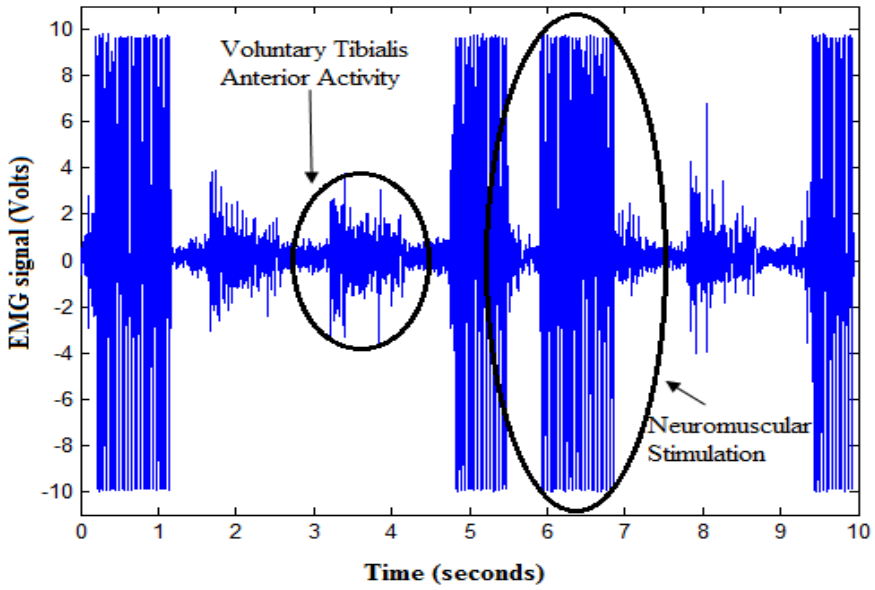

Figure 13: Tibialis anterior EMG differentiating voluntary contraction versus neuromuscular stimulation via the WalkAide.

After cleansing the skin with an alcohol swab, a bipolar electromyographic (EMG) electrode, see Table 5, was placed on the skin over the tibialis anterior and affixed with hypoallergenic adhesive tape. The electrode position was confirmed by manual stimulation from the WalkAide, ensuring that neuromuscular stimulation by the WalkAide and voluntary tibialis anterior activity could be differentiated (Figure 13). The WalkLink was placed around the subject's neck and connected to the WalkAide to acquire heel and tilt sensor data during ambulation. Finally, the F-scan insole was calibrated based on the subject's body weight using Tekscan software. Figure 14 illustrates a subject instrumented for a WalkAide gait session.

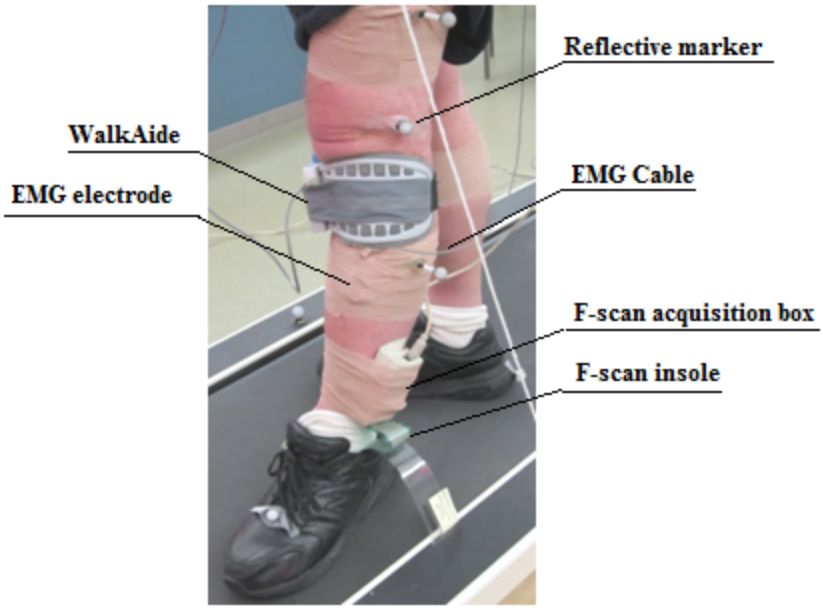

Figure 14: Subject instrumented for WalkAide gait analysis session.

3.3.2.2 Dynamic Gait Trials

Gait analysis trials were conducted on a treadmill (Landice L8, Landice Inc.; Randolph, NJ), randomly oriented in level, inclined (+7°) and declined (-7°) orientations. As the treadmill only allows level and inclined ambulation, the treadmill was positioned on a 7° declined wooden ramp, thereby supporting all treadmill orientations. For each treadmill orientation, the subject determined his/her comfortable walking speed; two minute walking trials were then conducted with 10 seconds of data acquired every 15 seconds. Kinematic and heart rate data were collected simultaneously. During WalkAide trials, tibialis anterior EMG data were acquired using a VICON analog channel, synchronizing the kinematic and EMG data. Plantar pressure data (F-scan insoles) were acquired on a separate computer. Tilt and heel sensor data, as well as stimulation data, were acquired using the WalkLink on a third computer (see Table 5); these data were transmitted to the VICON workstation via Bluetooth.

3.3.2.3 Static Pointer Trials

After completing the walking trials, static trials for virtual marker determination were conducted with the treadmill in the level orientation. The tip of the virtual marker wand was positioned over the respective anatomical location (e.g. ASIS, lateral malleolus, toe). As the subject quietly stood, the wand was "poked", pointing to the location of interest, thereby decreasing the distance between the wand's two reflective markers (Figure 15) and creating a reference position for the virtual maker in relation to other lower extremity markers.

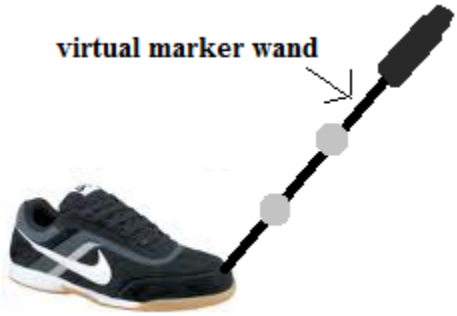

Figure 15: Virtual marker wand positioned to create virtual toe marker.

3.3.2.4 Static Knee Marker Trial

A final static trial was conducted in which knee alignment devices (KADs) were placed over the medial and lateral epicondyles of both limbs to better define the knee joint axis (Figure 16). These KADs replaced the lateral femoral epicondyle markers.

Figure 16: KAD (left) positioned over the femoral epicondyles during a static trial (right).

3.4 KINEMATIC DATA PROCESSING

Upon completion of each gait analysis session, the subject's data were first processed to obtain c3d kinematic data files. Kinematic data processing was completed at the Motion Analysis Laboratory at the Medical College of Wisconsin using VICON Workstation software (version 5.4) and MATLAB (R2010a, Mathworks; Natick, MA). This section summarizes raw marker data conversion to three-dimensional (3D) motion data, creation of virtual markers, and use of a PlugInGait model to integrate marker kinematic data, define lower extremity limb segments and calculate joint angles.

3.4.1 Construction of 3D Motion Data

Raw marker data from the 15 cameras were reconstructed using direct linear transformation to produce 3D marker position data. Each respective marker was manually labeled. Gaps (due to marker obstruction in the camera field of view) in marker data up to 10 frames were filled by linear interpolation. The marker motion data were filtered using a low pass quintic spline Woltring filter (mean square error of 20 corresponding to a noise tolerance of 20 mm^2) to smooth data. These video data processing steps are summarized in Figure 17.

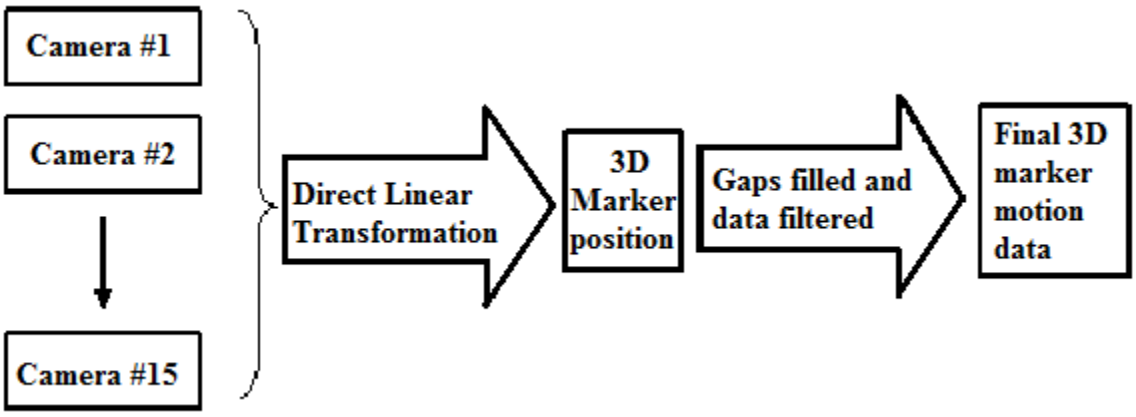

Figure 17: Flow chart illustrating conversion of raw video data to 3D marker data.

3.4.2 Creation of Virtual Markers

Virtual markers for the ASIS, sacrum, knee, lateral malleoli, and toes were created as needed using MATLAB code (J. Long, Medical College of Wisconsin), the virtual marker wand data, and reference marker data; this procedure is summarized in Figure 18. The static virtual marker trial data files were opened and the individual camera views from the 15 cameras were reconstructed to produce the 3D motion data. The two markers on the wand and the three reference markers (see Table 6) were labeled. The position of the virtual maker during the static trial was determined and a vector relating the virtual marker to the reference markers was created. This vector was then used to create the virtual marker in all dynamic trial data files.

Table 6: Virtual marker reference

Virtual marker	**Reference markers**
Right/left ASIS	Right pelvis
	Left pelvis
	Sacrum
Sacrum	Right pelvis
	Left pelvis
	Replacement sacral marker superior to anatomical location
Knee	Thigh plate marker 1
	Thigh plate marker 2
	Thigh plate marker 3
Right/left lateral malleolus	Right/left Toe
	Right/left medial malleolus
	Right/left heel
Right/left toe marker for toe clearance	Right/left toe
	Right/left medial or lateral malleolus
	Right/left heel

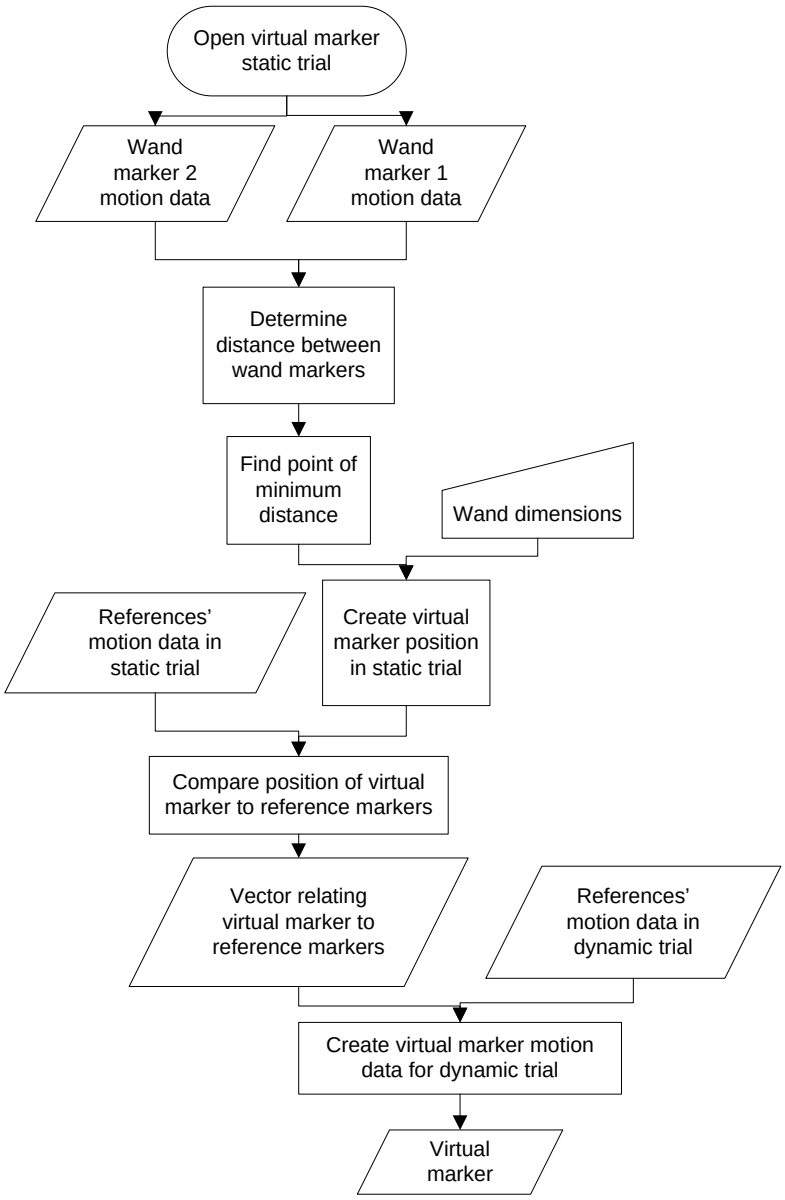

Figure 18: Procedure for creation of virtual markers.

3.4.3 Calculation of Lower Extremity Joint Angles

The PlugInGait model used to label the markers also defines the respective limb segments (pelvis, right/left femur, right/left tibia, right/left foot) in terms of these markers. The segments (rigid bodies) were defined by three non-collinear markers as follows:

- pelvis: right and left ASIS markers and the sacral marker
- femur: ASIS and knee markers, thigh wand
- tibia: knee and ankle markers and the tibia wand
- foot: heel, toe and ankle markers.

The origin of each segment was located at the distal end of the respective segment. The first axis of each segment was defined by the line joining the distal and proximal joint centers of the segment, the second axis was defined by the main axis of rotation of the distal joint, and the third axis was perpendicular to the first two axes (Figure 19).

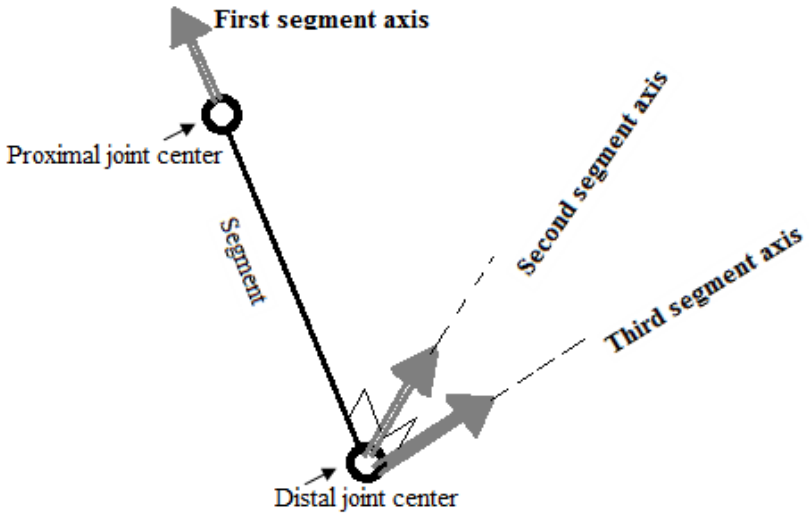

Figure 19: Segment axes definition.

Joint centers were calculated based on marker locations and anthropometric measurements. The hip joint centers were based on an offset applied to the ASIS markers; this offset calculation incorporated leg length (L), inter-ASIS distance (IAD), and pelvic geometry [51]. The knee joint centers were based on the hip joint center, the knee and thigh wand markers, and the knee offset (KAD static trials). The ankle joint center was determined using the knee joint center, the tibia marker, the ankle marker, and the ankle offset (half ankle width) [51]. Euler angles about these respective joint centers were calculated using VICON Workstation.

All marker motion data, joint angles, and analog data were stored in binary c3d files.

3.5 GAIT PARAMETERS DATA PROCESSING

The aforementioned c3d files were further manipulated using MATLAB code to identify gait events [toe off (TO) and heel strike (HS)] and quantify temporal parameters, gait asymmetry and toe clearance measures.

3.5.1 Gait Events and Temporal Parameters

The velocity of the toe and heel markers, calculated using the finite difference method, was used to identify HS and TO events – both of which occur at local minima. HS occurred at the minimum velocity of the heel marker, immediately after the period of maximum heel marker velocity (i.e. swing phase) – see Figure 20. TO occurred at the minimum velocity of the toe marker, just prior maximum toe marker velocity or swing phase (Figure 21). The frame numbers corresponding to these HS and TO events were written to Excel files for use in subsequent algorithms.

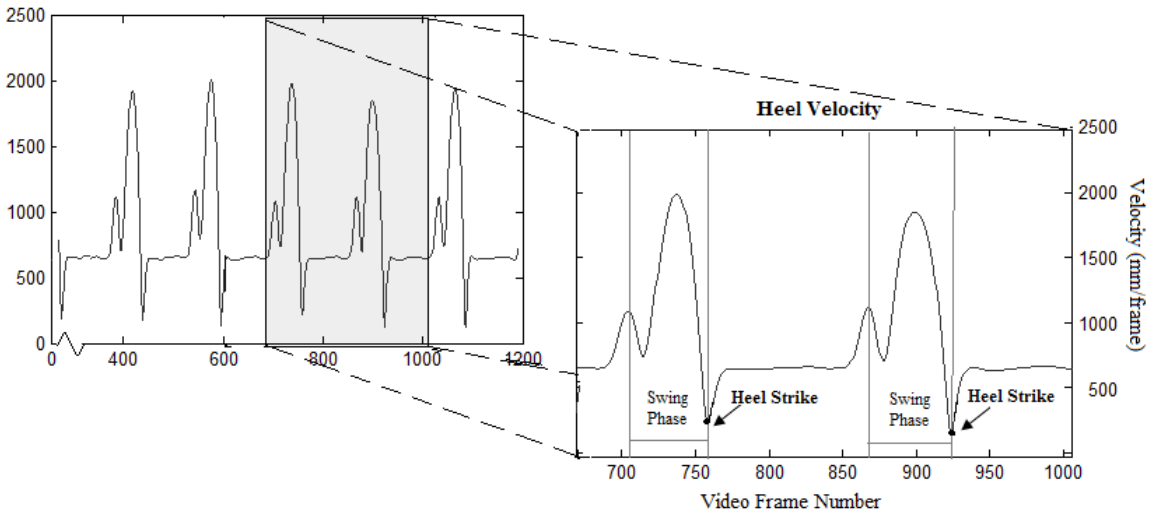

Figure 20: Heel marker velocity illustrating HS and swing.

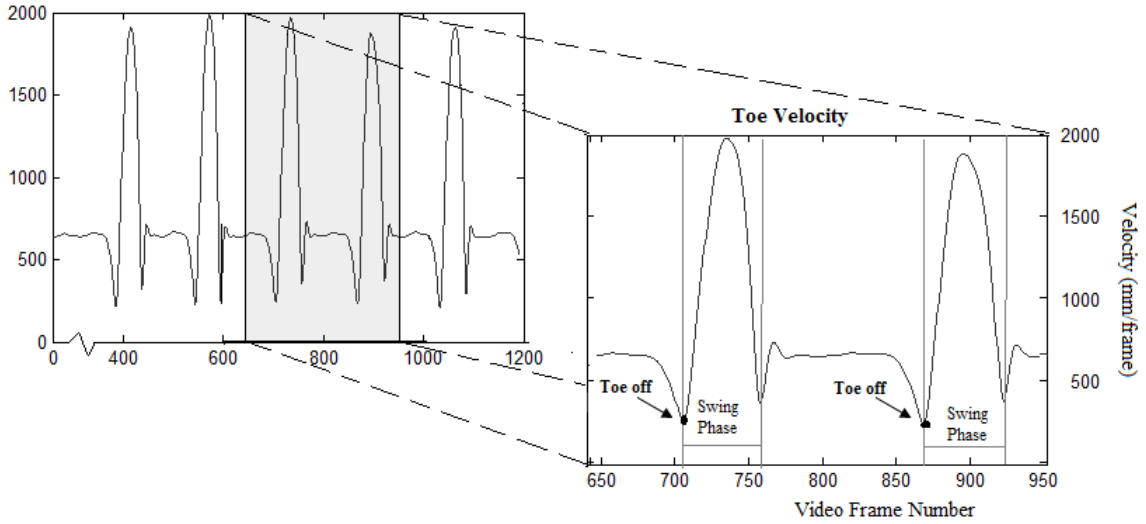

Figure 21: Toe marker velocity illustrating TO and swing.

Temporal parameters such as stance and swing duration, single limb support, and double limb support periods for both the affected and unaffected limbs were calculated based on these HS and TO events, see Figure 22 and Section 2.2.2. Cadence and gait cycle duration were similarly evaluated based on the HS events of both the affected and unaffected limbs. Cadence was defined as the number of gait cycles or steps (one less than the number of ipsilateral heel strikes) divided by the time elapsed between the first and last HS (in minutes).

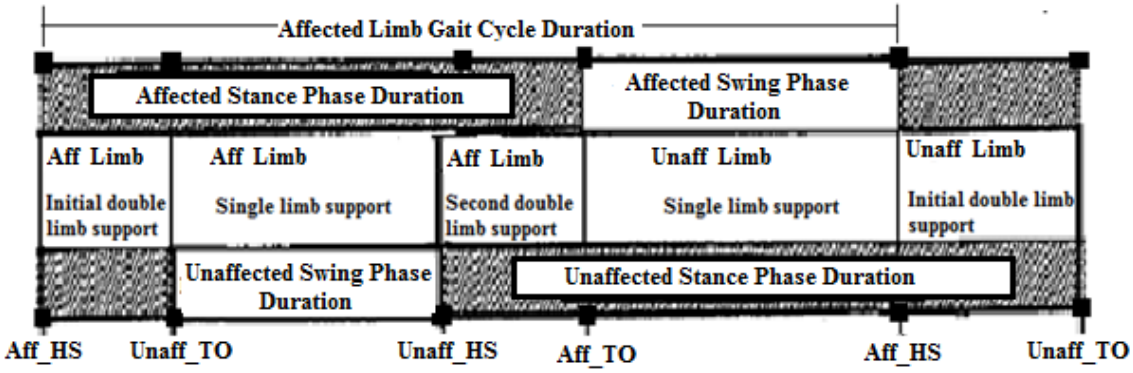

Figure 22: Temporal parameters based on gait events (adapted from [14]).
Aff=affected limb and Unaff= unaffected limb

3.5.2 Functional Phased of Stance and Swing

Each stance phase (ipsilateral HS to ipsilateral TO) was separated into four functional phases [loading response (LR), mid-stance (MSt), terminal stance (TSt) and pre-swing (PS),] see Section 2.2.1 and Table 7; this data processing procedure is summarized in Figure 23 (left).

Table 7: Functional phases of stance

Gait Phase	Beginning	End
LR	Ipsilateral HS	Contralateral TO
MSt	Contralateral TO	First 1/2 of the frames to contralateral HS
TSt	First 1/2 of the frames to contralateral HS	Contralateral HS
PS	Contralateral HS	Ipsilateral TO

Similarly, each swing phase (ipsilateral TO to ipsilateral HS) was separated into three functional phases [initial (IS), mid (MS), and terminal swing (TS)], see Section 2.2.1 and Table 8. While the original swing phase durations were based on functional definitions of the phases described by Perry [12], the variability in these phases (per cent gait cycle) was substantial. As such, the functional swing phases simply divided the swing phase into thirds, see flow chart in Figure 23 (right) and Table 8 (second row).

Table 8: Functional phases of swing

	Gait Phase	Beginning	End
Based on functional definition [12]	IS	Ipsilateral TO	Ipsilateral toe crosses contralateral heel
	MS	Ipsilateral toe crosses contralateral heel	Vertical tibia
	TS	Vertical tibia	Ipsilateral HS
Alternative definition: thirds	IS	Ipsilateral TO	First 1/3 of the frames to ipsilateral HS
	MS	First 1/3 of the frames to ipsilateral HS	Second 1/3 of the frames to ipsilateral HS
	TS	Second 1/3 of the frames to ipsilateral HS	Ipsilateral HS

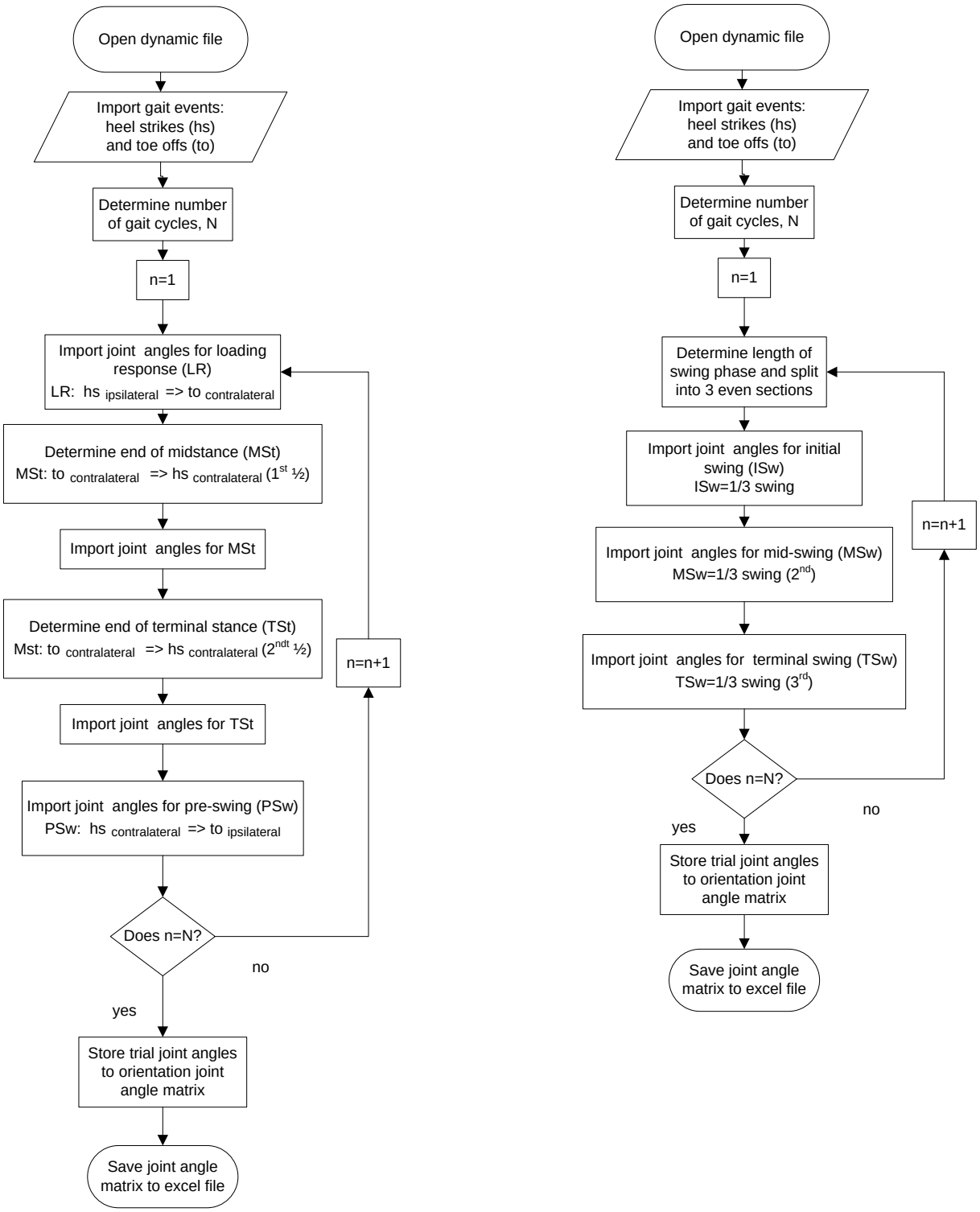

Figure 23: Flow chart summarizing the division of gait cycle data into stance (left) and swing (right) functional phases.

The hip, knee and ankle joint motion in the sagittal plane was divided into these functional phases of stance and swing for each gait cycle. The mean, minimum, and maximum joint motion for each functional phase were quantified (Figure 24) and stored in a spreadsheet for each subject, joint and treadmill orientation for plotting and subsequent analysis.

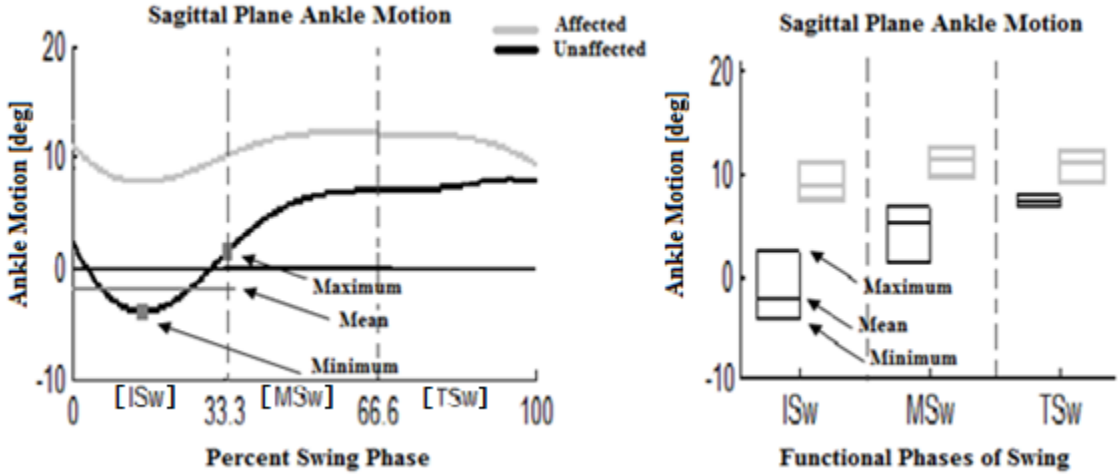

Figure 24: Sample ankle motion time series (left) and associated mean, minimum and maximum sagittal plane ankle angle for the affected and unaffected limbs for the functional phases of swing (right).

3.5.3 Gait Asymmetry

As discussed in Chapter 2, effective treatment of foot drop for post-stroke subjects is expected to result in improved gait symmetry. However, current measures of gait symmetry (or asymmetry) are limited since they incorporate the affected limb in the normalization, are unable to account for sign changes which occur in kinematic data and ROM differences between subjects or joints, and lack the resolution needed to observe small inter-limb differences. As such, a new parameter to quantify gait asymmetry for kinematic data was proposed.

This new parameter, gait asymmetry (GA), was defined as the difference in the respective joint angle parameter between the affected and unaffected limbs normalized by the ROM of the unaffected limb joint:

$$GA = \left(\frac{K_{affected} - K_{unaffected}}{ROM_{unaffected}} X\ 100\right) \quad \text{(Eq 3)}$$

where $K_{affected}$ is the kinematic parameter of interest for the affected limb (e.g. minimum, maximum or mean sagittal plane joint motion during the various stance and swing functional phases), $K_{unaffected}$ is the kinematic parameter of interest for the unaffected limb, and $ROM_{unaffected}$ is the mean range of motion of the specific joint (e.g. hip, knee or ankle) on the unaffected side for the entire gait cycle.

Unlike the various measures of gait asymmetry used previously in the literature to assess temporal gait asymmetry or asymmetry between the limbs in free versus constrained conditions (see Section 2.6.2.2), the above GA measure accounts for sign changes in the kinematic data as well as differences in ROM between subjects or joints. These advantages are illustrated in Figure 25 which contrasts this asymmetry measure for minimum ankle joint motion during IS with alternative measures used previously in the literature. The difference in ankle joint minima between the affected and unaffected limbs in both the top and bottom curves is 12°, although the top curves reflect a sign change (flexion=dorsiflexion=positive; extension=plantar flexion=negative). The denominator for GA, 25° ROM for the unaffected limb, is based on that ROM of the entire gait cycle although only ankle motion during swing is shown in this figure. Since the relative difference between limbs is 12° for both graphs, the symmetry (or asymmetry) measure should be insensitive to the change in sign. Previously reported measures of symmetry, symmetry ratio (SR) and symmetry index (SI), demonstrate excessive sensitivity to these sign changes; the new GA measure is insensitive to this sign change.

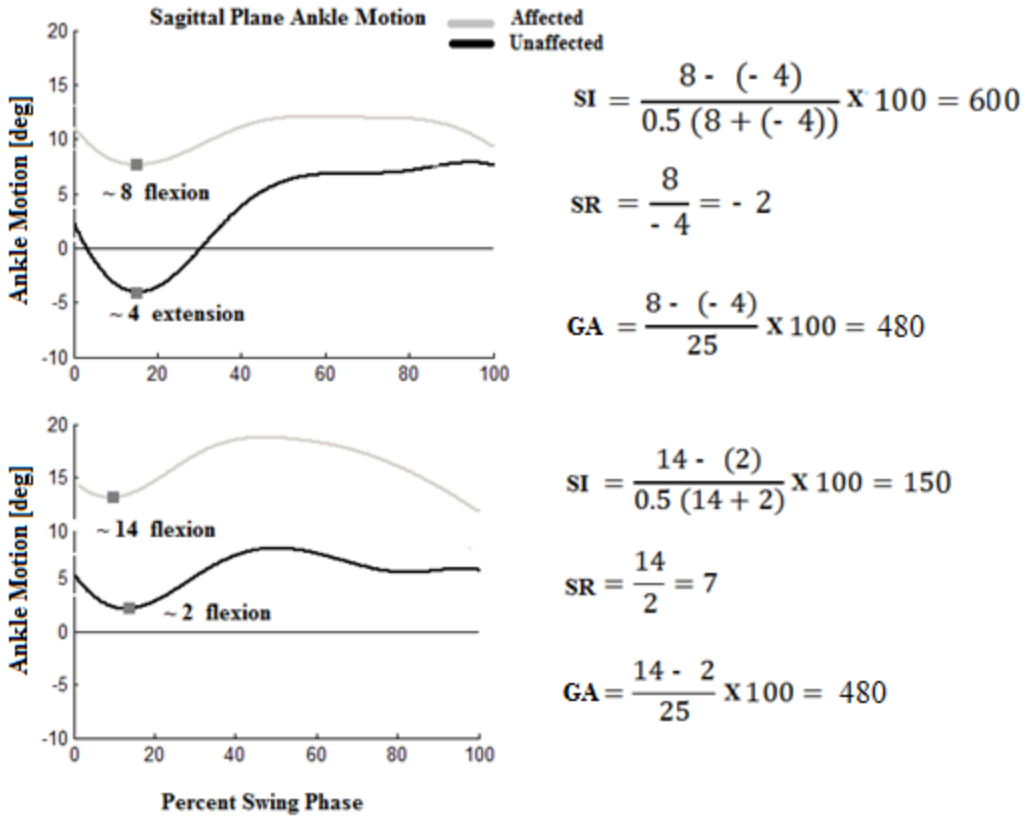

Figure 25: Comparison of GA with prior symmetry measures reported in the literature for ankle kinematic data with (top) and without (bottom) a sign change (flexion = positive, extension = negative). Each parameter was evaluated for minimum ankle joint motion during IS.

Figure 26 contrasts these measures for ankle versus knee kinematic data during mid-stance, demonstrating that GA also facilitates comparison of gait asymmetry between joints. The difference in mean joint motion between the affected and unaffected limbs during mid-stance was 12° for both joints; however, the ROM of the unaffected knee differs from that of the ankle (55° versus 25°, respectively). Only the GA measure accounts for the difference in joint ROM or scale.

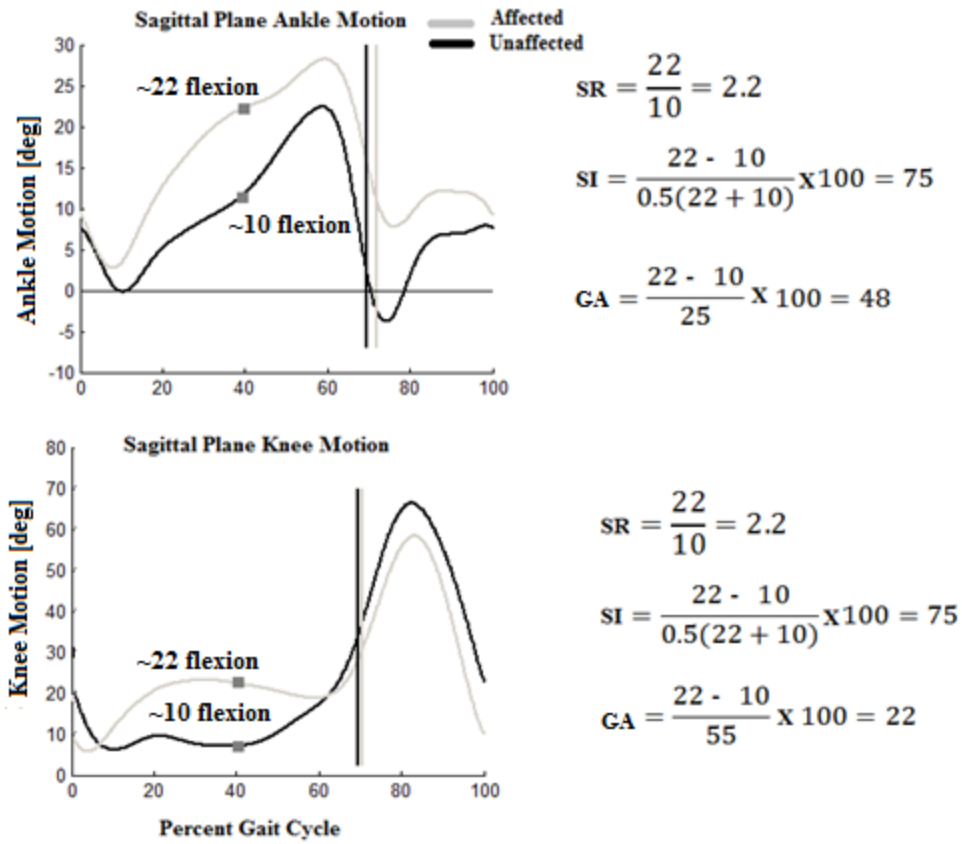

Figure 26: Comparison of GA with prior symmetry measures reported in the literature for ankle versus knee motion during mid-stance.

3.5.4 Toe Clearance

To investigate the efficacy of the two treatments (AFO, WalkAide) and assess whether the treatment minimized the risk of falling, toe clearance was assessed. Increased toe clearance may reflect a decrease in the risk of stumbling or falling, although excessive toe clearance of the affected versus unaffected or "normal limb" may indicate unnecessary energy expenditure. Therefore, both the magnitude of toe clearance and the difference in toe clearance between the affected and unaffected sides were evaluated.

To account for inclined and declined treadmill positions, three markers were placed on the treadmill to define the treadmill orientation (Figure 27). These three markers were used to determine the plane of the treadmill relative to the laboratory coordinate system. The distance between the virtual toe marker and the plane of the treadmill was calculated throughout swing.

Toe clearance was defined as the minimum distance between the virtual toe marker and the plane defined by the treadmill. These procedures are summarized in Figure 28.

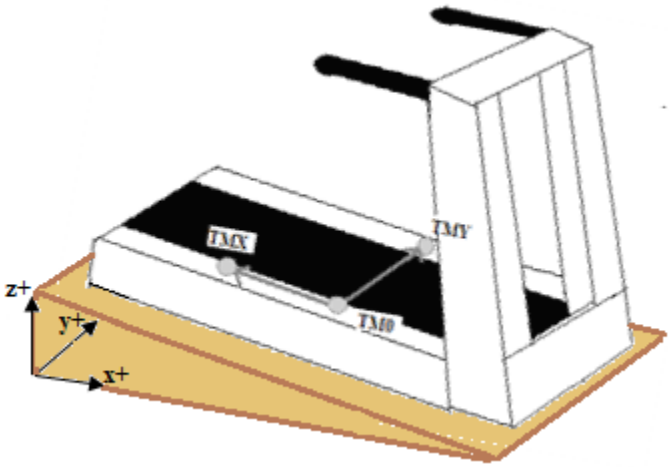

Figure 27: Markers (TM0, TMX, TMY) used to define treadmill plane relative to the laboratory-based coordinate system (XYZ).

3.6 WALKAIDE DATA PROCESSING

As discussed in Chapter 2, the clinician programs the WalkAide using the tilt sensor and tibial angles during level walking. However, McIntosh et al. [41] observed that lower extremity joint kinematics change on inclined and declined surfaces. Therefore, if WalkAide programming is based solely on level walking, stimulations may be missed or occur at different times throughout the gait cycle during inclined and declined ambulation.

In addition to the tilt sensor, the WalkAide system also includes a heel sensor. The clinician can program the WalkAide using either the tilt or the heel sensor. The heel sensor data may not change as significantly during non-level walking and therefore heel sensor-based stimulation may be more reliable for ambulation on non-level surfaces. Theoretical stimulations based on the WalkAide heel sensor data were determined to investigate whether the heel sensor may be a more reliable control parameter for stimulation during non-level walking.

To investigate potential missed and extraneous stimulations for both tilt and heel sensor-based stimulation, WalkAide stimulation reliability was determined. WalkAide tilt and heel sensor-based stimulation initiation and termination timing were also investigated to determine

whether stimulation timing changes with inclined and/or declined walking with either WalkAide sensor.

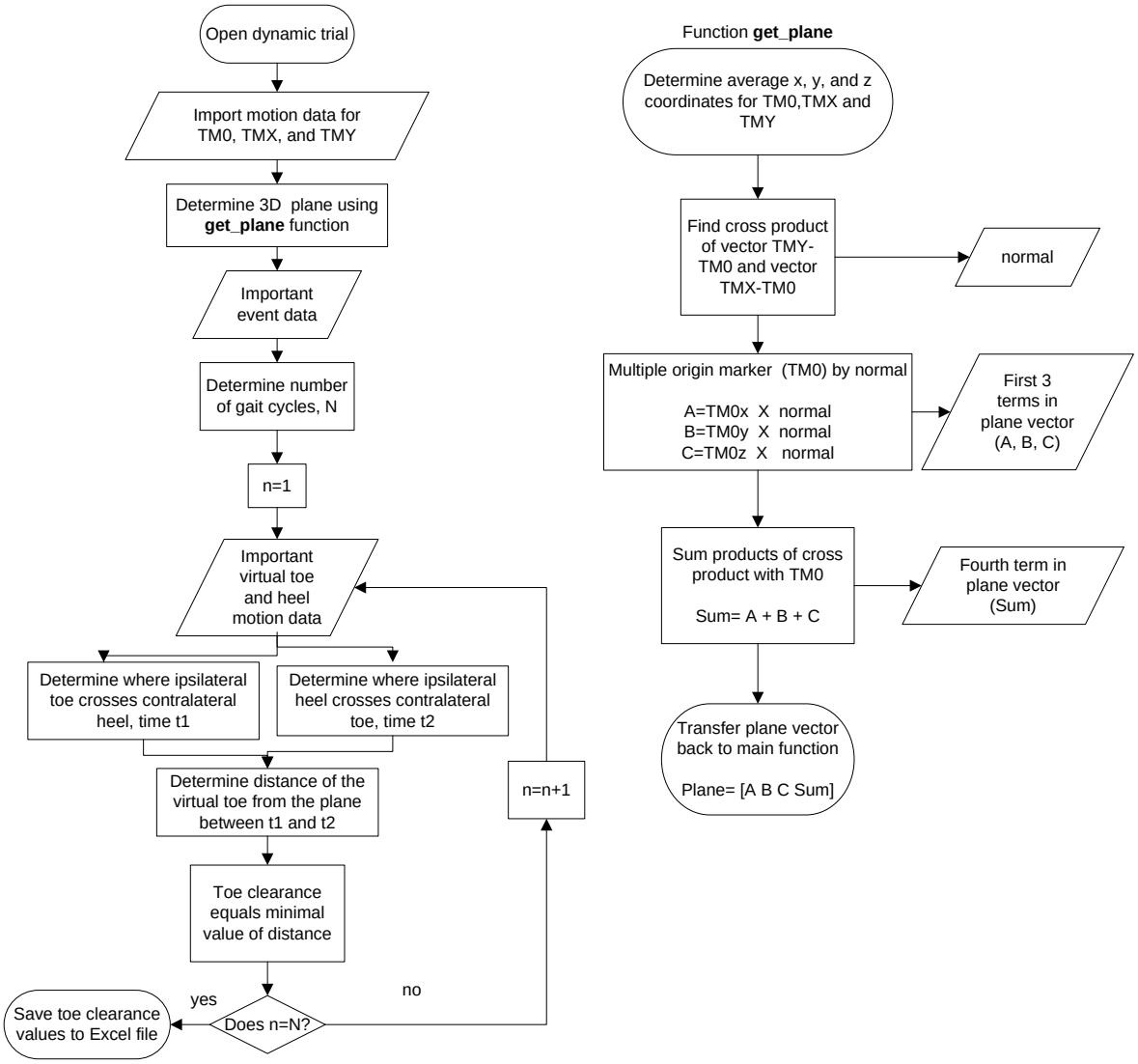

Figure 28: Flow chart summarizing the procedures for determining toe clearance (left) and treadmill plane (right).

3.6.1 Determination of Tilt Sensor-based Stimulation using EMG

Anterior tibialis EMG data (sampled at 1800 Hz) were used to identify the specific initiation and termination frame of each stimulation (Figure 29). The EMG data were high pass filtered (250 Hz cut-off frequency, 10^{th} order Butterworth) to eliminate 60 Hz noise and voluntary muscle activity and then rectified and low-pass filtered (8 Hz cut-off frequency, 8^{th} order Butterworth) to produce a linear envelope – see Figure 30.

Threshold detection was then applied to the linear enveloped EMG to identify the on/off stimulation or stimulation initiation/termination periods. Based on preliminary analysis of all EMG and WalkAide data files for the first four subjects, a threshold value of 37% mean peak EMG consistently differentiated these stimulation periods. All frames in which the EMG magnitude exceeded the threshold were stored in a "stimulation on" array. The frame corresponding to the beginning of each "stimulation on" period was stored as the stimulation initiation; the final frame of each "stimulation on" period was stored as the corresponding stimulation termination. The stimulation initiation and termination frames determined from the EMG data were later used to determine if the reliability and timing of the tilt sensor-based programming changed with treadmill orientation.

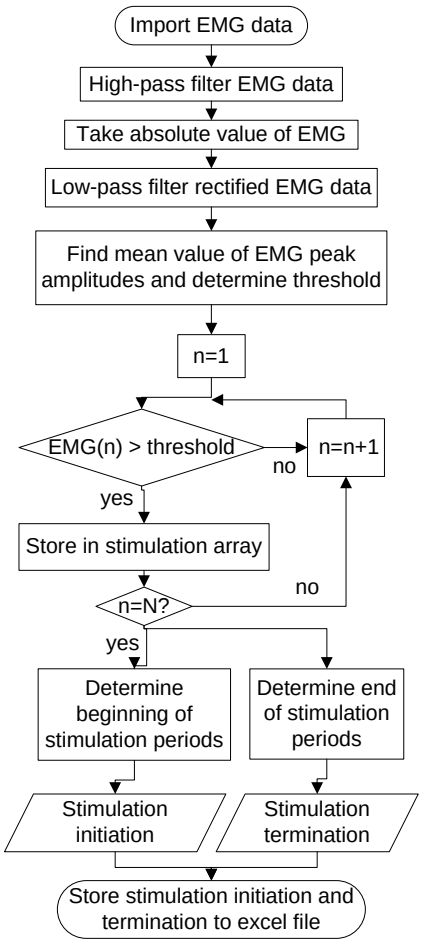

Figure 29: Flow chart summarizing the procedure used to determine neuroprosthesis stimulation initiation and termination based on tibialis anterior EMG data.

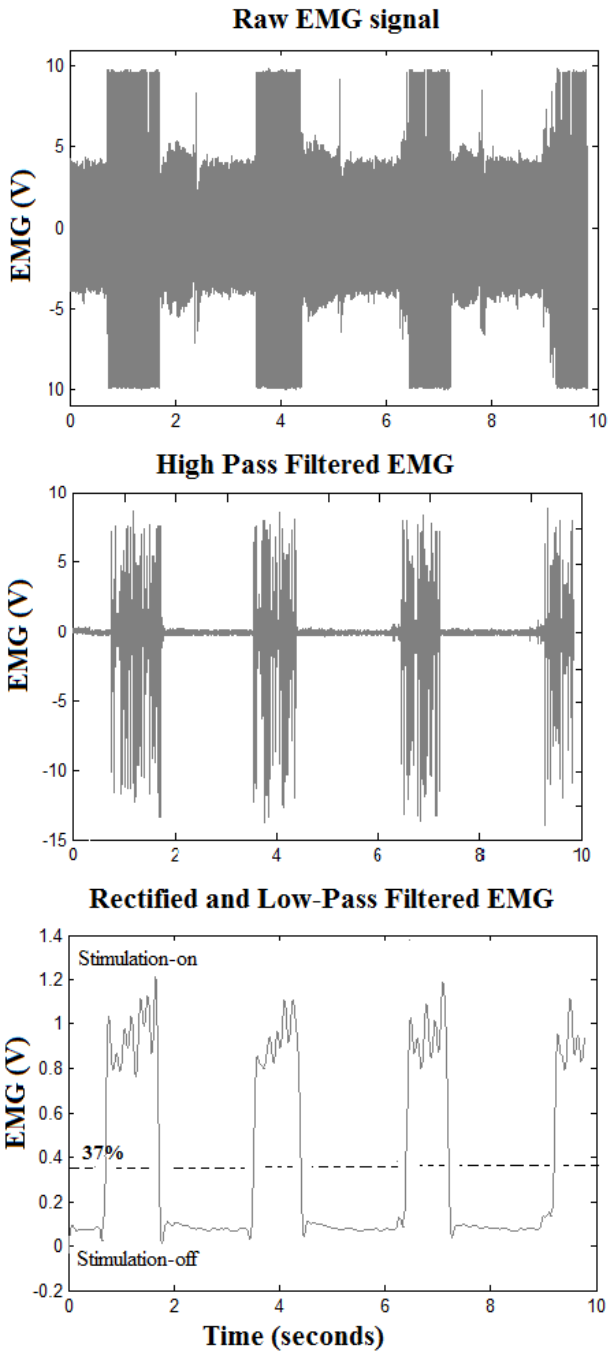

Figure 30: EMG data processing contrasting raw tibialis anterior EMG (top), high pass filtering to remove 60 Hz noise and voluntary contraction (middle), and resultant linear envelope (bottom) illustrating the WalkAide stimulation periods.

3.6.2 Clinical WalkAide Parameters and Programming

The WalkAide is typically programmed clinically using the tilt sensor since this sensor can be used without shoes. The tilt sensor measures the approximate angle of the tibia. The

clinician controls the stimulation by setting and refining several parameters: stimulation initiation/termination thresholds, wait time, and minimum/maximum stimulation duration. These control parameters are defined below.

1) Stimulation initiation and stimulation termination thresholds: Tibial tilt increases as the knee flexes and decreases as the knee extends, see Figure 31. The stimulation initiation threshold is defined as the tilt threshold that will initiate a new stimulation. The stimulation termination threshold is defined as the tilt threshold that will end or terminate the stimulation

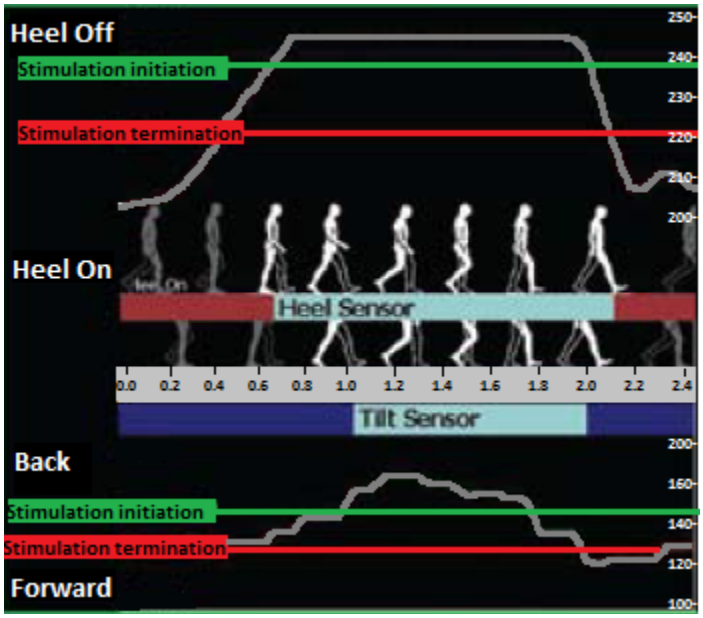

Figure 31: Stimulation initiation and termination thresholds based on the WalkAide tilt (bottom gray waveform) and heel sensor (top gray waveform) data [4].

2) Wait time: The wait time is defined as the minimum time interval between stimulation termination and a new stimulation.

3) Minimum stimulation duration: The minimum stimulation duration is defined as the minimum stimulation time.

4) Maximum stimulation duration: The maximum stimulation duration is the maximum duration of a sustained stimulation. If stimulation termination threshold has not been achieved by this time, the stimulation will be terminated based on this setting.

After all the parameters have been set, WalkAide stimulation (tilt sensor control) occurs as follows: stimulation begins when the tilt angle exceeds the stimulation initiation threshold. Stimulation continues until the minimum stimulation duration is reached. Stimulation continues further until the tilt angle falls below the stimulation termination threshold or the maximum stimulation duration has been reached. A new stimulation may not be initiated until the minimum wait time has passed and the stimulation initiation threshold has again been exceeded.

During clinical programming, the WalkAide is initially programmed with default parameter values (e.g., stimulation initiation/termination for heel and tilt sensors: 255/1; minimum/maximum stimulation duration: 0.5/1.0 seconds; wait time: 0.4 seconds). The clinician monitors the subject's ambulation, focusing on the subject's walking speed, ankle dorsiflexion and knee extension during loading response, knee flexion during swing and potential hip hiking or circumduction during swing. Based on these observations, the clinician adjusts the default parameters as needed for the specific subject. For example, the minimum stimulation duration is increased if the subject is a slow walker; the stimulation termination angle is reduced if the subject has prolonged loading response, requiring ankle dorsiflexion for a longer period of time following swing termination).

3.6.3 Theoretical Heel Sensor-Based Stimulation

The WalkAide is typically programmed and controlled using the tilt sensor. However, the clinician also has the option to program and control WalkAide stimulation using heel sensor data. This method requires that the clinician set the heel loading thresholds for stimulation based on heel loading (stance) and heel unloading (terminal stance and swing), see Figure 32 (top). Note that the heel sensor data are in arbitrary units; high values represent no load and low values correspond to sensor loading.

In addition to tilt sensor and stimulation data, WalkAide heel sensor data were acquired during gait analysis sessions. Theoretical heel sensor-based stimulation was estimated using the

parameters designated during clinical tilt sensor programming (e.g. wait times, minimum and maximum stimulation durations) and setting the respective stimulation initiation and termination based on sample (3-4 ten second level walking trials) heel sensor data.

The stimulation thresholds were now based, not on tilt data, but on heel sensor data. Due to the inverse nature of the heel sensor output, unloading corresponded to periods in which the heel data were high; loading corresponded to period in which the heel data were low - see Figures 32 and 33. A single threshold was used to differentiate heel loading and unloading periods for a given subject. This threshold was set in terms of the heel sensor loading range such that extraneous stimulations due to foot movement were minimized; threshold values ranged from 1-15% for the respective subjects, see Figure 33 (bottom).

This threshold define the preliminary stimulation initiation (heel unloaded) and termination (heel loaded) times. The first frame immediately prior to the heel sensor exceeding the threshold, signifying an unloaded heel sensor, was the stimulation initiation time. The first frame after the heel sensor value fell below the threshold, signifying a loaded heel sensor, was the stimulation termination time.

The preliminary stimulation initiation and termination times were adjusted based on the wait time and minimum/maximum stimulation duration. Each 10 second data trial was reviewed to ensure that relative timing of subsequent initial theoretical heel sensor-based stimulations did not occur during the wait time previously defined during clinical tilt sensor based programming. Any theoretical stimulations occurring during this wait time were delayed such that the wait time was satisfied, defining a new stimulation initiation frame.

Each 10 second data trial was reviewed a second time to ensure that each theoretical heel sensor-based stimulation duration met or exceeded the minimum stimulation duration previously defined during clinical tilt sensor based programming. If the stimulation duration was less than the minimum duration, the stimulation termination frame was delayed such that the minimum stimulation duration was satisfied.

Finally, each 10 second data trial was reviewed a third time to ensure that each theoretical heel sensor-based stimulation duration did not exceed the maximum stimulation duration previously defined during clinical tilt sensor based programming. If the stimulation duration was greater than the maximum, the stimulation termination frame was adjusted such that the stimulation duration satisfied the maximum stimulation duration setting.

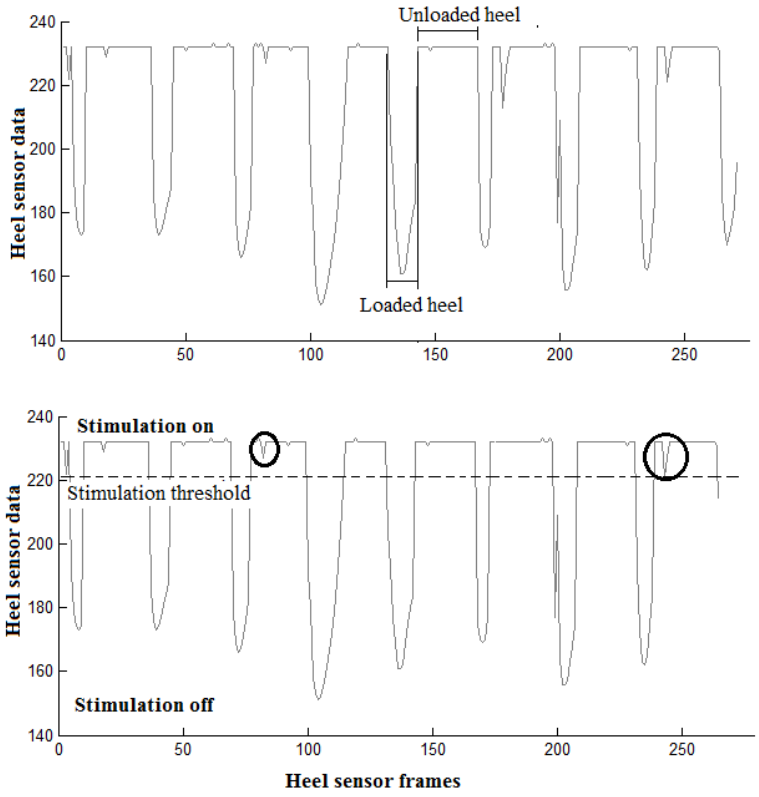

Figure 32: WalkAide heel sensor data illustrating loading and unloading periods (top) and corresponding stimulation periods (bottom) using threshold detection techniques to minimize extraneous stimulation due to small variations in heel sensor data (bottom).
Note, the WalkAide heel sensor data are reported in arbitrary units.

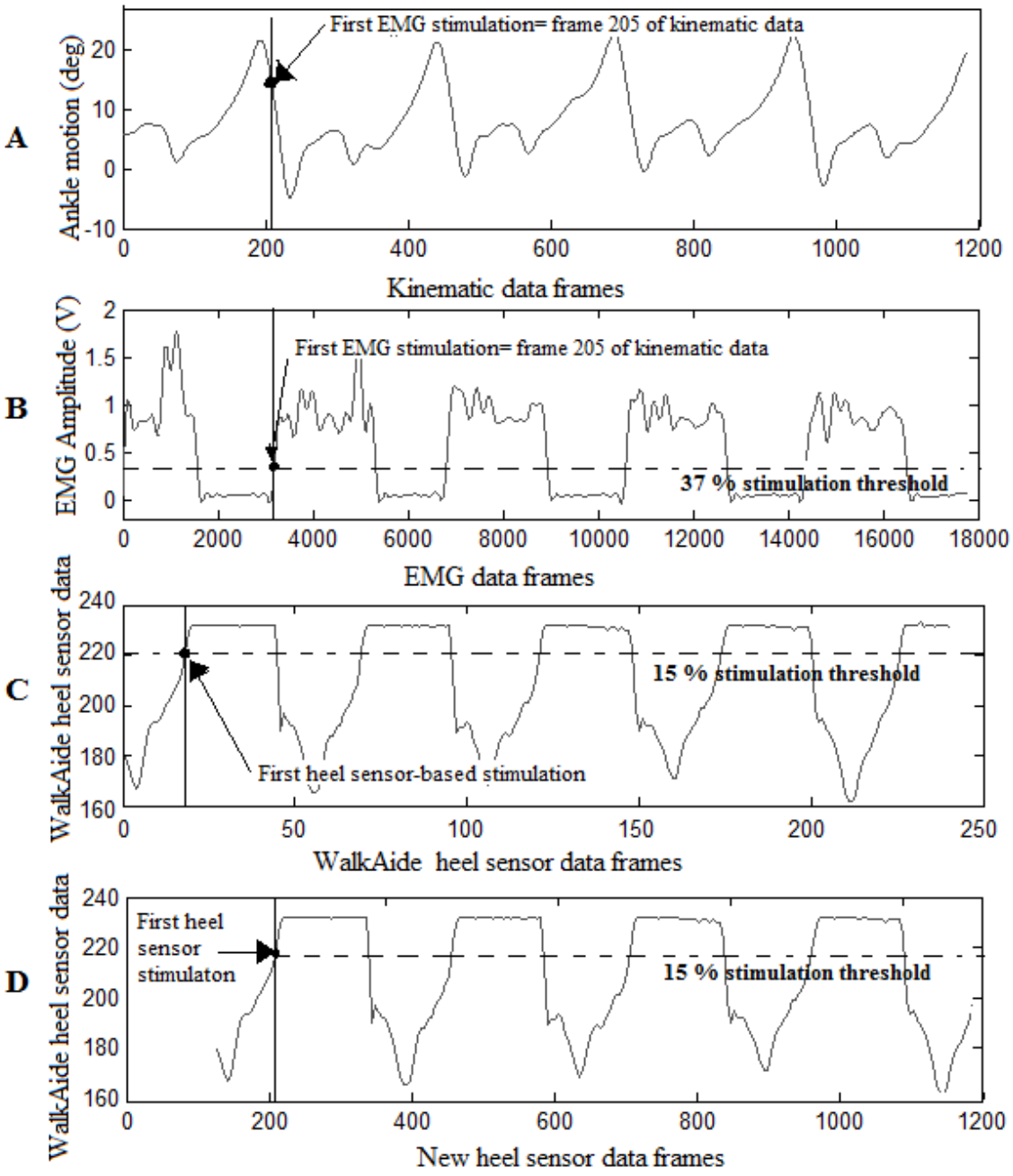

Figure 33: Ankle angle as a function of "kinematic" frame number (A, 120 Hz), EMG data as function of "EMG" frame number (B, 1800 Hz), WalkAide heel sensor data as a function of "WalkAide" frame number (C, 25 Hz), and WalkAide heel sensor data as function of "kinematic" frame number (D, 120 Hz).

3.6.4 Stimulation Reliability

One of the hypotheses motivating this research was that as WalkAide stimulation initiation and termination is based on tibial angle data during level walking, missed or extraneous

stimulations may occur during inclined and declined ambulation. Missed and extraneous stimulations may decrease with heel sensor-based stimulation if heel sensor data (and loading/unloading thresholds) are less sensitive to treadmill. The stimulation reliability (StR) for both tilt and heel sensor-based stimulation was characterized.

Prior to determining the StR (and timing), the timing of WalkAide heel sensor data was adjusted to facilitate synchronization with the kinematic data. The WalkAide heel sensor data were sampled at 25 Hz, much slower than that for kinematic (120 Hz) and EMG (1800 Hz) data used to determine the gait cycle and stimulation initiation/termination, respectively. To convert and synchronize the theoretical heel sensor-based stimulation initiation and termination frames (25 Hz) to the respective gait cycle frames (120 Hz), the following equation was used:

$$Stim_{k\_frame} = \left[\left(WA_{frame} - WA_{stim1}\right)X\frac{120}{25}\right] + K_{stim1} \qquad \text{(Eq 8)}$$

where $Stim_{k\_frame}$ is the heel sensor-based stimulation "kinematic" frame, WA_{frame} is the WalkAide heel sensor stimulation frame, WA_{stim1} is the WalkAide frame of the first stimulation onset, and K_{stim1} is the "kinematic" frame of the first stimulation onset.

StR of the WalkAide tilt and heel sensors was determined using the gait cycle period (defined in terms of heel strike events) and stimulation initiation frame data of each respective sensor. Specifically, StR was defined as:

$$StR = \frac{Stim}{GC} \qquad \text{(Eq 4)}$$

where GC is the number of gait cycles per trial and Stim refers to the number of stimulation periods per trial. The number of gait cycles per trial was one less than the number of affected limb HS's. The number of stimulation periods per trial was the number of stimulation initiations that occurred between the first and last HS events of that respective trial. A StR value of one, therefore, indicated that the WalkAide stimulated during each gait cycle; StR values less than one indicated missed stimulations and values greater than one reflected extraneous stimulations. StR

values for both tilt and heel sensor stimulations were evaluated to investigate the effect of treadmill orientation on StR, and whether StR was improved using heel sensor-based stimulation.

3.6.5 Stimulation Timing

In addition to potential effect of orientation on the StR of tibial angle-based programming, the timing of the stimulation initiation and termination may also be affected by treadmill orientation. As discussed in Chapter 2, ankle and knee kinematic data during non-level walking differ from that during level walking. As such, the tibial angle is also expected to change during non-level walking. Since the WalkAide stimulation programming is based on tibial angle thresholds during level walking only, inaccurate or inconsistent stimulation timing may occur during non-level walking. Heel sensor-based stimulation may decrease the potential stimulation inaccuracy and inconsistency on non-level surfaces. To investigate the accuracy and consistency of the WalkAide tilt and heel sensor-based stimulation timing, the stimulation initiation and termination times for each sensor were investigated.

The gait events (HS and TO) and stimulation initiation and termination frames were used to determine the timing of the respective WalkAide stimulation initiation and termination in terms of percent gait cycle. For each stimulation, the stimulation initiation frame was compared to the HS event frames. If the stimulation initiation frame fell between the two consecutive HS's, the StI timing can be expressed as a function of gait cycle:

$$StI = \frac{StI_{frame} - HS(n)}{HS(n+1) - HS(n)} \qquad \text{(Eq 5)}$$

where StI_{frame} is the stimulation initiation frame, HS(n) is the frame corresponding to the first HS, HS(n+1) is the frame of the subsequent HS, and n is the respective gait cycle. In Figure 33 for example, StI_1 falls between hs_1 and $hs_{2;}$ it is therefore part of gait cycle 1 and occurs at 45% of the gait cycle (mid-stance for this subject). The relative stimulation termination timing (StT) was

similarly calculated, substituting the stimulation termination frame, StT_{frame}, for StI_{frame} in Eq 5. For the example shown in Figure 33, StT_1 occurs at ~95% of gait cycle 1, during terminal swing.

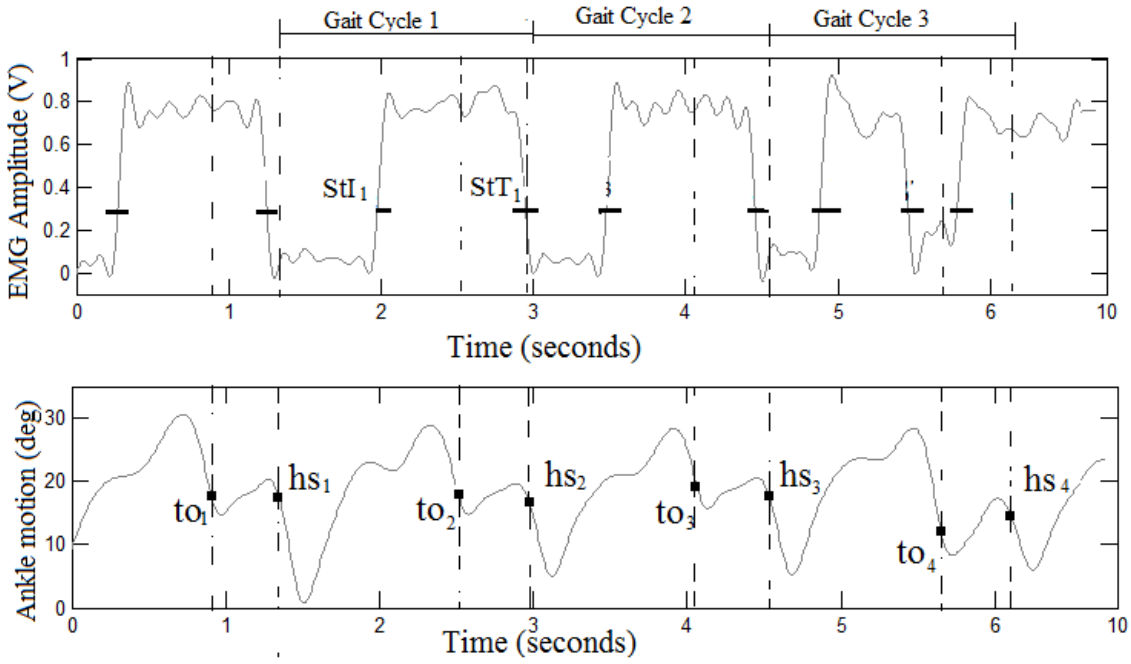

Figure 34: Stimulation periods (top) illustrating the stimulation initiation (StI) and termination (StT) timing during tilt sensor based stimulation; the corresponding sagittal plane ankle motion data are also shown (bottom), illustrating the respective HS and TO events for each gait cycle.

Since peroneal nerve stimulation provides ankle dorsiflexion and therefore facilitates toe clearance during swing for individuals with foot drop, the StI timing was also expressed relative to the start of swing, as shown below:

$$StI_{swing} = P_{swing} - StI \qquad \text{(Eq 6)}$$

where P_{swing} is the onset of swing (% gait cycle), and StI is the stimulation initiation (% gait cycle). Positive StI_{swing} values reflect stimulation initiation prior to swing, perhaps during the pre-swing functional stance phase. Negative StI_{swing} values indicate that the stimulation initiation occurred during swing; these negative StI_{swing} values reflect delayed stimulation that will likely result in insufficient toe clearance during swing.

During able-bodied walking, the ankle dorsiflexors are also active during early to mid-stance to prevent foot slap. As such, WalkAide stimulation of the tibialis anterior may continue

through early stance. StT timing was therefore expressed relative to the end of swing to determination whether stimulation terminated after swing (>100% gait cycle), extending into early stance of the subsequent gait cycle.

$$StT_{swing} = StT + 100 \qquad \text{(Eq 7)}$$

The mean StI and StT timing during each trial, as well as the variability (standard deviation) in these parameters, was calculated for each subject and treadmill orientation for both tilt and heel sensor-based stimulations. StI and StT timing were compared amongst treadmill orientations to determine if the orientation influences the sensor stimulation timing.

3.7 F-SCAN DATA PROCESSING

Since kinetic data could not be acquired due to the treadmill, F-scan insoles were used to acquire plantar pressure data for each limb during ambulation in both treatments. However, initial attempts to contrast heel loading between the AFO and WalkAide treatments were unsuccessful as the rigid plantar surface of the AFO created wrinkles in the F-scan sensor, introducing elevated pressure artifacts (Figure 34). Therefore, F-scan insoles were used during WalkAide trials only, facilitating comparison of WalkAide heel sensor data with F-scan heel loading data.

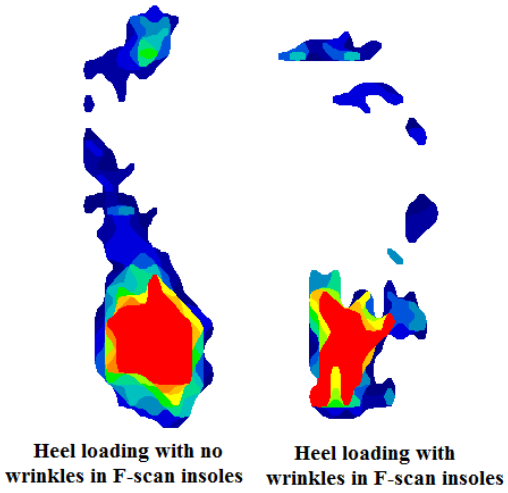

Figure 35: Plantar pressure errors or artifacts introduced by rigid AFO plantar surface [no AFO (left), AFO (right)].

Due to complications with the WalkAide heel sensor, heel sensor data were not acquired for two subjects (S7 and S8). As such, the magnitude and duration of theoretical heel sensor-based stimulation was based on the F-scan sensor.

This section summarizes the plantar pressure data processing to assess heel loading, the comparison of F-scan heel loading versus WalkAide heel sensor loading, and the estimated heel loading-based stimulation, stimulation reliability and stimulation timing using the F-scan for the aforementioned two subjects (S7 and S8).

3.7.1 Heel Plantar Pressure

The F-scan plantar pressure data (sampled at 50 Hz) were reviewed using Tekscan software. A box approximating the WalkAide heel sensor dimensions (2 cm by 2 cm) and location was defined to assess F-scan heel loading (Figure 35). For each 8 second data trial, mean heel pressure data for the respective heel box region were evaluated.

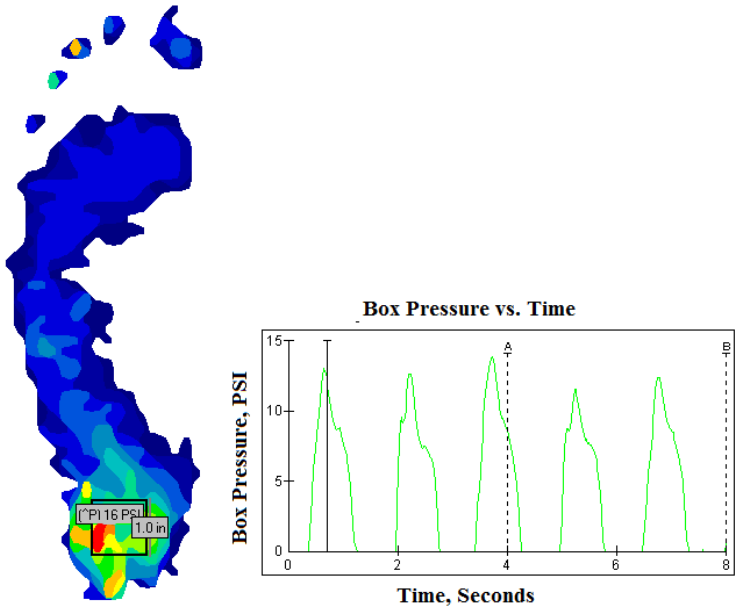

Figure 36: Heel loading via the F-scan insole [plantar pressure and heel box (left) and corresponding mean heel box loading time series (right)].

3.7.2 Comparison of F-Scan versus WalkAide Heel Loading

For the six subjects for whom WalkAide heel sensor data were available, the F-scan heel loading data were compared to that obtained with the WalkAide, to validate the WalkAide heel sensor for possible stimulation control.

The heel loading and unloading periods based on the F-scan heel box were determined using threshold detection, see Figure 36. Near-zero (psi) loading was considered unloading periods, while times in which the pressure increased above zero were considered loading periods. Similar to the analysis of the WalkAide heel sensor, a threshold was used to define specific loading/unloading periods. This threshold was based on a specific percent of the loading (2-13%).

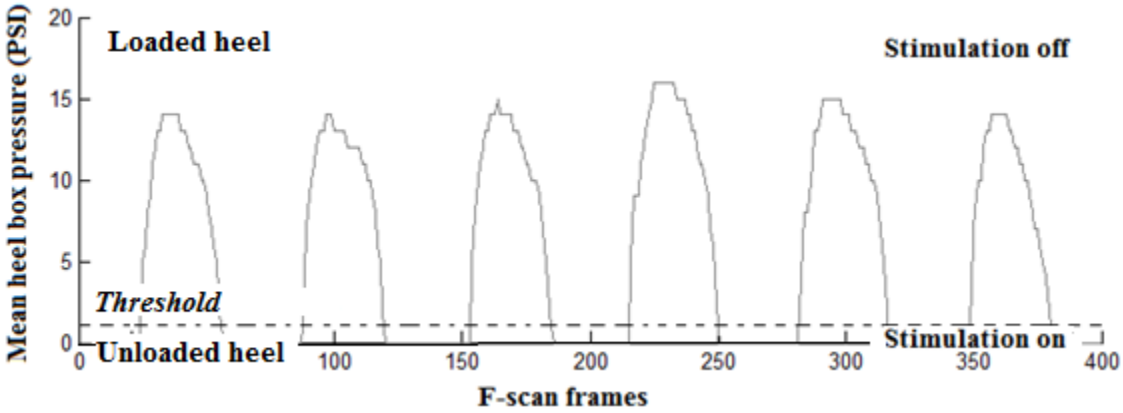

Figure 37: F-scan heel loading and unloading based on threshold detection techniques.

The mean and maximum pressure during loading, and the duration of loading and unloading periods of the F-scan box were compared to the WalkAide heel sensor data collected from six of the study's subjects. The mean and maximum heel box values and WalkAide heel sensor values for each loading period were determined for all trials and each treadmill orientation. The duration of each loading and unloading period based on both the heel box and the WalkAide heel sensor were also determined for all trials and each treadmill orientation. The F-scan heel box values were plotted against the corresponding WalkAide heel sensor value in a scatter plot. Separate plots were created for each of the four variables: maximum pressure during loading,

mean pressure during loading, heel loading duration, and heel unloading duration. Linear regression was performed for the four graphs to determine the potential correlation between F-scan insole heel box data and WalkAide heel sensor data.

3.7.3 Estimation of F-Scan Heel Box Based Stimulation

Due to complications with the WalkAide heel sensor, heel sensor data were not acquired for two subjects (S7 and S8). For these subjects, the F-scan heel box (Section 3.7.1) was used to estimate theoretical heel loading and unloading, with thresholds set at 0.1 psi, see Figure 36. The frame immediately before the box value exceeded the threshold was considered stimulation initiation, while the first frame after the box value dropped below the threshold was considered stimulation termination. Stimulation initiation and termination frames were determined for all trials of subjects S7 and S8.

Since F-scan data were sampled at 50 Hz, contrary to the kinematic (120 Hz), EMG (1800 Hz), and WalkAide (25 Hz) data, and the data could not be synchronized through hardware, the F-scan pressure under the entire foot was used to determine HS and TO. To estimate HS and TO events, a "box" corresponding to the entire plantar surface was created. The mean plantar pressure for this box was evaluated; periods of zero pressure were considered swing (i.e. the foot was off the ground) and the periods of loading were considered stance (i.e. the foot was on the ground). The frames corresponding to the beginning and end of stance were stored as HS and TO frames, respectively. With frames for the HS, TO, stimulation initiation, and stimulation termination, the StR and relative StI and StT (percent gait cycle) could be calculated using Equation 4 and Equation 5, respectively.

3.8 STATISTICAL ANALYSIS AND HYPOTHESIS TESTING

Statistical analyses were conducted to test the research hypotheses, contrasting treatments in terms of temporal and kinematic symmetry (Hypothesis 1a) and efficacy or toe clearance

(Hypothesis 1b), treadmill orientation effect on the WalkAide tilt sensor-based stimulation in terms of stimulation reliability (Hypothesis 2a) and timing (Hypothesis 2b), and WalkAide tilt versus heel sensors in terms of stimulation reliability (Hypothesis 3a) and timing (Hypothesis 3b).

To determine whether parametric or nonparametric analyses should be conducted, the normality of the data distribution was assessed. Power tests were also performed to verify sample size necessary to determine significant differences in each of the respective study parameters.

3.8.1 Hypothesis 1 testing

The first research hypothesis was that the use of a neuroprosthesis for post-stroke individuals with drop foot will improve gait symmetry [temporal symmetry (SR and gait asymmetry index) and sagittal plane kinematic GA] and efficacy (toe clearance) for all treadmill orientations when compared to ambulation with an AFO. The normality of respective parameters was assessed using the Kolmogorov-Smirnov test (K-S test) [52]. This test indicated that both the temporal parameters and toe clearance were non-normally distributed. Therefore, this hypothesis was tested using nonparametric statistical analyses.

Initial analysis included Wilcoxon signed rank tests (paired, two-sided) across subjects to identify parameters that differ significantly:

1) between limbs (e.g. affected vs. unaffected) of the same treatment and
2) between treatment (e.g. AFO versus WalkAide) of the same limb for given treadmill orientation.

This test assumes that the two data sets or pairs (affected vs. unaffected limbs and AFO vs. WalkAide treatments for the same subject) are independent random samples from continuous, symmetric distributions [52]. The null hypothesis for these Wilcoxon signed rank tests was that no difference exists between median measures of the respective pair. Significance levels of 5% ($p=0.05$) and 1% ($p=0.01$) were documented.

Since measures of kinematic gait asymmetry were based on differences between limbs, only a single measure of asymmetry was calculated for each parameter (e.g. mean ankle or knee angle during IS). The standard errors of the respective parameter for both the affected and unaffected limb were individually determined and combined, using equation 9, to determine the standard error for the treatment ($SE_{treatment}$):

$$SE_{treatment} = \sqrt{SE_{affected}{}^2 + SE_{unaffected}{}^2} \qquad \text{(Eq 9, adapted from [52])}$$

where $SE_{affected}$ is the standard error of the respective parameter for the affected limb and $SE_{unaffected}$ is the standard error of the respective parameter for the unaffected limb.

Using these standard errors for both the WalkAide and the AFO treatments, a Z-test was performed (Eq 10) for each subject, contrasting the two treatments:

$$Z - score = \frac{GA_{AFO} - GA_{WalkAide}}{\sqrt{SE_{AFO}{}^2 + SE_{WalkAide}{}^2}} \qquad \text{(Eq 10, adapted from [52])}$$

where SE_{AFO} is the standard error of the AFO treatment, $SE_{unaffected}$ is the standard error of WalkAide treatment, GA_{AFO} is the gait asymmetry for the AFO, and $GA_{WalkAide}$ is the gait asymmetry for the WalkAide – all for the respective parameter of interest. The null hypothesis for these Z-tests was that the asymmetry measures for the WalkAide and AFO do not differ. This test assumes that the asymmetry measures are random samples from populations with known means and variances, and are independent of one another [52]. Significance levels of 5% (p=0.05) and 1% (p=0.01) were again documented. To compare the GA of all subjects, the absolute value of each GA was determined for both treatments. A Wilcoxon rank sum test was then performed to determine if the GA differed significantly between treatments across all study subjects.

Since all parameters were determined to be non-normally distributed based on the K-S test, nonparametric Freidman testing of the temporal parameters, toe clearance and gait asymmetry measures was conducted to identify parameters that differ significantly with treadmill orientation for a given treatment (AFO/WalkAide) and limb (affected/unaffected). The Friedman

test assumes that all observations are mutually independent and that all data have the same continuous distribution. The Friedman tests were set up in a two-way layout (e.g. rows= blocks=subjects, columns=factors= treadmill orientations) to test the effects of the treadmill orientations on the given parameter for specific limb while using one treatment [52]. The null hypothesis for the Friedman test was that the treadmill orientation did not affect the given parameter; significance levels of 5% (p=0.05) and 1% (p=0.01) were documented. If significant differences were found, post-hoc testing (Wilcoxon sign rank tests with the Bonferroni correction for multiple comparisons [53]) was conducted to assess whether these differences occurred between level-inclined, level-declined, or inclined-declined treadmill orientations.

3.8.2 Hypothesis 2 testing

The second research hypothesis stated that WalkAide tilt sensor-based stimulation reliability (Hypothesis 2a) and timing (Hypothesis 2b) will differ for ambulation on level versus on non-level (e.g., level versus incline and level versus decline) surfaces. The WalkAide StR and stimulation timing were found to be non-normally distributed based on the K-S test. Therefore non-parametric Friedman testing was again conducted to determine whether StR or stimulation timing (StI and StT) differed significantly non-level walking (e.g., level versus incline and level versus decline) for tilt sensor-based stimulations. Post-hoc Wilcoxon signed rank tests with Bonferroni correction for multiple comparisons were completed to assess whether these differences occurred between level-inclined or level-declined walking. Significance levels of 5% (p=0.05) and 1% (p=0.01) were documented for all Friedman tests.

3.8.3 Hypothesis 3 testing

The third research hypothesis contrasts WalkAide tilt versus heel sensor-based stimulation in terms of stimulation reliability (Hypothesis 3a) and timing (Hypothesis 3b). During non-level walking, the heel sensor is hypothesized to result in improved StR and less

variable stimulation timing. Wilcoxon signed rank testing was conducted to determine whether StR and variability in stimulation timing (StI and StT) differed significantly between sensors (tilt and heel) during the non-level (inclined and declined) walking trials. Significance levels of 5% (p=0.05) and 1% (p=0.01) were documented.

3.8.4 Power tests

Large sample sizes are often required when analyzing data with non-parametric statistical tests. To determine whether additional subjects are needed to detect statistically significant differences, power analyses were conducted. These power analyses were used to determine both the power associated with the current sample size (n=8 corresponding to the full population subset) and the sample size necessary to achieve the desired power of 0.80 at the 0.05 level of significance. These tests were repeated for each of the temporal parameters, temporal symmetry measures, gait asymmetry measures, and toe clearance values.

Three separate *a priori* power analyses were conducted using G-Power (v3.1.2, [54] to determine the number of subjects necessary to detect statistically significant differences at the 0.05 level of significance, assuming small (0.2), medium (0.5) and large (0.8) effect sizes [55] and non-normal data distributions.

A priori and post hoc power analyses based on the observed effect sizes in the respective parameters (temporal parameters, temporal symmetry, GA, and toe clearance) were also conducted, both to determine the number of subjects necessary to detect statistical significance and the statistical power associated with the current sample size. A priori power analysis for each Wilcoxon rank sum analysis was computed using G-Power to determine the sample size needed to achieve the desired power (P=0.80). Priori power test input included the distribution (non-normal), mean and standard deviation of the respective groups for the given parameter, and significance level (α=0.05) [55]. Post hoc test inputs included the sample size (n=8), distribution (non-normal), and the mean and standard deviation of the respective groups for a given parameter

(e.g. affected limb AFO/WalkAide data for temporal parameters, AFO/WalkAide GA measure for kinematic asymmetry, and affected limb AFO/WA values for toe clearance).

3.9 SUMMARY

Post-stroke individuals with foot drop who met research selection criteria were recruited from the Milwaukee area and tested in two foot drop treatments, an AFO and WalkAide neuroprosthesis. After acclimating to each device for four weeks, gait analysis (two minute walking trials on a level, inclined and declined treadmill) was conducted. Acquired data included lower extremity joint kinematics and subject heart rate; for the WalkAide sessions data included plantar pressure and tibialis anterior EMG, as well as WalkAide tilt sensor, heel sensor and stimulation data.

Using heel strike and toe off event markers, the temporal parameters were calculated. These gait events were also used to review kinematic data during the functional phases of swing and stance. Gait asymmetry between the affected and unaffected limbs at the knee and ankle was assessed for both treatments (AFO/WalkAide). Temporal parameters, asymmetry measures and were used to contrast the temporal and kinematic gait symmetry (Hypothesis 1a) and efficacy in terms of toe clearance (Hypothesis 1b) for the two treatments. The WalkAide tilt sensor stimulation reliability and initiation/termination timing were used to investigate the reliability (Hypothesis 2a) and stimulation timing consistency (Hypothesis 2b) of the WalkAide stimulation on level and non-level surfaces for the tilt sensor-based programming. Stimulation reliability (Hypothesis 3a) and variability in initiation/termination timing (Hypothesis 3b) were contrasted for tilt sensor and theoretical heel sensor-based programming to determine if the heel sensor produced more reliable and consistent stimulation during non-level walking.

CHAPTER 4: RESULTS

Two gait analysis sessions were conducted on each of the eight subjects who completed this study, once while ambulating in the study-provided AFO and the other while ambulating in the WalkAide. Kinematic and heart rate data were collected for all sessions while plantar pressure, EMG, and WalkAide tilt, heel sensor, and stimulation timing data were collected during WalkAide gait sessions.

Gait events, HS and TO, were determined using the velocity of the heel and toe markers, respectively. These events were used to determine temporal parameters such as cadence and stance duration. Each trial was divided into the respective gait cycles of the affected and unaffected limbs, as well as the functional phases of stance and swing. Gait asymmetry, which compared the dissimilarity of the mean, maximum and minimum sagittal plane motion of the ankle and knee, was calculated for each functional phase. The ankle ROM and toe clearance during swing phase were also determined. Temporal parameters, gait asymmetry, ankle ROM, and toe clearance were used to compare the effects of treatments.

WalkAide tilt sensor-based stimulation reliability and stimulation initiation and termination timing were determined using the gait events and anterior tibialis EMG data. Theoretical heel sensor-based stimulation reliability and timing were determined using the gait events and WalkAide heel sensor data. The WalkAide heel sensor and F-scan data were reviewed to determine whether these data were correlated so that F-scan plantar pressure data might be use to estimate stimulation initiation/termination for subjects for whom WalkAide heel sensor data were not available.

This chapter summarizes the results from the respective gait analysis sessions.

4.1 SUBJECT CHARACTERISTICS

4.1.1 Subject Information

Eight post-stroke individuals were selected from the greater Milwaukee area for participation in this study. Subjects were between the ages of 48 – 76 years, had experienced a stroke at least six months prior to testing (mean of 3.8 years), and had an average lower extremity Fugl-Meyer score of 24.4. Three subjects had right hemiparesis and five subjects had left hemiparesis. All subject characteristics are summarized in Table 9.

Subjects S2 and S7 exhibited large variability in WalkAide stimulation reliability and initiation/termination timing data (see Section 4.5.3). Analyses were conducted both with (All) and without (Group A) these subjects.

Table 9: Subject Characteristics
***study outliers: data excluded from Group A analyses**
[+] original stroke 9.5 years (mini stroke 4.5 years) prior to research participation
[++] original stroke 1 month before stroke which caused foot drop
WA=WalkAide

	Sex	Age (years)	Affected Side	Lower extremity Fugl-Meyer	Time since stroke (years)	Weight (kg)	Height (cm)	AFO use prior to study	First study treatment
S2*	M	51	L	27	4.5[+]	123.15	185.42	articulated	WA
S4	M	48	R	23	7.5	75.07	185.42	solid	WA
S5	M	76	L	24	0.5	82.33	163.83	solid	AFO
S6	F	65	R	23	2.5	60.21	154.94	solid	WA
S7*	M	53	L	24	8	85.73	177.8	solid	AFO
S8	M	51	R	21	2	126.32	180.34	solid	AFO
S10	F	56	L	28	2	105.91	167.64	none	WA
S12	M	58	L	25	3.5[++]	119.29	177.8	solid	AFO
Mean (std)		57.3 (9.2)		24.4 (2.3)	3.8 (2.7)	97.3 (24.8)	174.1 (10.9)		

4.1.2 Functional Assessment

To assess differences in subject impairment, Modified Ashworth Scale (Table 10), passive ankle ROM (Table 11) and muscle strength (Table 12) measures were acquired during the

initial fitting for the first study device and at the beginning of each gait session. For all subjects, passive ankle dorsiflexion was greater after WalkAide acclimation (e.g. beginning of the WalkAide gait session) than after AFO acclimation. Treatment did not consistently affect Modified Ashworth Scale scores, plantar flexion ROM or ankle dorsiflexion strength.

Heart rate was monitored during all walking trials to identify any potential subject fatigue during testing. As shown in Table 13, the mean heart rate did not vary by more than 15 beats per minute within a given gait session and never exceeded 110 beats/minute. Subject fatigue was not observed during testing.

Table 10: Modified Ashworth Scale scores

	Ashworth		
	Initial Fitting	WalkAide	AFO
S2	3	2	3
S4	3	1	2
S5	0	3	3
S6	1	1	0
S7	3	2	2
S8	4	4	3
S10	1	0	2
S12	4	3	4

Table 11: Passive ankle ROM

	Dorsiflexion ROM (deg)			Plantar flexion ROM (deg)		
	Initial Fitting	WalkAide	AFO	Initial Fitting	WalkAide	AFO
S2	5	7	2	35	35	30
S4	10	20	15	30	35	35
S5	10	8	5	30	35	30
S6	10	8	5	24	28	30
S7	5	5	5	25	30	25
S8	5	5	3	25	25	35
S10	5	10	5	30	25	30
S12	5	5	5	30	30	25

Table 12: Ankle dorsiflexion strength based on MMT

	Dorsiflexion strength		
	Initial Fitting	WalkAide	AFO
S2	2-	3	3
S4	1	1	1
S5	2	2+	2+
S6	3	3	3
S7	2	3-	3-
S8	2	3+	2+
S10	2+	3-	2
S12	3-	3-	3-

Table 13: Mean heart rate (beats/min) for subjects, treatments and treadmill orientations

		S2	**S4**	**S5**	**S6**	**S7**	**S8**	**S10**	**S12**
	Decline	100	84	78	95	92	108	77	100
	Level	94	80	84	97	90	105	86	102
	Incline	89	81	78	100	92	94	81	102
WA	**Mean**	**94**	**82**	**80**	**97**	**91**	**102**	**81**	**101**
	Decline	77	77	90	80	81	101	85	98
	Level	72	76	91	78	80	101	88	100
	Incline	81	78	100	82	79	110	91	102
AFO	**Mean**	**77**	**77**	**94**	**80**	**80**	**104**	**88**	**100**

4.2 TEMPORAL PARAMETERS

Cadence, gait cycle duration, stance and swing phase durations, and double limb support (DLS - initial, second and total) durations were computed using the gait events. Cadence was determined for each ten second walking trial; phase and support durations were evaluated for each full gait cycle within a trial for both the affected and unaffected limbs. Mean temporal data were evaluated for the respective affected and unaffected limbs for each subject and treadmill orientation. The Wilcoxon rank sum test was used to compare affected versus unaffected limb values and WalkAide versus AFO treatments for each treadmill orientation.

4.2.1 Temporal Parameters for All Subjects

The mean cadence and gait cycle duration of all subjects are summarized in Figures 38 both as a function of treatment (AFO versus WalkAide) and limb (affected versus unaffected) for each treadmill orientation. As the research hypotheses address treatment efficacy, only the treatment differences will be presented. The mean cadence ranged from 61.2 to 65.7 steps/min and the mean gait cycle duration ranged from 1.81 to 1.94 seconds. Although the cadence was generally faster with the AFO versus WalkAide, these differences between treatments were not statistically significant. The gait cycle duration was typically prolonged for AFO ambulation during inclined walking; during declined and level walking, the gait cycle duration was prolonged for WalkAide ambulation. However, these treatment differences in gait cycle duration were not statistically significant.

Although no statistically significant differences in gait cycle duration were found between treatments, the respective stance and swing durations were also examined. The mean stance and swing durations of all subjects with respect to both limb (affected vs. unaffected) and treatment (WalkAide vs. AFO) are summarized in Figure 39. The mean stance duration ranged from 1.31 to 1.46 seconds, regardless of treatment or treadmill orientation, with greater inter-subject variability noted during WalkAide ambulation The stance duration was prolonged (for both the affected and unaffected limbs) with the WalkAide vs. AFO for all treadmill orientations; these differences, however, were only statistically significant for level walking (both limbs) and declined walking (unaffected limb only).

The mean swing [or single limb support (SLS) of the contralateral limb] duration ranged from 0.50 to 0.56 seconds, regardless of treatment or treadmill orientation. Greater inter-subject variability in swing duration was again noted during WalkAide ambulation. For the unaffected limb, longer swing (or longer affected limb SLS) durations were observed during AFO versus WalkAide ambulation for all treadmill orientations; these differences were statistically significant

for level and inclined orientations. In contrast, for the affected limb, longer swing (or longer unaffected SLS) durations were observed with the WalkAide versus AFO during declined walking (not significant); longer affected limb swing durations were observed with the AFO during level (not significant) and inclined (significant) walking.

The gait cycle was further subdivided into periods of DLS; the initial, second and total periods of DLS were examined. While the WalkAide's primary objective is affected limb dorsiflexion during swing, the WalkAide may also stimulate the tibialis anterior during the transition from swing to stance (initial DLS) and stance to swing (second DLS). Continued anterior tibialis stimulation during initial DLS may enhance ankle stability during loading response, while WalkAide stimulation during the second DLS period may limit plantar flexion needed for push off. Examining the durations of the initial and second DLS periods may provide insight into weight acceptance (e.g. loading response) and weight transfer, respectively, as well as insight into ankle function when examining subjects with unilateral deficits such as stroke [19]. Examination of the total DLS period facilitates identification of balance problems [19].

The mean periods of initial, second and total DLS are presented as a function of limb (affected vs. unaffected) and treatment (WalkAide vs. AFO) in Figure 40. Since weight acceptance occurs during initial DLS, prolonged initial DLS duration may indicate that the treatment does not fully address weight acceptance issues such as ankle and knee stability during loading response. Prolonged second DLS periods may indicate that the treatment restricts plantar flexion, potentially decreasing the efficiency of gait. The initial period of DLS was often prolonged during WalkAide (range: 0.42 to 0.49 sec) versus AFO (range: 0.37 to 0.47 sec) ambulation, with the exception of the unaffected limb during declined ambulation; all of these results were statistically significant. The latter DLS period was also prolonged during WalkAide (range: 0.43 to 0.47 sec) versus AFO (range: 0.37 to 0.41 sec) ambulation for the unaffected limb on all treadmill orientations and the affected limb on the level treadmill orientation; all of these differences were statistically significant. The affected limb exhibited prolonged second DLS for

the AFO versus the WalkAide during declined and inclined ambulation; these differences were statistically significant. Finally, although the total DLS period was generally longer with the WalkAide (0.87 to 0.92 sec) than for AFO (0.80 to 0.89 sec) ambulation, these differences were only statistically significant during level ambulation.

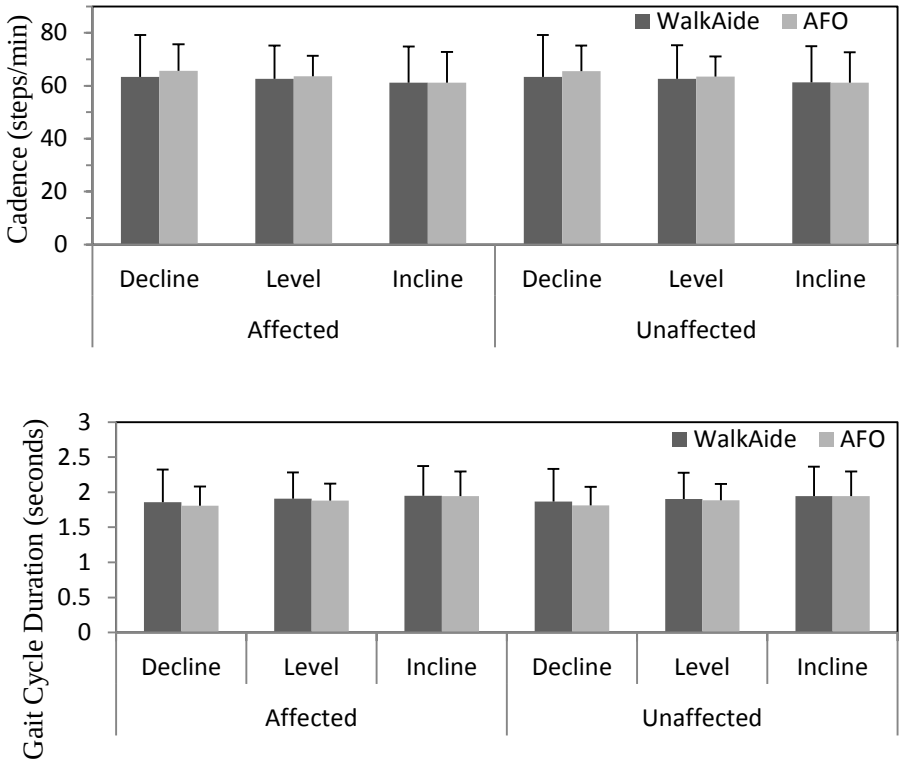

Figure 38: Mean cadence (top) and gait cycle duration(bottom) for all subjects during two minute walking trial, contrasting treatments (WA vs. AFO: $^{+}$, $^{++}$ denotes a statistically significant differences at 0.05 and 0.01 levels, respectively) and limbs (unaffected vs. affected: ^{W}, ^{A} denotes a statistically significant differences at 0.05 level for WA and AFO ambulation, respectively).

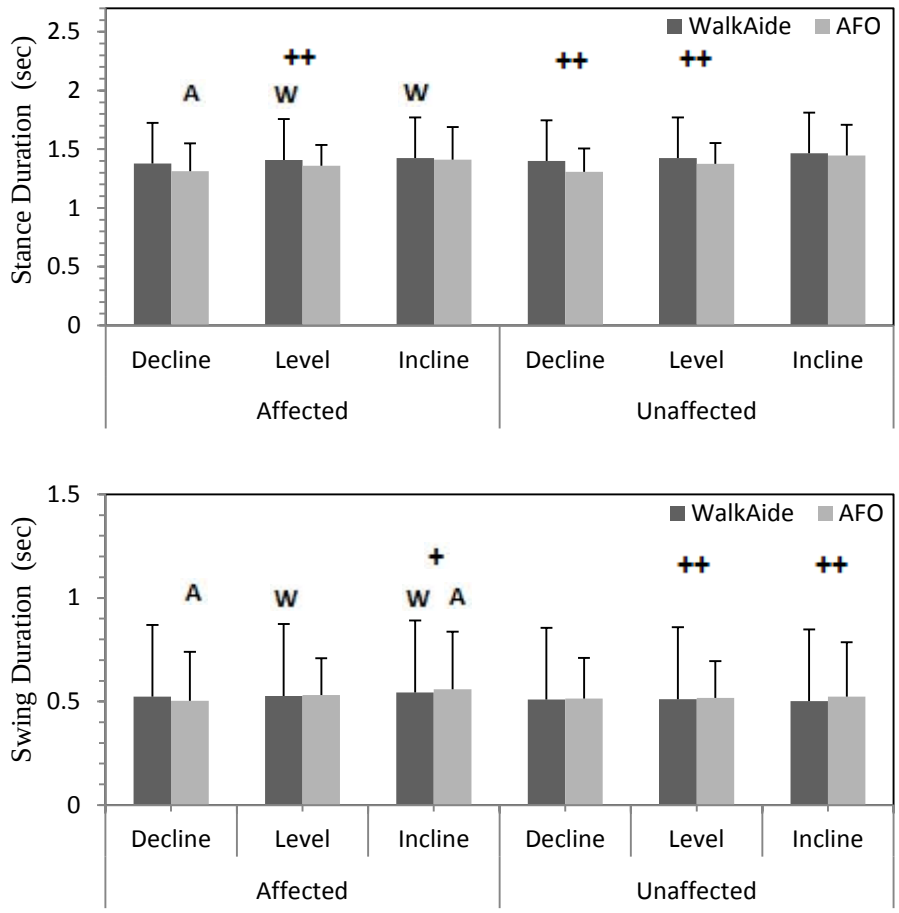

Figure 39: Mean stance (top) and swing (bottom) duration of all subjects, contrasting treatments (WA vs. AFO: $^{+}$, $^{++}$ denotes a statistically significant differences at 0.05 and 0.01 levels, respectively) and limbs (unaffected vs. affected: ^{W}, ^{A} denotes a statistically significant differences at 0.05 level for WA and AFO ambulation, respectively).

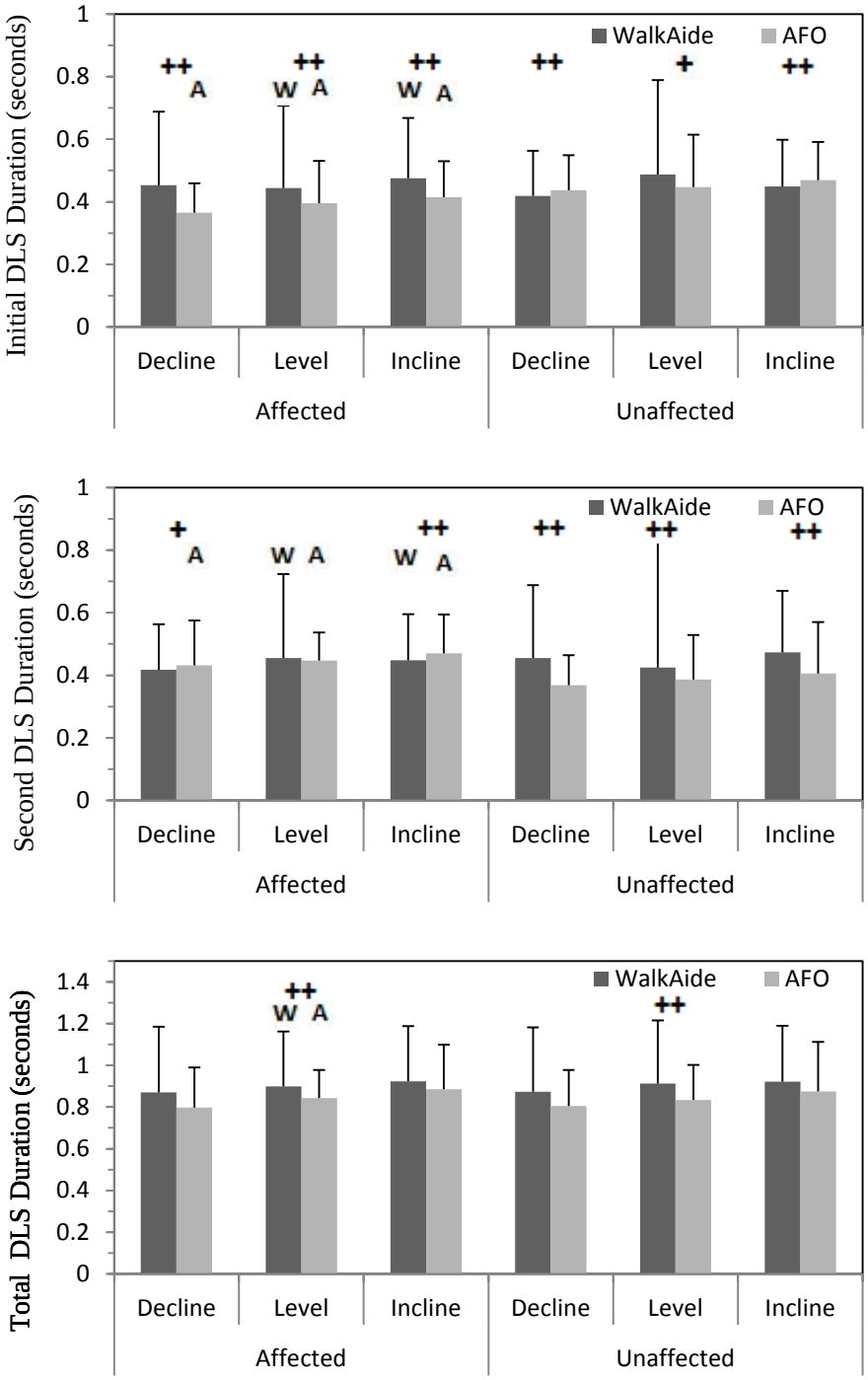

Figure 40: Initial (top), second (middle) and total (bottom) DLS duration, contrasting treatments (WA vs. AFO: $^{+}$, $^{++}$ denotes a statistically significant differences at 0.05 and 0.01 levels, respectively) and limbs (unaffected vs. affected: ^{W}, ^{A} denotes a statistically significant differences at 0.05 level for WA and AFO ambulation, respectively), for all subjects.

4.2.2 Temporal Parameters for Group A

As subjects S2 and S7 exhibited large variability in WalkAide stimulation reliability and initiation/termination timing data (see Section 4.5.3), the temporal parameters were also reviewed for Group A, a subject subset that excluded data from these two subjects. Cadence values were largely unchanged for this subject subset (Figure 41 versus Figure 38). As for all subjects, although the cadence was generally faster with the AFO than with the WalkAide, these differences were not statistically significant. Gait cycle durations were also unchanged for this population subset (Figure 41 versus Figure 38). While differences in gait cycle duration were observed between treatments, these differences were not statistically significant.

The stance duration for the Group A subset differed little from that of all subjects (Figure 42 versus Figure 39). The mean stance duration ranged from 1.37 to 1.50 seconds, regardless of treatment or treadmill orientation; greater inter-subject variability was again noted during WalkAide ambulation. Similar to that for the full population, the stance duration was generally prolonged with the WalkAide versus AFO; however, statistically significant differences were only observed during level ambulation.

Excluding subjects S2 and S7 from the analysis demonstrated fewer significant differences in swing duration between treatments (Figure 42 versus Figure 39). For the Group A subset, both the affected and unaffected limbs exhibited longer swing durations during WalkAide versus AFO ambulation on all orientation; these differences were statistically significant for the affected limb during declined ambulation and for the unaffected limb during inclined walking.

The duration of the initial, second and total periods of DLS were also evaluated for the Group A subject. Differences between WalkAide and AFO treatment (and affected and unaffected limbs) for Group A are shown in Figure 43 (versus Figure 40 for all subjects). While the inter-subject variability in DLS duration was reduced with the Group A subset, greater variability remained with the WalkAide versus AFO trials. The results demonstrated fewer

statistically significant differences between treatments when compared to the analysis of all study subjects (Figure 43 versus Figure 40). The initial period of DLS of the affected limb for the Group A subset was shorter during AFO (range: 0.37 to 0.43 sec) versus WalkAide (range: 0.42 to 0.43sec) ambulation on all treadmill orientations. Contrary to all subjects, the initial period of DLS of the unaffected limb was prolonged during AFO (range: 0.47 to 0.50 sec) versus WalkAide (range: 0.45 to 0.49 sec) ambulation on all treadmill orientations; this difference was statistically significant for declined ambulation only. Additionally, prolonged latter DLS periods for the affected limb were observed during AFO (range: 0.47 to 0.50 sec) versus WalkAide (range: 0.45 to 0.49 sec) ambulation for all orientations, but were only statistically significant during declined ambulation. For the unaffected limb, prolonged latter DLS was observed with the WalkAide compared with the AFO for all treadmill orientation; statistically significant differences were observed during declined and level walking only. Finally, the total DLS period was prolonged during WalkAide (range: 0.87 to 0.93 sec) compared to AFO (range: 0.84 to 0.93 sec) use for all treadmill orientations; these treatment differences were only statistically significant during level walking.

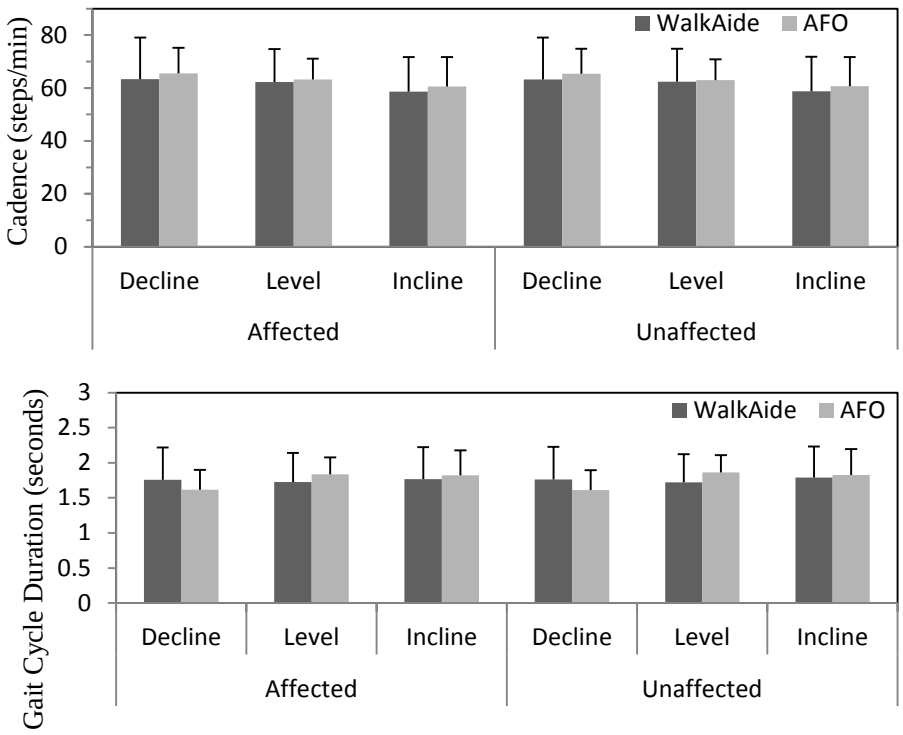

Figure 41: Mean cadence (top) and gait cycle (bottom) duration of Group A subset during two minute walking trial, contrasting treatments (WA vs. AFO: $^{+}$, $^{++}$ denotes a statistically significant differences at 0.05 and 0.01 levels, respectively) and limbs (unaffected vs. affected: ^{W}, ^{A} denotes a statistically significant differences at 0.05 level for WA and AFO ambulation, respectively).

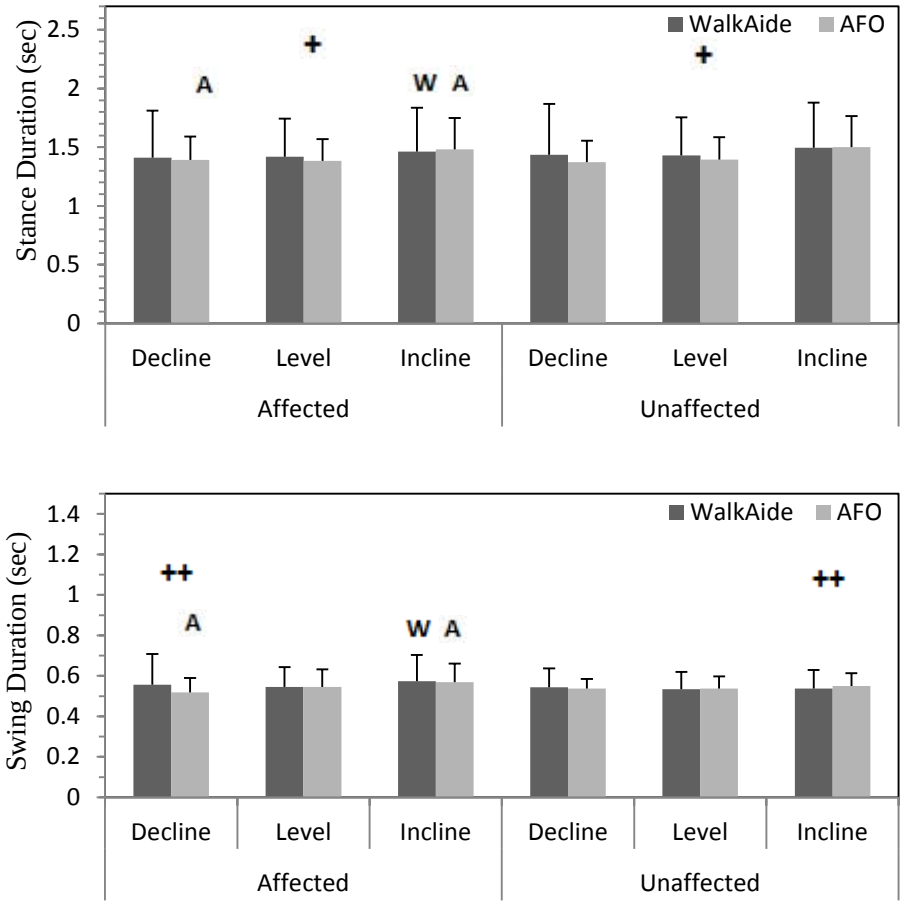

Figure 42: Mean stance (top) and swing (bottom) durations for Group A subset, contrasting treatments (WA vs. AFO: $^{+}$, $^{++}$ denotes a statistically significant differences at 0.05 and 0.01 levels, respectively) and limbs (unaffected vs. affected: ^{W}, ^{A} denotes a statistically significant differences at 0.05 level for WA and AFO ambulation, respectively).

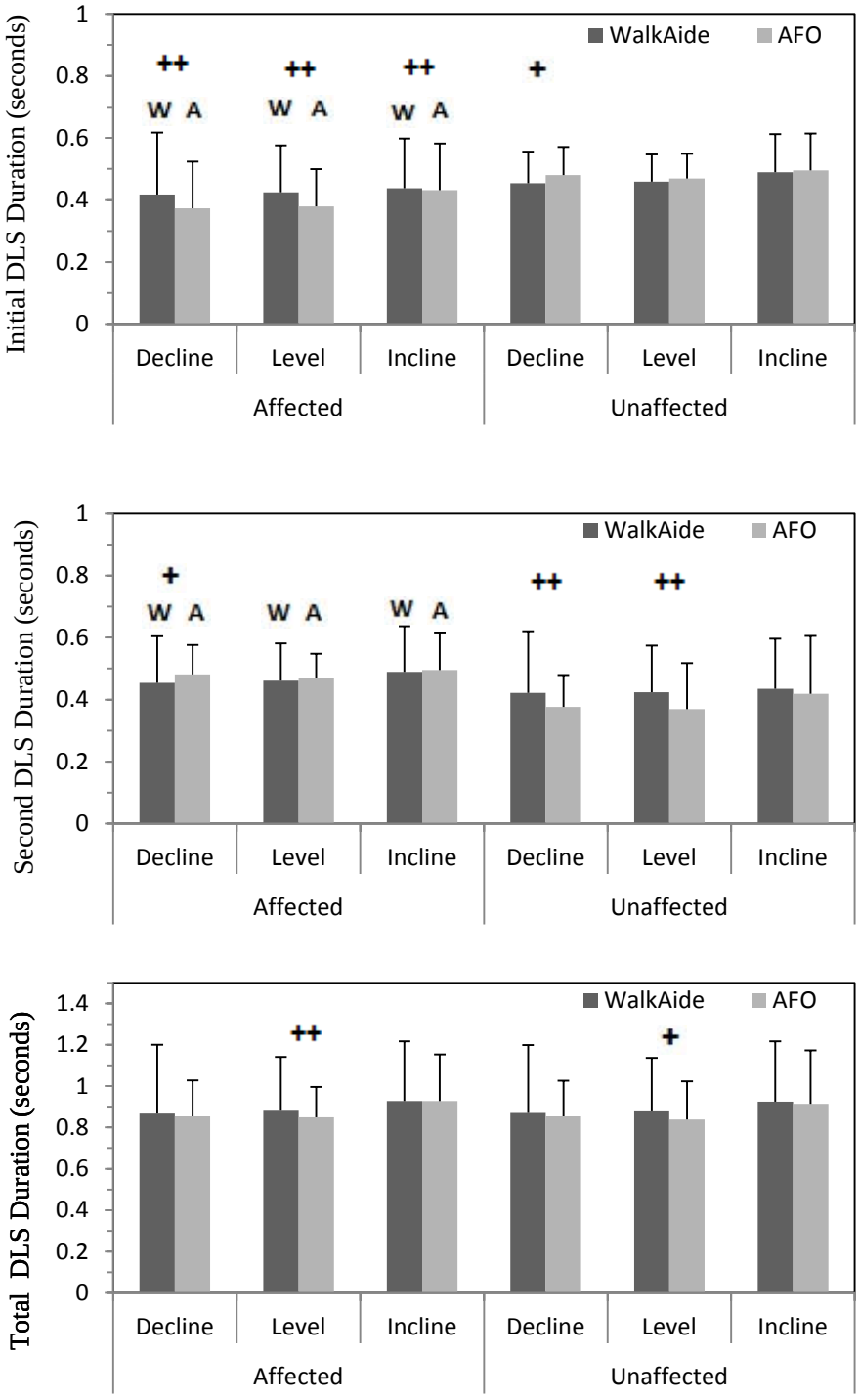

Figure 43: Initial (top), second (middle) and total (bottom) DLS duration, contrasting treatments (WA vs. AFO: $^{+}$, $^{++}$ denotes a statistically significant differences at 0.05 and 0.01 levels, respectively) and limbs (unaffected vs. affected: ^{W}, ^{A} denotes a statistically significant differences at 0.05 level for WA and AFO ambulation, respectively), for Group A subset.

4.2.3 Swing Time Variability

Prior research of post-stroke individuals ambulating in both an AFO and a neuroprosthesis has been limited (see Section 2.5). In addition to walking speed and PCI, another parameter of interest is swing time variability of the unaffected limb. Since swing duration of the unaffected limb is also the SLS of the affected limb, the unaffected limb swing time variability can provide insight into gait stability, balance and fall risk [39]. This measure is independent of walking speed and therefore may indicate changes in a subject's balance and fall risk that may not be identified by walking speed alone.

Swing time variability was quantified in terms of the coefficient of variation (CV) of the unaffected limb's swing duration. The swing time variability was determined for all subjects and again for the Group A subset, as shown in Figure 44. While the swing time variability was less during AFO (range: 3.07 to 3.87%) versus WalkAide (range: 3.96 to 4.53%) ambulation, these differences were not statistically significant.

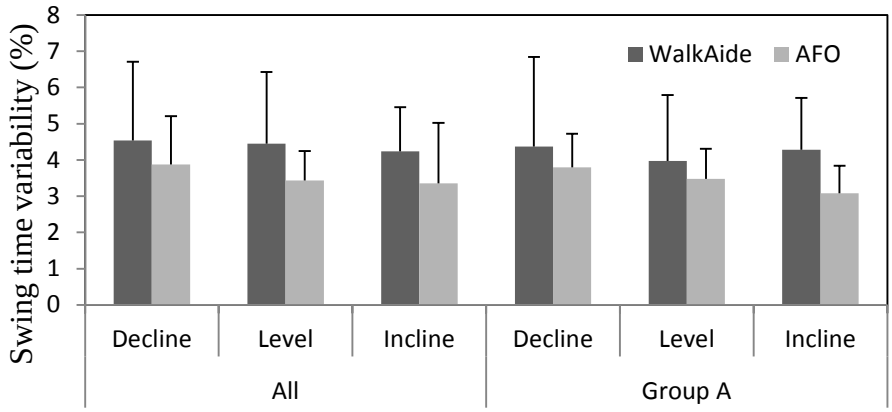

Figure 44: Swing time variability of the unaffected limb of all subjects and the Group A subset during WalkAide and AFO ambulation for all treadmill orientations.
+, ++denotes a statistically significant difference (0.05 and 0.01 levels, respectively)

4.2.4 Symmetry Ratio

In previous studies, gait symmetry was commonly evaluated by comparison of temporal parameters such as stance and swing duration. The symmetry ratio (SR), see Section 2.6.2.2, is

simply a ratio of the respective temporal parameter for the affected versus unaffected limb. As such, SR's of 1.0 reflect greater symmetry. A SR less than one indicates that the respective parameter is shorter for the affected limb; a SR greater than one indicates that the respective parameter is longer for the affected limb.

The SR for the stance and swing durations was determined for all subjects and again for Group A, as shown in Figure 45. The SR for both stance and swing ranged from 0.97 to 1.01. No significant differences in stance and swing SRs between treatments (WalkAide versus AFO) were observed for any treadmill orientation for either all subjects or for the Group A subset.

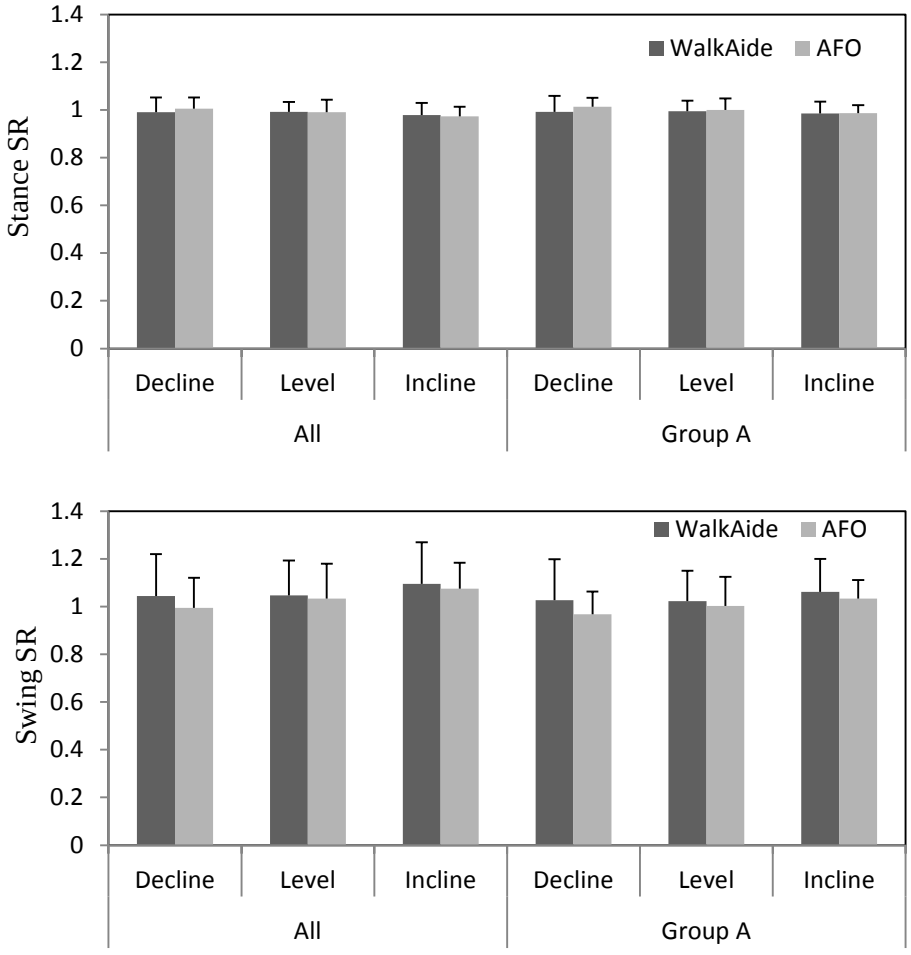

Figure 45: Mean symmetry ratio (SR) during two minute walking trial for stance (top) and swing (bottom) phase durations of all subjects and the Group A subset.
+, ++ denotes a statistically significant difference (0.05 and 0.01 levels, respectively)

4.2.5 Gait Asymmetry Index

In addition to the SR, the gait asymmetry index has also been used to quantify symmetry of temporal parameters [39]. Contrary to the SR which was normalized by the unaffected limb value, the gait asymmetry index is normalized with respect to the average value of the affected and unaffected limbs. The gait asymmetry index is similar to the SI (Section 2.6.2.2), differing only with respect to the denominator; the gait asymmetry index is double that of the SI. The gait asymmetry index has been used to compare the swing times of the affected and unaffected limbs.

The gait asymmetry index for swing duration was determined for all subjects and again for the Group A subset, as shown in Figure 46. The mean gait asymmetry index for swing duration was greater during WalkAide versus AFO ambulation for all treadmill orientations, reflecting greater temporal swing asymmetry with the WalkAide. However, these differences were not statistically significant for either all subjects or the Group A subset.

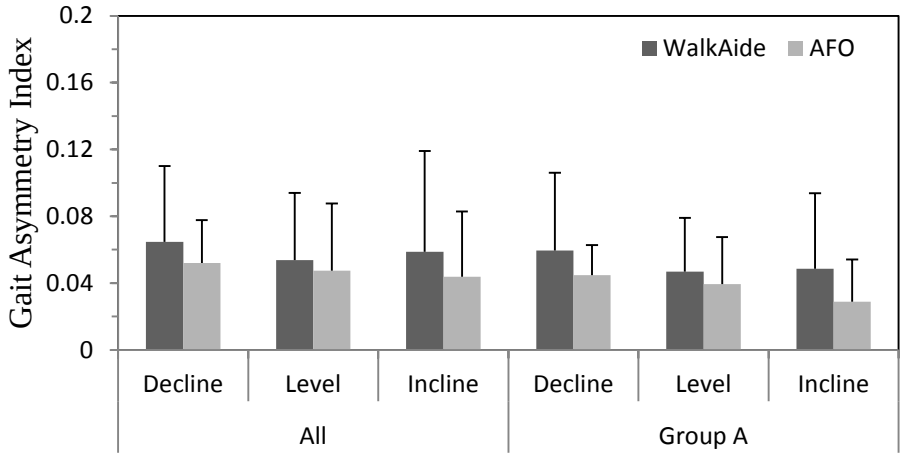

Figure 46: Mean gait asymmetry index for swing of all subjects and the Group A subset, contrasting WalkAide and AFO treatments for all treadmill orientations. +, ++ denotes a statistically significant difference (0.05 and 0.01 levels, respectively)

4.2.6 Key Temporal Parameter Findings

There were no trends or statistically significant differences in cadence or gait cycle duration between treatments for the full subject population or for the Group A subset.

For both all subjects and the Group A subset, the stance duration was a longer during WalkAide versus AFO use, with statistically significant differences observed for level and decline (all subjects only) walking. For all subjects, both limbs exhibited statistically greater swing durations with the AFO than the WalkAide during inclined walking. For the Group A population subset, the unaffected swing duration was greater with the AFO than the WalkAide during inclined walking. The swing duration of the affect limb was greater during WalkAide ambulation on all treadmill orientations, although these differences were statistically significant for declined walking only.

The initial period of DLS of the affected limb was significantly longer during WalkAide versus AFO ambulation for all treadmill orientations, for both all subjects and the Group A subset. The second period of DLS for the affected limb was significantly longer for AFO versus WalkAide ambulation on non-level surfaces for all subjects; these differences were statistically significant during declined walking only for the Group A subset. Finally, for all subjects and the Group A subset, both limbs exhibited prolonged total DLS during WalkAide versus AFO ambulation, although these differences were statistically significant during level walking only.

4.3 GAIT KINEMATICS

4.3.1 Gait Asymmetry

In contrast to the aforementioned SR and gait asymmetry index used to investigate temporal symmetry between the affected and unaffected limb, kinematic asymmetry between the affected and unaffected limb can be investigated using the GA measure proposed and defined in Section 3.5.2. This new measure facilitates comparison between subjects and joints as values are normalized with respect to the joint ROM of the unaffected limb; this measure also accounts for changes in kinematic sign that may occur during gait.

GA was evaluated for the mean, minimum, and/or maximum sagittal plane motion of each lower extremity joint for each functional phase of stance and swing, see Appendix B for mean GA for both the knee and ankle for the full gait cycle and Appendix C for individual knee and ankle GA . As this research investigates treatment for drop foot, ankle and knee joint motion were of primary interest. In addition, as drop foot causes problems with foot clearance during swing, the primary functional phases of interest were: initial swing (IS), midswing (MS) and terminal swing (TS), as well as the respective transition regions [stance to swing: pre-swing (PS), swing to stance: loading response (LR)]. The potential GA measures were further narrowed so only clinically relevant measures (mean, minimum or maximum values) were reported. For example, the maximum ankle GA during MS was reported to help assess whether difference in maximum dorsiflexion during swing might providing insight into foot clearance mechanisms.

GA of the knee for both all subjects and the Group A population subset are summarized in Figures 47 and 48, respectively (subject specific data are presented in Appendix C). For all subjects, sagittal plane knee GA was greater during the swing (AFO: 20.5-34.2, WalkAide: 15.1-35.6) than the stance (AFO: 11.0-19.0, WalkAide: 12.6-21.4) functional phases. Knee GA for the Group A subset increased with respect to all subjects during the stance functional phases (AFO: 12.1-22.8, WalkAide: 13.2-24.8), but were similar to that for all subjects for the swing functional phases (AFO: 15.6-33.4, WalkAide: 16.6-43.4). Greater knee GA was observed during WalkAide versus AFO ambulation during initial and mid-swing during both level and inclined walking. However, no statistically significant differences between treatments were found for any treadmill orientation during any of the functional gait phases of interest for either all subjects or the Group A subset.

Similarly, sagittal plane ankle GA for both all subjects and the Group A subset are summarized in Figures 49 and 50, respectively (subject specific data are presented in Appendix C). As for knee GA, ankle GA was greater during the swing (range: 16.4-50.6) than the stance (range: 16.4-36.5) functional phases of interest for AFO ambulation; during WalkAide

ambulation, ankle GA was similar during swing (range: 13.2-26.4) and stance (range: 16.2-37.1). Removal of subjects S2 and S7 from Group A analysis resulted in greater ankle GA during stance (AFO: 18.4-40.7, WalkAide 17.9-44.5), but decreased ankle GA during swing (AFO: 23.3-58.0, WalkAide: 10.4-27.2). Although no trends were noted during the various functional phases of stance, greater ankle GA was observed during AFO (range: 16.4-50.6) versus WalkAide (range: 13.2-26.4) ambulation for all functional swing phases during both level and inclined walking for both the full population and the Group A subset. Statistically significant differences in minimum (AFO: 50.6-58.0, WalkAide: 19.7-26.4) and mean (AFO: 44.5-51.6, WalkAide: 14.6-22.8) ankle GA during IS were observed for both level and inclined walking for both all subjects and Group A. The increased ankle GA during AFO ambulation during mid and terminal swing was not statistically significant.

The effects of treadmill orientation on ankle GA can be seen by contrasting the top (declined), middle (level) and bottom (inclined) graphs in Figures 49 and 50. For all subjects and the Group A subset, the ankle GA during AFO use increased as the treadmill progressed from declined to inclined, especially during the swing functional phases; ankle GA during WalkAide ambulation was not affected by treadmill orientation. Specifically, the mean and minimum ankle GA during IS significantly increased, as confirmed by Friedman tests, during AFO ambulation as the treadmill orientation progressed from declined to inclined for all subjects and for the Group A subset. Subsequent post-hoc testing of IS demonstrated significant differences in mean ankle GA between level and inclined treadmill orientations.

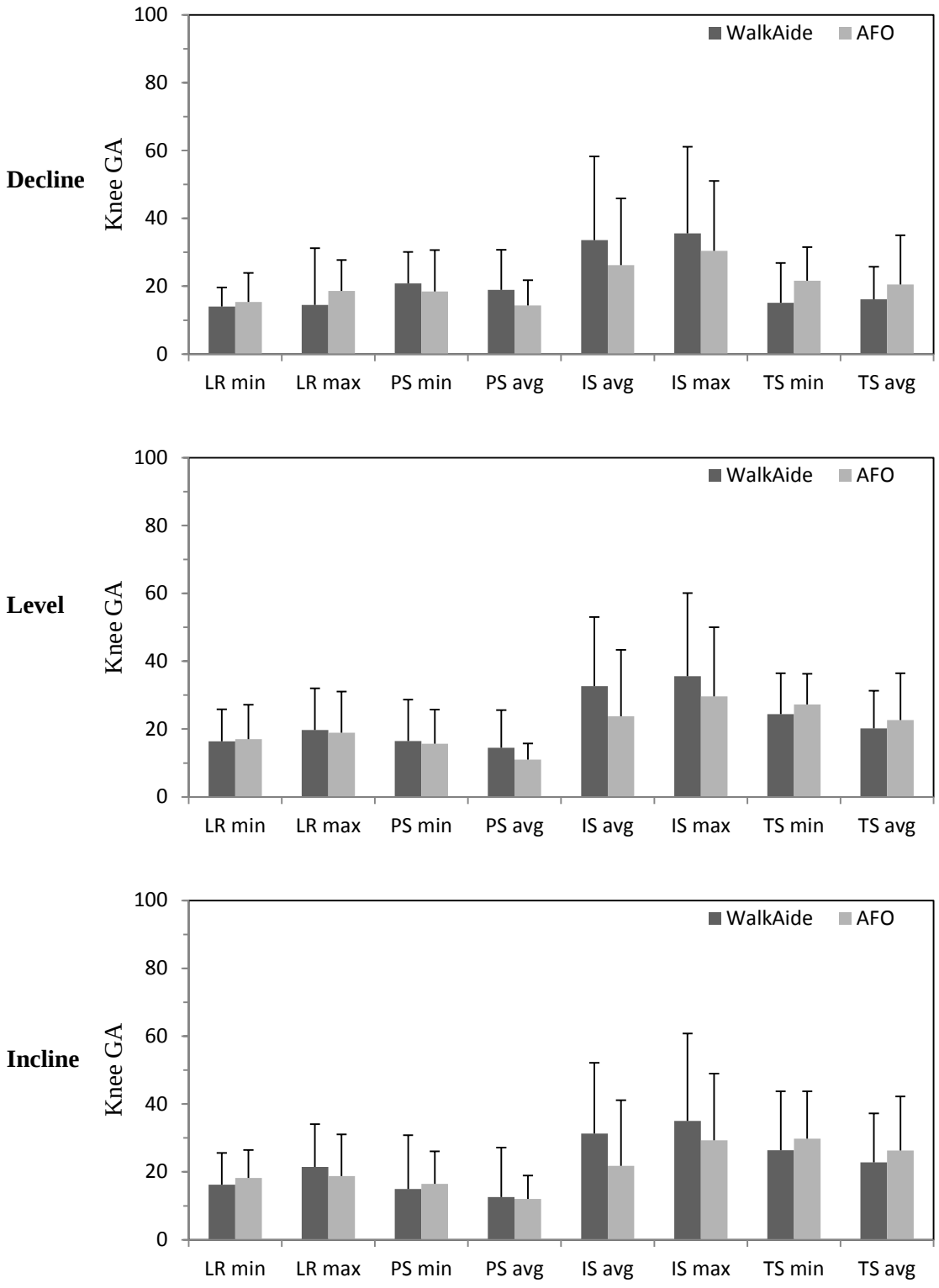

Figure 47: GA of knee motion for functional gait phases for all subjects during declined (top), level (middle), and inclined (bottom) treadmill walking.
+, ++ denotes a statistically significant difference (0.05 and 0.01 levels, respectively)

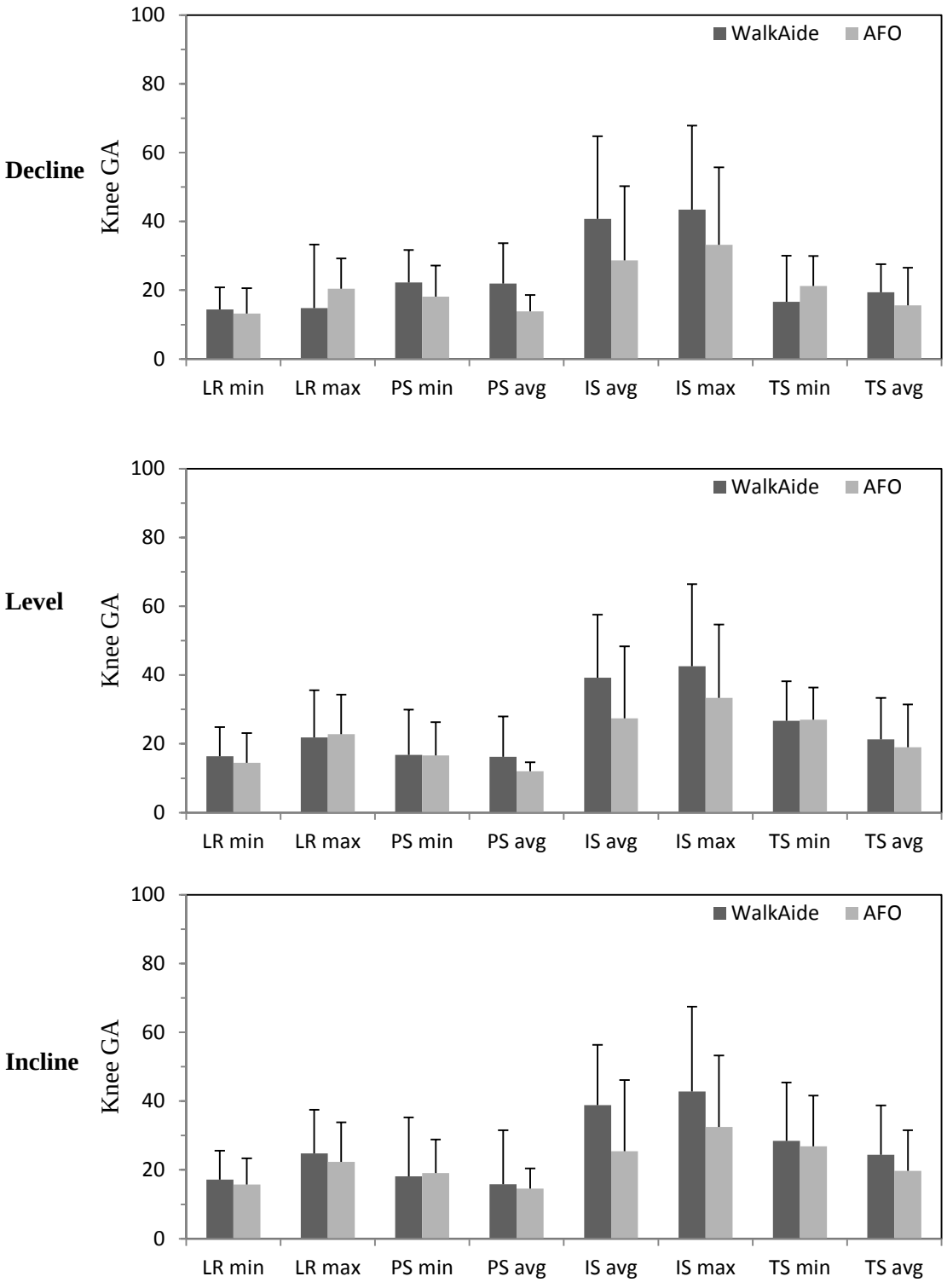

Figure 48: GA of knee motion for functional gait phases for the Group A subset during declined (top), level (middle), and inclined (bottom) treadmill walking.
+, ++ denotes a statistically significant difference (0.05 and 0.01 levels, respectively)

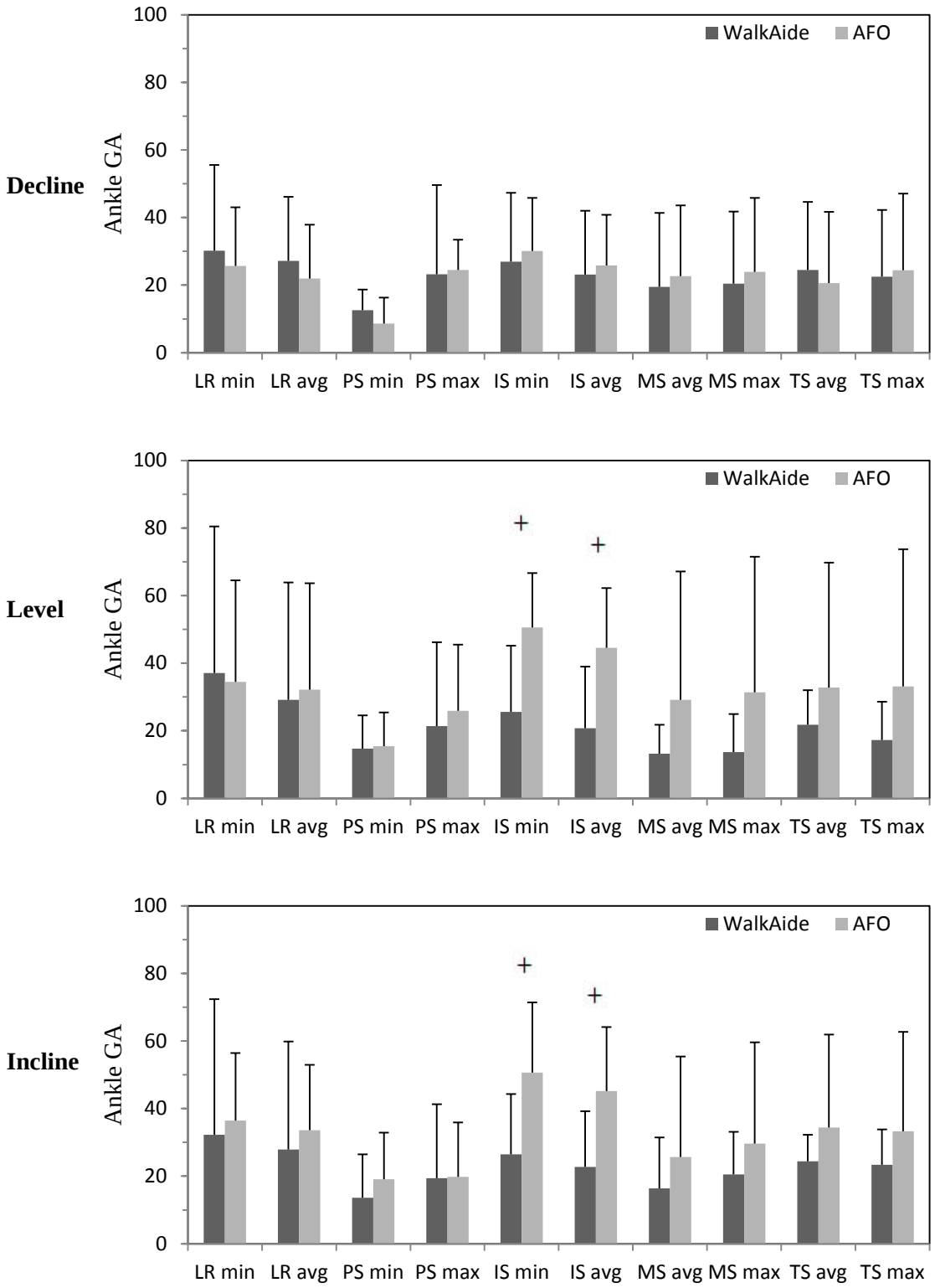

Figure 49: GA of ankle motion for functional gait phases for all subjects during declined (top), level (middle), and inclined (bottom) treadmill walking.
+, ++ denotes a statistically significant difference (0.05 and 0.01 levels, respectively)

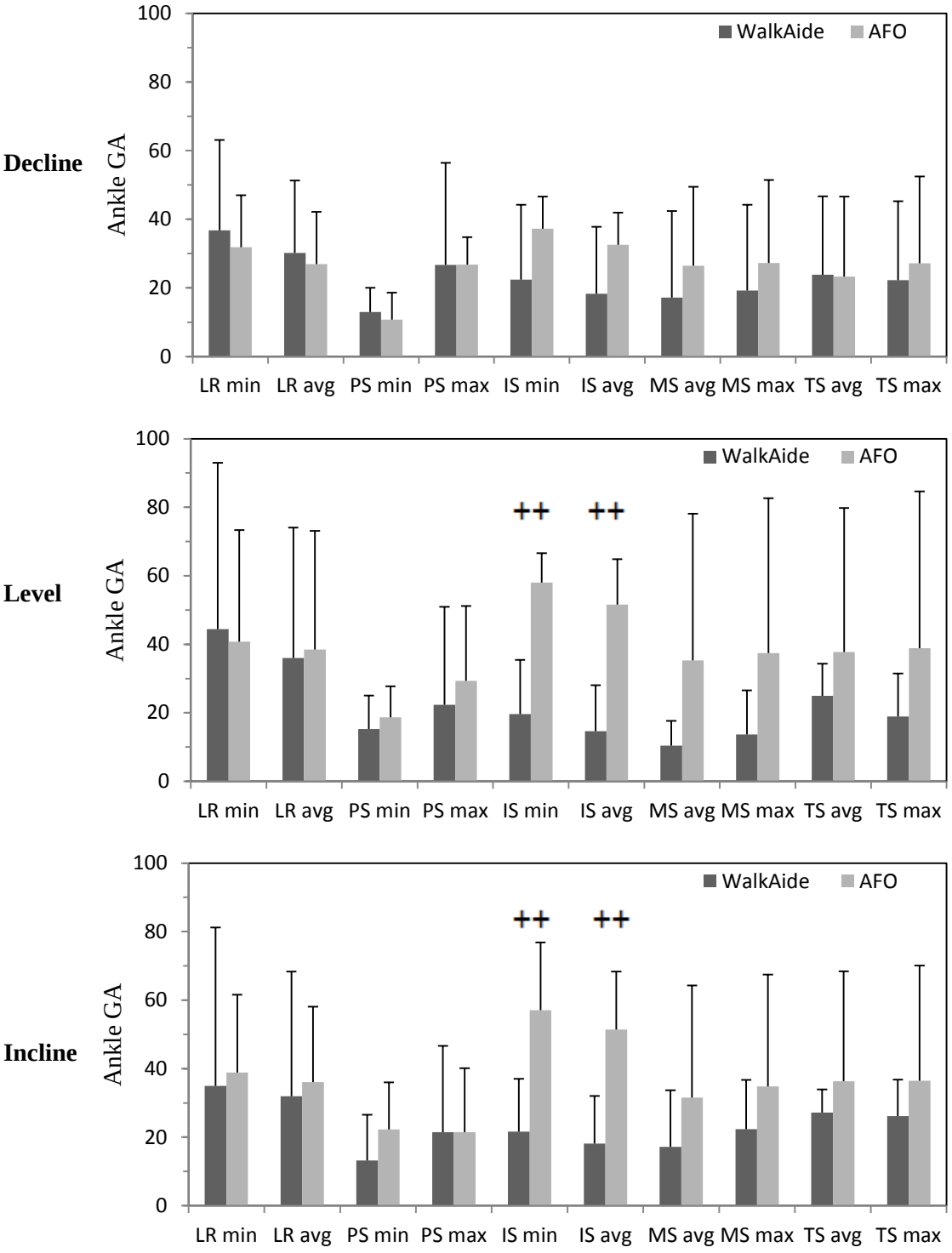

Figure 50: GA of ankle motion for functional gait phases for the Group A subset during declined (top), level (middle), and inclined (bottom) treadmill walking.
+, ++ denotes a statistically significant difference (0.05 and 0.01 levels, respectively)

4.3.2 Ankle Range of Motion

4.3.2.1 Unaffected Ankle Range of Motion

The treatment differences in ankle GA during swing may be attributed, at least in part, to differences in unaffected ankle ROM, as ankle GA is normalized with respect to the unaffected ankle ROM over the full gait cycle, see Equation 3 in Section 3.5.3. To further investigate the statistically significant differences in sagittal plane ankle GA between treatments, unaffected ankle ROM during gait was investigated for both AFO and WalkAide ambulation.

Unaffected ankle ROM differed between the WalkAide (range: 22.1-24.3°) and AFO (range: 22.8-24.4°) treatments, as shown in Figure 51. These differences, however, were not significant. Additionally, while statistically significant differences in ankle GA were observed with treadmill orientation, such differences were not observed in unaffected ankle ROM.

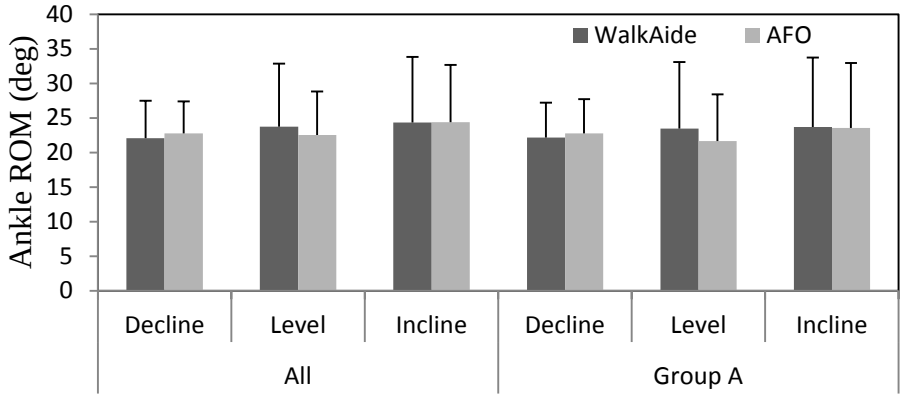

Figure 51: Mean sagittal plane unaffected limb ankle ROM for all subjects and the Group A subset contrasting treatments and treadmill orientation.
+, ++ denotes a statistically significant difference (0.05 and 0.01 levels, respectively)

4.3.2.2 Affected Ankle Range of Motion

Restricted active and passive ankle ROM has been correlated with impaired balance (goniometric measure of passive and active ROM, [56]) and energy efficiency (active ROM during gait, [57]). As such, the passive ankle ROM was measured (without AFO) before each gait session and the active ROM during gait (AFO on) of the affected versus unaffected limbs

were contrasted. Table 14 shows the subject specific passive ROM and Figure 52 displays the mean passive ROM before each treatment gait session. The passive ankle ROM was significant greater for the unaffected limb compared to the affected limb before the WalkAide gait session when examining all subjects; although the same trend was observed for Group A, it was not statistically significant. No difference in passive ROM was observed before the AFO gait session.

The ankle ROM of the affected and unaffected limbs differed on an individual subject basis (Figure 53, subject S12) and across subjects (Figure 54). The mean ankle ROM of the affected (All: 17.0- 21.0°, Group A: 15.3-20.9°) limb was less than that for the unaffected (All: 22.0-24.4°, Group A: 21.6-23.7°) limb. These differences were statistically significant for all treadmill orientations during AFO ambulation, and during inclined walking with the WalkAide (Group A subset).

Table 14: Passive ankle ROM before WalkAide and AFO gait sessions

	WalkAide		**AFO**	
	Affected	**Unaffected**	**Affected**	**Unaffected**
S2	37	40	22	30
S4	55	55	50	53
S5	43	45	42	35
S6	36	42	35	35
S7	35	40	30	30
S8	30	45	38	40
S10	35	40	37	35
S12	35	45	30	40
Mean (std)	**38° (8)**	**44° (5)**	**36°(8)**	**37°(7)**

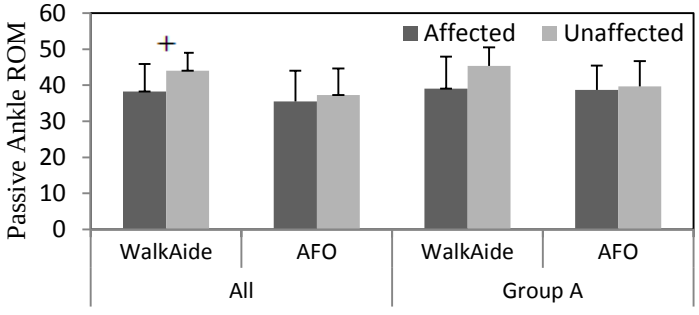

Figure 52: Mean passive ankle ROM of the affected and unaffected ankles measured before each gait session, for all subjects and Group A.

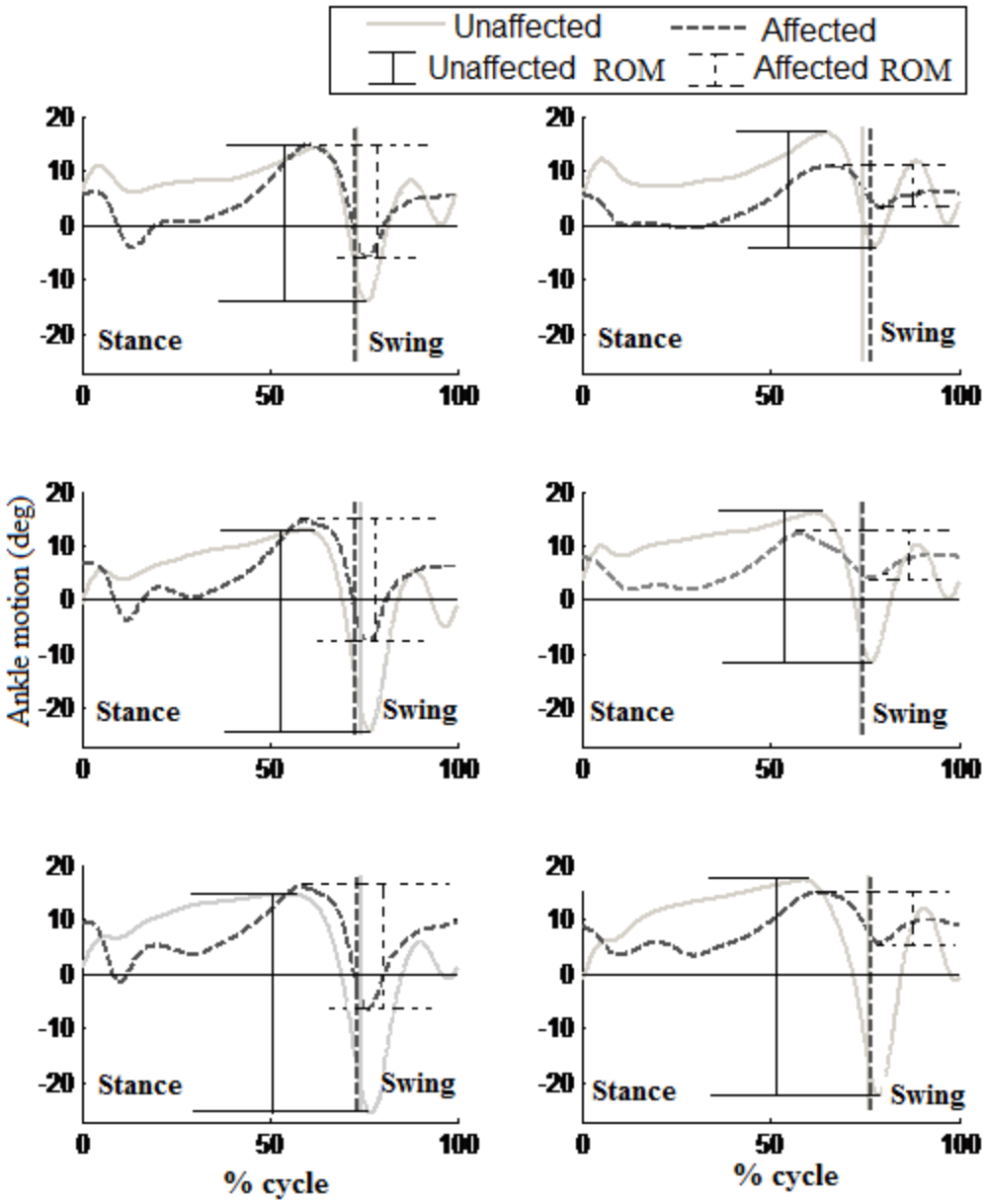

Figure 53: Representative sagittal plane ankle motion (+: dorsi, -: plantar) during WalkAide (left) and AFO (right) ambulation during declined (top), level (middle) and inclined (bottom) treadmill walking for subject S12.

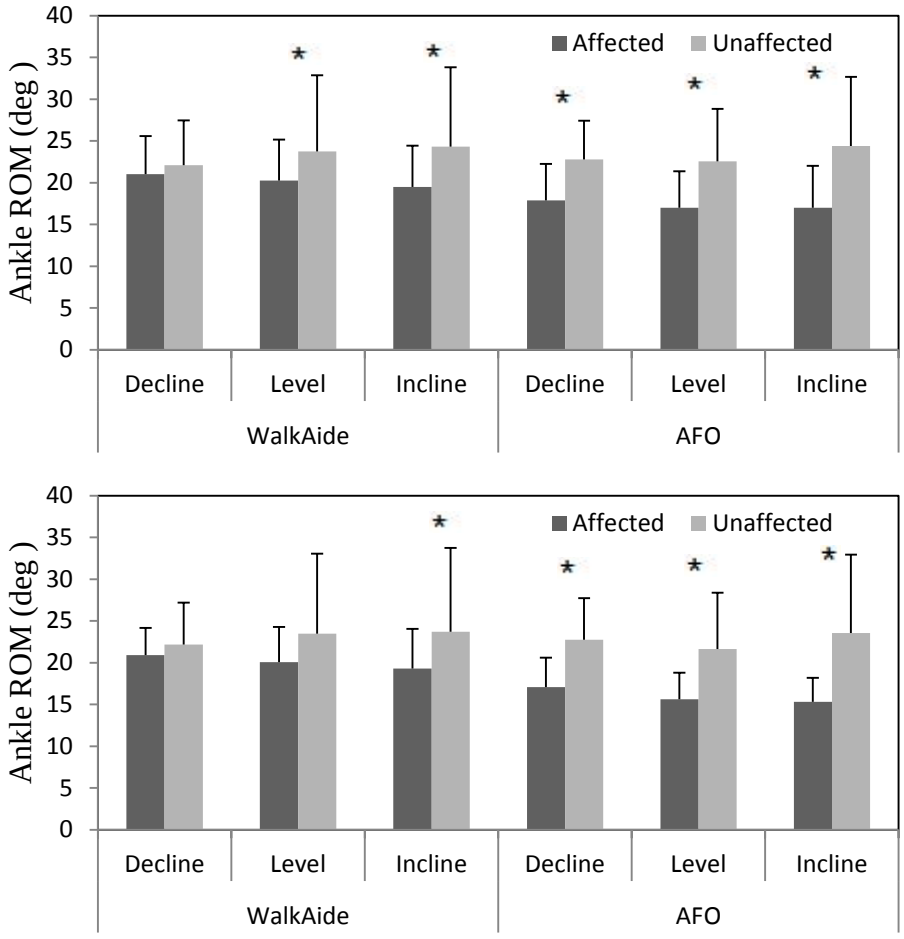

Figure 54: Mean sagittal plane ankle ROM during gait for all subjects (top) and the Group A subset (bottom), contrasting the affected versus unaffected limbs.
+, ++ denotes a statistically significant difference (0.05 and 0.01 levels, respectively)

4.3.3 Ankle Plantar Flexion and Dorsiflexion

As noted in section 4.3.1, differences in ankle GA between treatments were observed during IS (but not during MS or TS). These differences might be due to differences in the GA denominator (unaffected ankle ROM, section 4.3.2) and/or the numerator (mean and minimum ankle position). These potential differences in the GA numerator (i.e. affected-unaffected), namely the minimum (peak plantar flexion, or minimum dorsiflexion if plantar flexion not achieved) ankle motion, were investigated. Also, since a new measure for GA was used, the difference in maximum (peak dorsiflexion) ankle motion was also investigated to verify that the measure could accuracy detect both differences and the lack of difference in GA.

The difference in minimum ankle motion between the affected and unaffected limbs during IS is shown in Figure 55 for all subjects and the Group A subset. As the unaffected limb of the full subject population was plantar flexed for all treadmill orientations, the difference in minimum ankle position between the affected and unaffected limbs was positive. Positive differences were also seen during AFO use and during level and inclined walking with the WalkAide for the Group A subset. During declined WalkAide ambulation for the Group A subset, many subjects (S4, S5, S6, S7, and S8) did not achieve plantar flexion with either the unaffected or affected limbs. In fact, the minimum ankle position was more dorsiflexed for the unaffected limb during IS, resulting in negative differences in minimum ankle position. While the difference in minimum ankle motion between limbs was larger for all subjects and Group A during AFO ambulation compared with WalkAide ambulation on all treadmill orientations, no statistically significant differences between treatments were found.

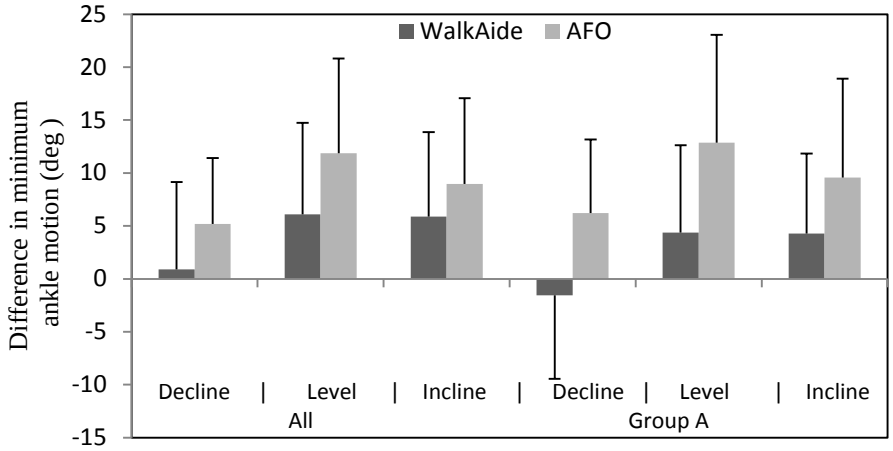

Figure 55: Difference in minimum ankle motion between the affected and unaffected limbs during IS for all subjects and Group A.
+, ++ denotes a statistically significant difference (0.05 and 0.01 levels, respectively)

The difference in affected versus unaffected parameters used to calculate GA (maximum/dorsiflexion motion during MS and TS and minimum/plantar flexion during IS) were also examined to verify the accuracy of the GA measure. Although the tibialis anterior of the affected limb was stimulated with the WalkAide to produce dorsiflexion similar to the unaffected

limb, while the AFO only provides structural support with a fixed angle, no significance difference in GA were observed. Since there was no difference in the denominator of the GA measure, there should not be a difference in the numerator of the measure. Figure 56 shows the difference in maximum ankle motion (i.e. dorsiflexion) between the affected and unaffected limbs during MS and TS for all subjects and Group A. As expected, there was no statistically significant difference in the relationship between the affected and unaffected limbs during WalkAide ambulation when compared to AFO ambulation.

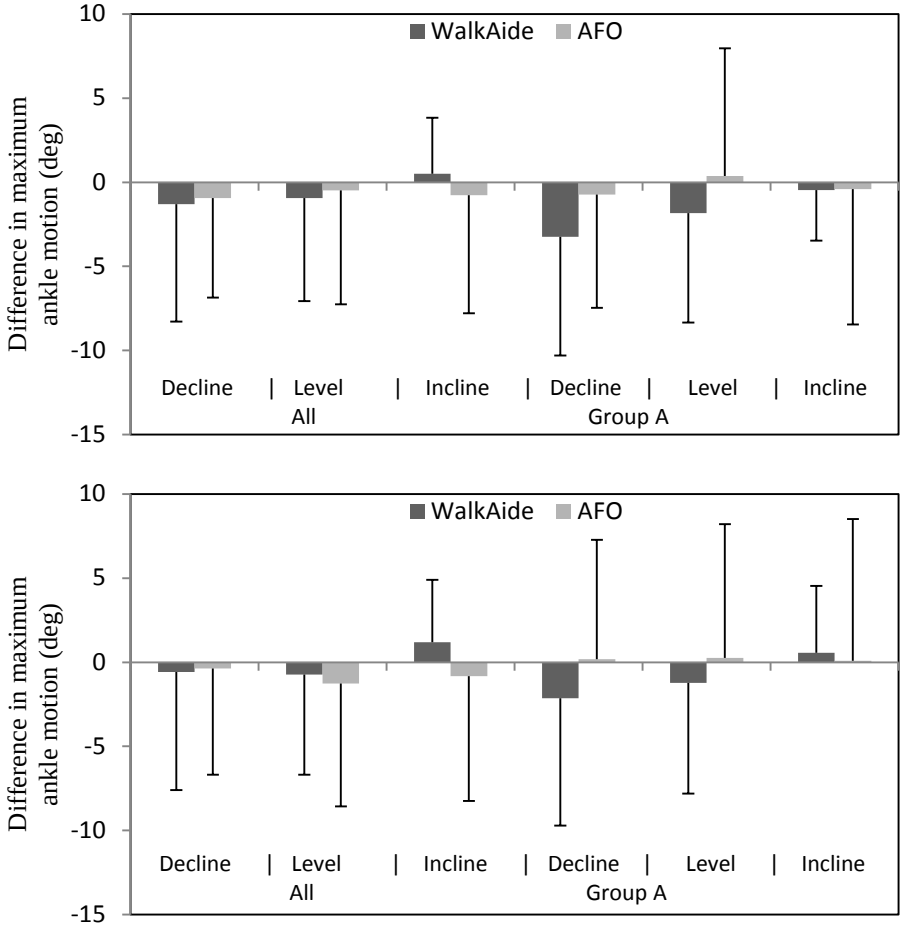

Figure 56: Difference in peak ankle dorsiflexion between the affected and unaffected limbs during midswing (top) and terminal swing (bottom) for all subjects and the Group A subset. +, ++ denotes a statistically significant difference (0.05 and 0.01 levels, respectively)

4.3.4 Ankle Angular Velocity

The ankle dorsiflexion rate or angular velocity during IS influences toe clearance and the efficiency of swing. To investigate the treatment efficacy with respect to toe/foot clearance during swing, the peak ankle angular velocity of the affected limb during IS was determined.

The maximum angular velocity of the affected limb during IS for all subjects and the Group A subset for all treadmill orientations is shown in Figure 57. The peak ankle angular velocity was consistently higher during the WalkAide versus AFO trials; these differences with treatment were statistically significant for all treadmill orientations. No differences in angular velocity with treadmill orientation were noted for either treatment during subsequent post-hoc testing.

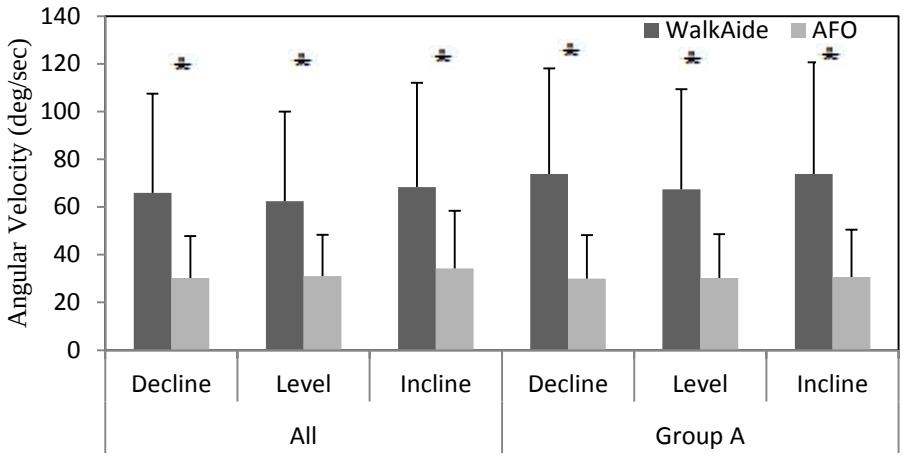

Figure 57: Peak ankle angular velocity (+: dorsi) during IS for all subjects and the Group A subset.
+, ++ denotes a statistically significant difference (0.05 and 0.01 levels, respectively)

4.4 TOE CLEARANCE

Individuals with drop foot have difficulty with foot clearance during swing, a problem that is addressed by both AFO and WalkAide treatments. To investigate the treatment efficacy of these devices (Hypothesis 1b), toe clearance of both the affected and unaffected limbs was determined for each full gait cycle for all treadmill orientations.

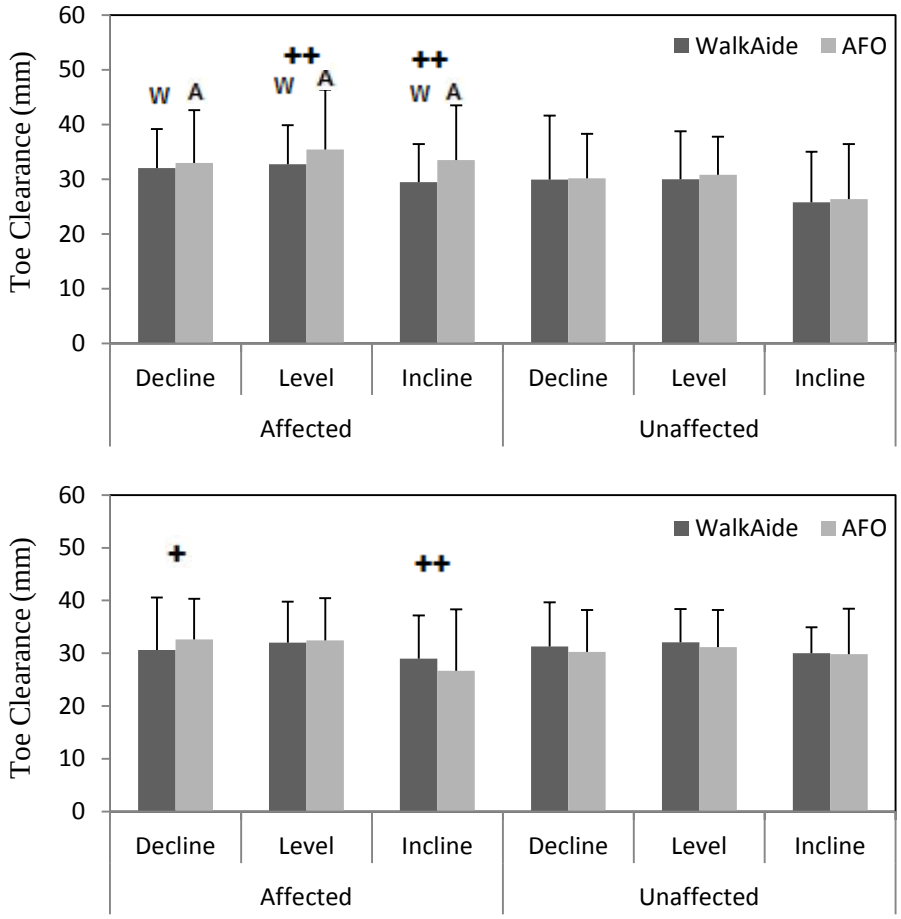

Figure 58: Toe clearance for all subjects (top) and Group A (bottom), contrasting treatments (WA vs. AFO: $^{+}$, $^{++}$ denotes a statistically significant differences at 0.05 and 0.01 levels, respectively) and limbs (unaffected vs. affected: ^{W}, ^{A} denotes a statistically significant differences at 0.05 level for WA and AFO ambulation, respectively).

Affected limb toe clearance was contrasted between treatments (Hypothesis 1b) to assess treatment efficacy. Greater toe clearance of the affected limb was observed during AFO versus WalkAide ambulation; these treatment differences were statistically significant for level and inclined walking. No statistically significant treatment differences were observed for the unaffected limb. For the Group A subset, the affected limb demonstrated greater toe clearance using the AFO during decline and level walking, although these differences were statistically significant during declined walking only. Significantly greater affected limb toe clearance was

seen with the WalkAide versus AFO during inclined walking. Again, no statistically significant differences between treatments were observed for the unaffected limb for the Group A subset.

Although increased toe clearance may indicate greater treatment efficacy or a reduced risk of falling for individuals with drop foot, greater toe clearance with the affected versus unaffected limb may require more energy during ambulation and may reflect a fear of falling. Since an optimal value for toe clearance is not known, the absolute value of the difference between the affected and unaffected limb toe clearance was also investigated.

The mean difference in toe clearance magnitude between the affected and unaffected limbs was calculated for each treadmill orientation and contrasted between treatments (see Figure 59). While the AFO treatment appeared to result in higher (although not statistically significant) mean toe clearance differences or more asymmetric toe clearance for all subjects, this trend was not apparent for the Group A population subset that excluded subjects S2 and S7. [As noted previously, the toe clearance of the affected limb was typically greater than that of the unaffected limb (e.g. negative difference in toe clearance)].

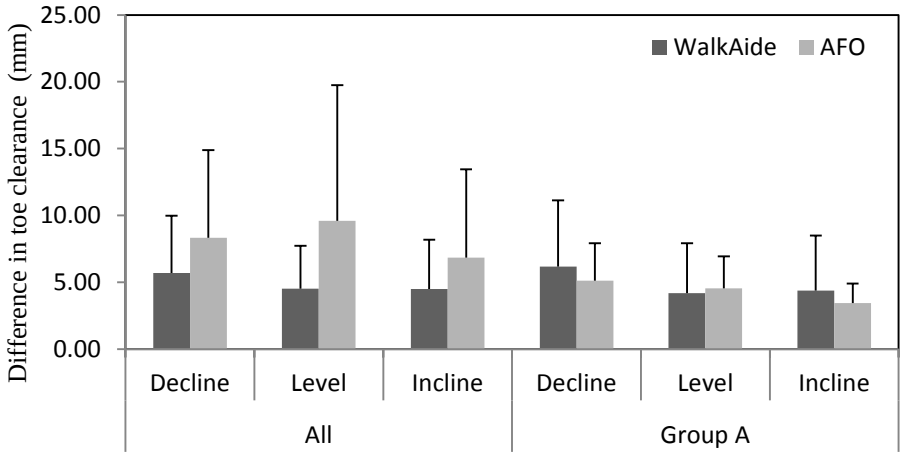

Figure 59: Mean difference in toe clearance magnitude during swing between the affected and unaffected limbs for all subjects and the Group A subset.
+, ++ denotes a statistically significant difference (0.05 and 0.01 levels, respectively)

4.5 WALKAIDE STIMULATION

The WalkAide is a neuroprosthesis designed to stimulate the ankle dorsiflexors during swing to assist with foot clearance for individuals with drop foot. In this study, these neuromuscular stimulation periods were identified using the anterior tibialis electromyogram and threshold detection techniques. The stimulation reliability (Hypothesis 2a) and initiation/termination timing (Hypothesis 2b) of the WalkAide, clinically programmed using the tilt sensor, were assessed using these data for level versus non-level walking. Estimated or theoretical heel sensor-based stimulation was determined using the WalkAide heel sensor (or F-scan sensors when WalkAide heel sensor data were not available) and heel sensor-based stimulation reliability (Hypothesis 3a) and initiation/termination timing (Hypothesis 3b) were compared with tilt sensor-based stimulation.

4.5.1 Stimulation Reliability

WalkAide stimulation reliability (StR) was determined for each subject. Ideally, each gait cycle would have one WalkAide stimulation, occurring during swing to provide the ankle dorsiflexion needed for toe clearance. Trials in which the number of gait cycles and WalkAide stimulations were equivalent correspond to a StR of one; StR values less than one reflect missed stimulations, while StR values greater than one are indicative of extraneous WalkAide stimulations.

4.5.1.1 Tilt Sensor-Based Stimulation Reliability

WalkAide tilt sensor-based StR was calculated for each subject for all treadmill orientations (see Figure 60). Four (S5, S8, S10, S12) of the eight subjects demonstrated consistent StR values of unity for all treadmill orientations. Three subjects (S2, S4, S6) exhibited missed stimulations for declined walking and extraneous stimulations for level and inclined walking.

Subject S7 demonstrated extraneous stimulations for all treadmill orientations. None of these differences in StR between treadmill orientations, however, were statistically significant.

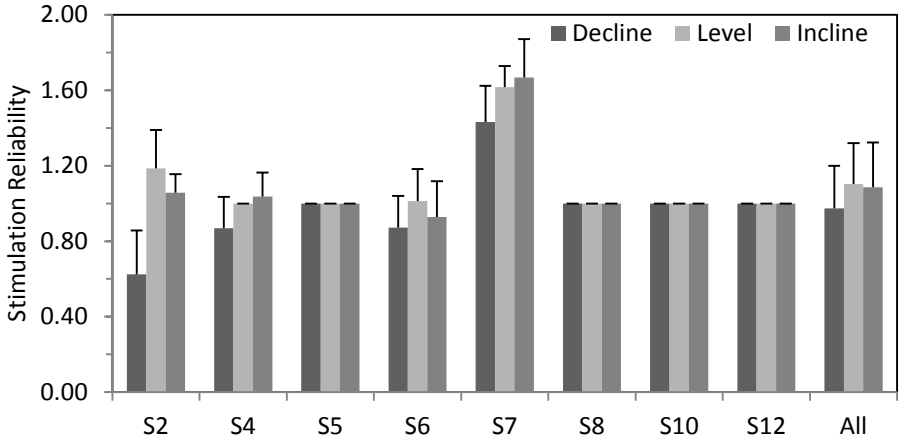

Figure 60: WalkAide tilt sensor-based StR for all subjects for all treadmill orientations.
+, ++ denotes a statistically significant difference (0.05 and 0.01 levels, respectively)

4.5.1.2 Heel-Sensor Based Stimulation Reliability

The WalkAide is typically programmed using the tilt-sensor to control stimulation, although the clinician also has the option to control stimulation using heel sensor data. Programming for heel sensor-based stimulation requires setting the stimulation heel loading thresholds for stance (loaded heel sensor) and swing (unloaded heel sensor).

WalkAide heel sensor data were acquired and threshold detection (in conjunction with tilt sensor stimulation programming parameters) was used to identify the respective stance and swing periods, defining the theoretical on (swing) and off (stance) stimulation times for each WalkAide walking trial. These WalkAide data were synchronized to the kinematic data as described in Section 3.6.2 such that the theoretical StR of heel sensor-based stimulation might also be estimated, as shown in Figure 61. [WalkAide heel sensor data were not available for subjects S7 and S8; heel sensor-based StR for these subjects were based on the F-scan insoles].

The theoretical heel sensor-based StR (Figure 61) can be contrasted with the clinically programmed tilt sensor-based stimulation (Figure 60) so as to test Hypothesis 3a. Two subjects

(S4, S5) exhibited extraneous (StR > 1) and/or missed (StR < 1) theoretical stimulations for non-level walking (both declined and inclined). Subject S4 had a combination of extraneous and missed stimulations for both declined and inclined ambulation, while S5 exhibited missed stimulations during inclined ambulation. These results differed from that observed using the clinically programmed tilt-sensor for which missed stimulations were observed for subject S4 during declined walking and extraneous stimulations were observed during inclined walking. Similarly, the extraneous theoretical stimulations with the heel sensor for subject S5 were not observed during tilt sensor-based stimulation (StR = 1). The remaining subjects showed a theoretical heel sensor-based StR of one for all treadmill orientations. The results were consistent with the tilt sensor-based stimulations for subjects S8, S10 and S12. However, improved StR was observed for subjects S2, S6 and S7 with the theoretical heel sensor-based versus clinical tilt sensor-based stimulation, although these sensor differences were not statistically significant. Combined theoretical heel sensor-based StR were approximately unity for all treadmill orientations; no statistically significant differences between sensors (tilt vs. heel) for any treadmill orientation were observed (Hypothesis 3a).

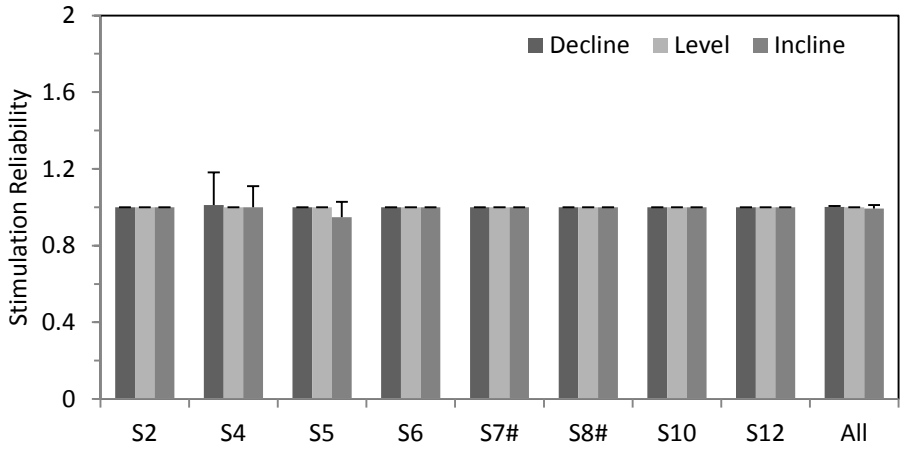

Figure 61: WalkAide heel sensor-based StR for all subjects for all treadmill orientations.
+, ++ denotes a statistically significant difference (0.05 and 0.01 levels, respectively)
 # F-scan insole used to estimate heel loading

4.5.2 Stimulation Initiation and Termination Timing

4.5.2.1 Tilt Sensor-Based Stimulation Timing

The neuromuscular stimulation periods based on EMG data were used to identify the stimulation initiation (StI) and stimulation termination (StT) times in percent gait cycle. Each StI value was then compared to the initiation of swing phase for that respective gait cycle such that the StI timing could also be expressed relative to swing (e.g. Did stimulation occur early, prior to swing, or late, after initial swing?). Similarly each StT value was expressed relative to the beginning of stance (e.g. Did stimulation cease after swing, during the subsequent stance phase?). Sample StI and StT times in both percent gait cycle and relative swing and stance periods are illustrated in Figure 62.

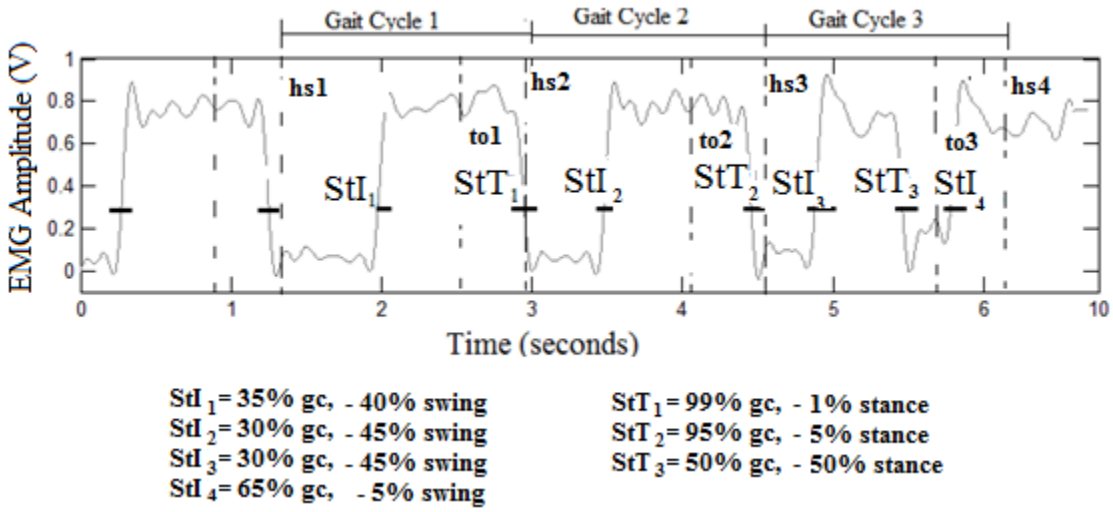

Figure 62: Sample StI and StT as defined in terms of both gait cycle (gc) and swing and stance, respectively.

While StI and StT can be easily calculated, outliers often existed due to extraneous stimulations related to anomalies in the tilt sensor data (e.g. movement artifact due to tremors or limb circumduction and/or tilt sensor alignment errors). As shown in Figure 63, omitting those gait cycles which included extraneous stimulations reduced much of the variability in the tilt sensor-based stimulation timing data. These outliers and excessive variability were particularly

prominent for subjects S2 and S7 for whom standard deviations greater than 15% gait cycle were noted for all treadmill orientations prior to outlier elimination. Even after eliminating these outliers, however, subject S2 exhibited standard deviations in stimulation timing greater than 10% gait cycle. For subject S7, omission of gait cycles with extraneous stimulations resulted in the elimination of nearly half of all gait cycles. [These extraneous stimulations and excessive variability warranted re-analysis of several parameters for the Group A subset that excluded these two subjects from the sample population.]

For most subjects (S2, S4, S7, S8, S10, and S12), StI occurred at approximately -20% swing, indicating that SI occurred during PS. In contrast, for subject S5, StI occurred during early swing (~5-10% swing, during IS); for subject S6, StI was seemingly early, occurring at approximately -30 to -40% swing (during mid or terminal stance). In general, StI was delayed as the treadmill orientation progressed from declined to inclined, occurring later in stance, closer to swing phase itself. Post-hoc testing revealed statistically significant differences in StI between declined/inclined treadmill orientations, both before and after eliminating outliers. StI occurred earlier in the gait cycle during declined walking (-15% swing) when compared with level (-10% swing); StI occurred later in the gait cycle during level (-10% swing) when compared to inclined (-7% swing) walking. Although differences were observed between level and inclined walking (Hypothesis 2b), these differences between level and non-level walking trials were not statistically significant.

For most subjects (S2, S4, S6, S7, S8, and S10), StT occurred during the first 20% of the subsequent gait cycle (i.e. 120% gait cycle), indicating that StT typically occurred during loading response of the following cycle. For subjects S6 and S7, StT occurred as late as 130% gait cycle for inclined walking. StT was also quite late for subjects S5 and S12 (130 to 150% gait cycle). For these subjects and the inclined walking trials of subjects S6 and S7, StT most likely occurred during the subsequent mid or terminal stance. Although differences in StT were observed between level and non-level walking (Hypothesis 2b) for some subjects (S4, S5, S6, and S7), no

statistically significant differences in StT timing were observed between level and non-level ambulation.

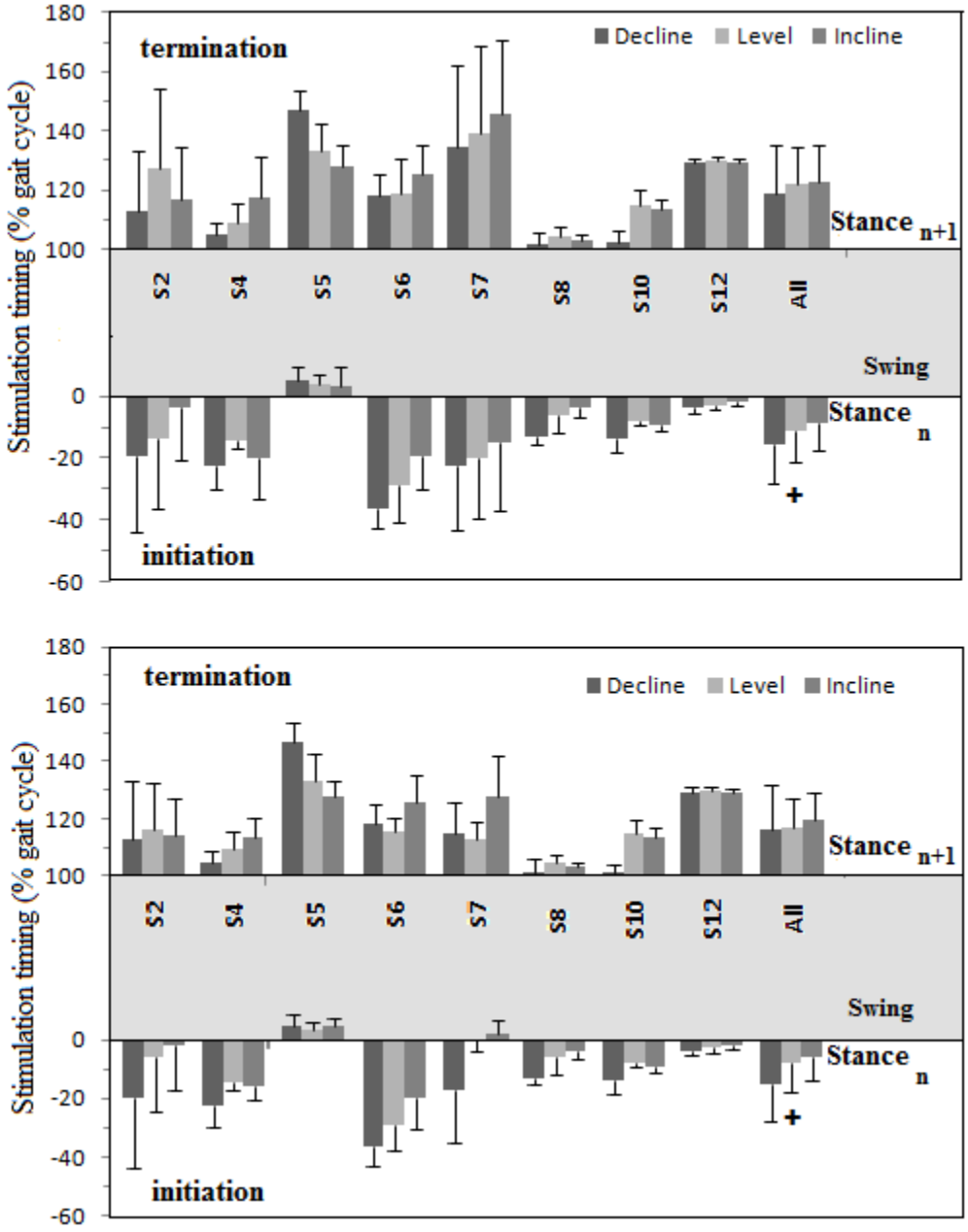

Figure 63: WalkAide tilt sensor-based stimulation timing before (top) and after (bottom) elimination of extraneous stimulations or outlier gait cycles.

Note that StI occurs prior to swing, during the previous stance phase (negative %cycle); StT occurs after swing, during the subsequent gait cycle (> 100% cycle).

4.5.2.2 Heel Sensor-Based Stimulation Timing

StI and StT were also estimated for theoretical heel sensor-based stimulation for all subjects. StI and StT were estimated based on WalkAide heel sensor data for the six subjects for whom WalkAide heel sensor data were available; StI and StT were estimated using F-scan data for subjects S7 and S8. As for the tilt sensor-based stimulation, these StI and StT data were calculated both before and after eliminating outliers (e.g. gait cycles with extraneous stimulations).

The results of the stimulation timing based on the WalkAide heel sensor (and F-scan insoles for subjects S7 and S8) before and after eliminating outliers can be seen in Figure 64. Outliers due to extraneous theoretical stimulations in this case may be attributed to heel pressure anomalies due to potential movement of the heel sensor or foot within the shoe, resulting in extra peaks in heel sensor loading. Although elimination of outliers did not reduce the intra-subject variability as dramatically as for tilt sensor-based stimulation, such elimination influenced the timing results for subjects S4 and S10. StI again occurred prior to swing (-20 to -25% swing before and after omitting outliers). Although StI occurred at approximately -20% swing for both tilt and heel sensor-based stimulations theoretical heel sensor-based StI demonstrated more consistent timing among subjects during non-level ambulation (Hypothesis 3b), with StI occurring outside this range for only one subject (S5). As for tilt sensor programming, no statistically significant differences in heel sensor-based StI were observed with treadmill orientation; additionally, no statistically significant differences in StI between sensors (tilt vs. heel) for any non-level treadmill orientations were observed.

As with tilt sensor-based stimulation, StT occurred during the subsequent stance period. These StT times, however, were slightly earlier (approximately 0-10% stance or 100-110% gait cycle) than that observed during tilt sensor-based (approximately +20% stance or 120% gait cycle) stimulation. Only one subject (S8) demonstrated theoretical heel sensor-based StT during

swing (95% gait cycle, -5% stance), prior to the subsequent gait cycle. Again, no statistically significant differences in heel sensor-based StT were observed between treadmill orientations and no significant differences StI were observed between sensors for non-level surfaces.

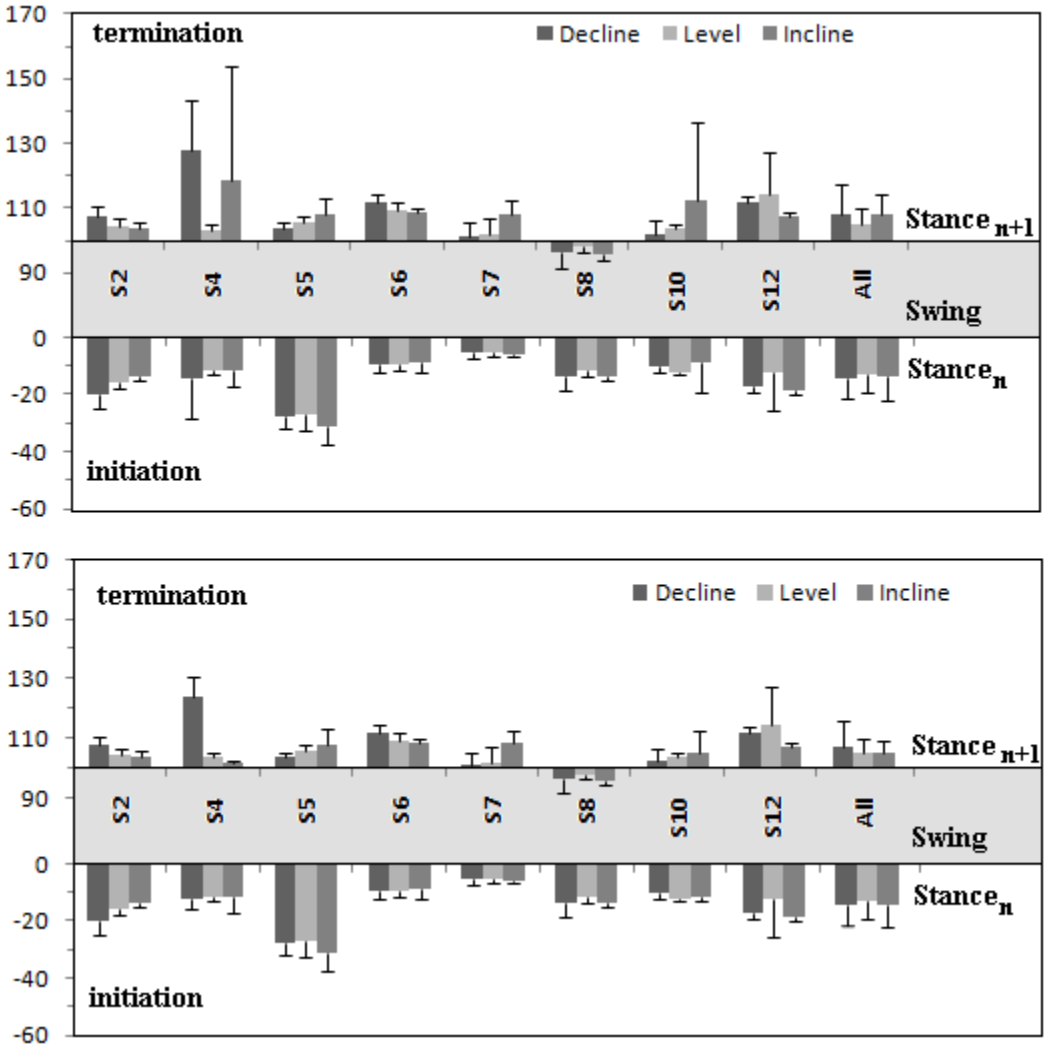

Figure 64: WalkAide heel sensor-based theoretical stimulation timing before (top) and after (bottom) elimination of extraneous stimulations or outlier gait cycles.
Note that StI occurs prior to swing, during the previous stance phase (negative %cycle); StT occurs after swing, during the subsequent gait cycle (> 100% cycle).
#: F-scan insole used to estimate heel loading.

While extra stimulations occurred with both tilt sensor and (theoretical) heel sensor-based stimulations, fewer extra stimulations were observed for the theoretical heel sensor-based stimulation (Hypothesis 3a), particularly for subjects S2 and S7 for whom nearly 50% of the gait cycles resulted in extra stimulation with the tilt sensor and less than 10% of the gait cycles

resulted in extra stimulations for theoretical heel sensor-based stimulation. Eliminating all gait cycles with extraneous stimulations resulted in less variability (smaller standard deviations) in StI and StT for both tilt and heel sensor-based stimulation. As shown in Figure 65, the StI and StT timing variability was consistently less for the theoretical heel sensor-based stimulation than for the clinical tilt sensor-based stimulation. However, these differences were statistically significant for StT during level walking only; no statistically significant differences in stimulation timing variability were observed during ambulation on non-level surfaces (Hypothesis 3b).

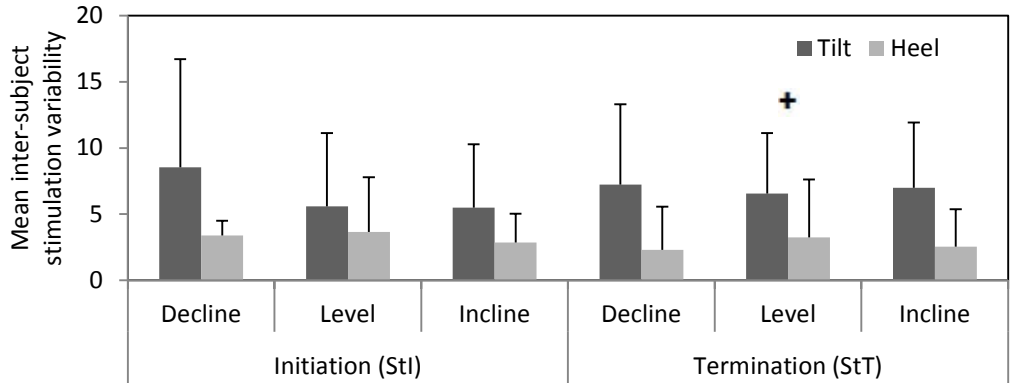

Figure 65: Tilt sensor- and theoretical heel sensor-based WalkAide stimulation timing [StI (left) and StT (right)] variability.
+, ++ denotes a statistically significant difference (0.05 and 0.01 levels, respectively)

4.5.3 Study Groups

Inconsistent WalkAide stimulation and high variability in StI and StT timing affects gait kinematics, influencing measures such as gait symmetry and toe clearance. StR values less than one, which indicate missed stimulations, result in minimal or no ankle dorsiflexion affecting toe clearance and gait symmetry measures. StR values greater than one (extraneous stimulations) result in anterior tibialis activation at various times during the gait cycle, providing potential ankle dorsiflexion that does not address foot clearance during swing. Due to the poor StR and high variability in StI and StT timing observed for subjects S2 and S7, the temporal and kinematic data for these two subjects were omitted in the aforementioned Group A subset analyses.

4.6 F-SCAN PLANTAR PRESSURE DATA

F-scan plantar pressure data were collected during WalkAide gait sessions and compared to the WalkAide heel sensor data for all subjects except S7 and S8 for whom WalkAide heel sensor data were not available. This comparison was used to assess the validity of using F-scan heel box loading data as a substitute for the WalkAide heel sensor in the determination of theoretical heel sensor-based stimulation timing.

Both the mean and maximum heel loading magnitudes were contrasted, as shown in Appendix D (Figure D1). As WalkAide heel sensor magnitude varied inversely with load (Figure 32), the WalkAide loading data were converted by subtracting the maximum unloaded value or bias. While the maximum heel loading magnitudes of the F-scan and heel sensor data were not correlated ($R^2 = 0.0004$), the mean heel loading magnitudes of these two sensors were positively, although modestly, correlated ($R^2 = 0.19$).

The mean heel loading (~stance) and unloading (~swing) duration for both the WalkAide and F-scan sensors was also contrasted. As shown in Figure D2, the heel load duration data for the F-scan versus WalkAide heel sensors were positively correlated, as illustrated by the relatively high correlation coefficient (0.91). While the heel unloading duration data for these two sensors were also positively correlated (Figure D3), the correlation coefficient for the unloading duration (0.76) was less than that for the loading duration (0.91). Based on these comparisons and positive correlations, the use of the F-scan sensor data as a substitute for the missing WalkAide heel sensor data (subjects S2 and S7) can be justified, particularly with respect to timing.

4.7 POWER ANALYSIS

As the post-stroke population in this study was relatively non-homogenous, the various measured parameters were not normally distributed and non-parametric tests (Wilcoxon rank sum, Friedman) were necessary for statistical analysis. Non-parametric tests do not assume normal distribution, are more robust, and can accommodate the variability in the post-stroke population; however, these non-parametric tests are less powerful than parametric tests. As such, larger sample sizes are often required [55]. To determine the number of subjects necessary to achieve statistical differences ($p>0.05$) with sufficient power ($P>0.80$), as well as to determine the power of the various parameters of the current study, *a priori* and post hoc power analyses were conducted.

Three *a priori* power analyses were conducted to determine the number of subjects necessary to detect statistically significant differences, assuming small (0.2), medium (0.5) and large (0.8) effect sizes [55]. Sample sizes were determined for both individual, non-paired parameter analysis (e.g. GA and toe clearance of the affected limb for the WalkAide versus AFO treatments) and paired, repeated measures analysis (e.g. 30 gait cycles per subject for various temporal parameters), as shown in Table 15. For individual, non-paired measures, 15- 208 subjects are necessary to detect large and small effect sizes, respectively. For paired, repeated measures analysis, 1-7 subjects are needed to detect large and small effect sizes, respectively.

Table 15: A *priori* power analysis to determine sample size
***assumes 30 steps per subject**

Effect	Effect size, d [55]	# subjects	# of subjects for repeated measures analysis*
Small	0.2	208	7
Medium	0.5	35	2
Large	0.8	15	1

A priori and post hoc power analyses based on the actual effect sizes of current study were also conducted both to determine the number of subjects necessary to detect statistical significance and the actual statistical power. The results of these power analyses are summarized

in Table 16 for the various temporal parameters. For the stance and total DLS durations on decline and level orientations, and the initial DLS on decline and incline orientations, a prior analysis revealed that eight subjects are sufficient to observe statistically significant differences ($p>0.05$) between treatments of the affected limb temporal parameters with sufficient power ($P>0.80$). Inclined swing duration and level initial DLS indicate that nine subjects are needed to detect statistically significant differences with sufficient power. Although statistically significant differences were observed for the affected limb second DLS period between treatments for declined and inclined walking, the statistical power for these parameters was low ($P<0.50$). Power analysis of the swing and second DLS duration during level walking indicate that more than 1000 subjects may be needed to detect statistically significant differences.

Table 16: *A priori* and post hoc power analyses to investigate treatment effects of the affected limb temporal parameters

Treatment effects: WalkAide versus AFO									
	Effect Size, d			Power, P (post hoc)			# of subjects (a *priori*)		
	Decline	Level	Incline	Decline	Level	Incline	Decline	Level	Incline
Stance Duration	0.22	0.19	0.03	0.84	0.71	0.07	6	8	260
Swing Duration	0.16	0.00	0.18	0.61	0.05	0.67	10	>1000	9
Initial DLS	0.40	0.18	0.36	0.99	0.67	0.99	2	9	2
Second DLS	0.07	0.00	0.15	0.16	0.05	0.49	58	>1000	13
Total DLS	0.26	0.27	0.12	0.94	0.95	0.37	4	4	18

The *a priori* and post hoc power analyses were also conducted for the non-repeated temporal gait symmetry measures, as shown in Table 17. Based on the SR data collected in this study, more than 100 subjects are required to observe a significant difference in stance and/or swing phase SR on non-level surfaces; more than 1,000 subjects are required to detect statistically significant differences in these parameters for level walking. Similarly, more than 100 subjects are required to detect statistically significant differences in the temporal gait asymmetry index between treatments for both level and non-level walking. These large populations are necessary to observe statistically significant differences in temporal symmetry measures due to the small to moderate effect sizes ($0.02 \leq d \leq 0.33$) and low power (0.06-0.21).

Table 17: *A priori* and post hoc power analyses to investigate treatment effects of the temporal symmetry parameters

Treatment effects: WalkAide versus AFO									
	Effect size, d			Power, P (post hoc)			# of subjects (a priori)		
	Decline	Level	Incline	Decline	Level	Incline	Decline	Level	Incline
SR stance	0.16	0.02	0.13	0.11	0.06	0.09	446	>1000	682
SR swing	0.32	0.10	0.13	0.20	0.08	0.09	113	>1000	658
Gait Asymmetry Index	0.33	0.17	0.24	0.21	0.11	0.15	105	391	199

The *a priori* and post hoc power analyses were also conducted for the non-repeated GA measures of the knee and ankle, as shown in Table 18. For all the functional phases of swing, more than 100 subjects are required to determine statistically significant differences in knee GA between treatments. Fewer subjects (N≥10) are needed to detect treatment differences in ankle GA.

Table 18: *A priori* and post hoc power analyses to investigate treatment effects in GA during IS, MS and TS

Treatment effects: WalkAide versus AFO									
Knee GA	Effect Size, d			Power, P (post hoc)			# of subjects (a *priori*)		
	Decline	Level	Incline	Decline	Level	Incline	Decline	Level	Incline
IS max	0.22	0.26	0.25	0.07	0.08	0.07	331	237	269
MS avg	0.03	0.13	0.14	0.05	0.06	0.06	16742	995	899
TS avg	0.36	0.20	0.23	0.10	0.07	0.07	129	410	312
Ankle GA	Effect Size, d			Power, P (post hoc)			# of subjects (a *priori*)		
	Decline	Level	Incline	Decline	Level	Incline	Decline	Level	Incline
IS avg	0.16	1.32	1.26	0.06	0.67	0.63	665	11	12
IS min	0.16	1.39	1.25	0.06	0.71	0.62	665	10	12
MS avg	0.17	0.57	0.40	0.06	0.18	0.11	571	51	105
TS avg	0.19	0.41	0.49	0.06	0.11	0.15	481	101	69

Finally, the results for the *a priori* and post hoc power analyses for toe clearance treatment differences are summarized in Table 19. A priori tests reveal that only 2-4 subjects are necessary to observe statistically significant differences in the affected limb toe clearance for

level and inclined walking. Due to the small to moderate effect size ($0.02 \leq d \leq 0.11$) and low power (0.05-0.29), substantially larger subject populations ($N \geq 24$) are required to observe statistically significant treatment differences in the remaining (affected, level, or unaffected limb, all) toe clearance results.

Table 19: *A priori* and post hoc power analyses to investigate treatment differences in affected and unaffected toe clearance, and limb differences in toe clearance

Treatment effects: WalkAide versus AFO									
	Effect size, d			Power, P (post hoc)			# of subjects (a *priori*)		
Toe Clearance	Decline	Level	Incline	Decline	Level	Incline	Decline	Level	Incline
Affected Limb	0.11	0.28	0.46	0.29	0.96	0.99	24	4	2
Unaffected Limb	0.02	0.10	0.06	0.06	0.25	0.12	511	30	74
Difference in Toe Clearance	Effect size, d			Power, P (post hoc)			# of subjects (a *priori*)		
Between Limbs	0.47	0.68	0.44	0.14	0.23	0.13	75	38	85

4.8 SUMMARY

Temporal parameters, gait asymmetry, joint kinematics and WalkAide stimulation reliability and timing were examined to test the respective research hypotheses. Statistically significant differences between treatments in the stance and swing durations, as well as initial and second DLS periods were observed, demonstrating potentially improved balance with the AFO, and improved ankle function with the WalkAide. However, contrary to Hypothesis 1a, no statistically significant differences in temporal symmetry (i.e. SR and SI) were observed between treatments. Statistically significant differences between the mean and minimum ankle GA during IS between treatments were observed, with AFO ambulation demonstrating greater ankle GA during IS (supporting Hypothesis 1a), indicating greater kinematic symmetry of the ankle with the WalkAide. No statistically significant differences in unaffected ankle ROM (GA denominator) between treatments were observed; the differences in ankle GA may therefore be attributed to differences in minimum ankle motion between the limbs during swing (GA numerator).

Significantly greater affected limb toe clearance was found during AFO versus WalkAide ambulation (Hypothesis 1b). However, greater affected vs. unaffected limb toe clearance was observed with the AFO compared to the WalkAide, possibly indicating that the increased affected limb toe clearance may contribute to increased energy expenditure during AFO ambulation rather than increased treatment efficacy (Hypothesis 1b).

WalkAide stimulation reliability and timing were investigated for the clinically programmed tilt sensor. Tilt sensor-based stimulation reliability (Hypothesis 2a) was not affected by non-level walking. Stimulation initiation timing changed during level versus non-level walking (Hypothesis 2b); StI occurred earlier (closer to IS) as the treadmill progressed from declined to inclined orientation. In contrast, stimulation termination timing did not vary between level and non-level walking trials (Hypothesis 2b).

Finally, theoretical WalkAide heel sensor stimulation reliability and timing were compared with the clinically programmed tilt sensor stimulation. Theoretical heel sensor stimulation resulted in more consistent, optimal (unity) StR values (six of eight subjects for all treadmill orientations) than for tilt sensor-based stimulation (four of eight study subjects for all treadmill orientations), Hypothesis 3a. Stimulation timing, both StI and StT, during non-level walking was unaffected by the WalkAide sensors (tilt vs. heel, Hypothesis 3b).

These results and their clinical relevance will be discussed further in the Discussion chapter.

CHAPTER 5: DISCUSSION

In 1961 Liberson and colleagues first proposed a neuroprosthesis to treat foot drop [33]. Currently, there are three FDA approved neuroprostheses on the market [34]. Despite numerous studies demonstrating that the use of a neuroprosthesis increases walking speed and decreases energy cost, insurance reimbursement for neuroprostheses is often denied. Reimbursement denial is usually attributed to the lack of studies documenting conclusive evidence of the benefits of a neuroprosthesis over conventional therapies such as the AFO; only two studies have directly compared neuroprosthesis function to that of an AFO [39,40] (Section 2.5). In addition, while walking speed and energy cost are clinically relevant, these parameters provide limited insight into the control of walking and the impact of neuroprostheses on functional rehabilitation goals such as weight bearing symmetry, improved balance and decreased risk of falling. Therefore, research comparing neuroprosthesis and AFO function using more specific gait parameters may make neuroprostheses more readily available to stroke patients.

In addition to these limitations in functional measures used to date, previous studies investigated level walking only, ignoring non-level walking surfaces routinely encountered during household and community ambulation. Significant changes in joint kinematics occur during declined and inclined ambulation ([41], Section 2.6.1). Since WalkAide stimulation is based on the angle of the tibia, the reliability and timing of WalkAide tilt sensor-based stimulation may change during non-level walking, affecting toe clearance and causing an increased risk of falling.

Due to limitations in prior research involving neuroprostheses to treat foot drop in post stroke individuals, the goal of this research study was to conduct more thorough analyses of post-stroke ambulation, contrasting AFO and neuroprosthesis treatments during both level and non-level walking. The specific research hypotheses were:

1) Use of a neuroprosthesis for post-stroke individuals with drop foot will improve a) temporal and kinematic gait symmetry and b) toe clearance on both level and non-level surfaces when compared to an AFO,
2) Non-level walking will affect the WalkAide tilt sensor-based stimulation reliability and timing, resulting in a) missed or extraneous stimulations during non-level ambulation and b) changes in stimulation initiation/termination timing during ambulation on non-level surfaces, and
3) The use of the WalkAide heel sensor for stimulation control will result in a) improved stimulation reliability and b) more consistent stimulation timing with respect to tilt sensor based control during non-level ambulation.

The results of this study and the testing of the aforementioned hypotheses will be discussed in this chapter. Temporal parameters, gait symmetry, joint kinematics, toe clearance, WalkAide stimulation reliability and timing during ambulation on level and non-level surfaces, power analyses results, study limitations, and clinical implications will be discussed.

5.1 TEMPORAL PARAMETERS

Temporal parameters are the most common clinical measures used to assess the function of post-stroke individuals. Although these measures are limited in their ability to describe gait pathologies and compensatory mechanisms [39, 42], temporal and spatial parameters are useful in rehabilitation planning due their ease of measure, reproducibility, and correlation to overall functional status [22]. Walking speed is the most common temporospatial parameter measured when assessing stroke subjects' function; it has been used extensively to evaluate neuroprosthesis function. However, studies have shown that comprehensive gait evaluation should include descriptions of stance and swing phase duration and proportion, as well as measures of gait asymmetry [17, 19, and 22].

5.1.1 Cadence and Gait Cycle Duration

The mean cadence, a measure indirectly related to walking speed, was approximately the same for the affected and unaffected limbs during both WalkAide and AFO ambulation, as seen in Figure 38. Although no statistically significant differences in cadence were observed, subject cadence was generally faster with the AFO versus WalkAide for both limbs. Increased cadence with AFO use could indicate that the subjects either increased their walking speed or took shorter steps. The specific mechanism for the increased cadence could not be determined, however, as stride parameters are not acquired during treadmill ambulation.

The observed cadences (61 to 63 steps/min) for these post-stroke subjects were substantially lower than that of healthy, age-matched individuals (101.0 steps/min) during level over ground self-selected walking [58]. The cadence of these post-stroke subjects was more comparable to that of other post-stroke subjects, as summarized in Table 20. The cadence values reported in this study during treadmill ambulation without an assistive device (but with handrail support) were nearly 20% less than that reported in other post-stroke studies of over ground walking with an assistive device. The slower cadences of the current study (~63 steps/min) may be largely attributed to the elevated treadmill and moving belt, as well as increased subject impairment in comparison to post-stroke subjects who did not require assistive devices for ambulation (~89 steps/min, [45]). However, the post-stroke subjects in the current study also exhibited slower cadences than reported in previous studies examining neuroprostheses (86.1 steps/min, [60]) suggesting that the current study's subjects may be moderate, not high functioning, individuals – as confirmed by their lower extremity Fugl-Meyer scores (range: 21-28 and mean: 24.4; severe 0-19, moderate 20-28, mild >29 [61]).

The mean gait cycle duration, a parameter which varies inversely with cadence, was also compared between treatments, as shown in Figure 38. During inclined walking, the gait cycle duration was typically prolonged for AFO ambulation. In contrast, a prolonged gait cycle

duration was observed with WalkAide use during declined and level walking. However, these treatment differences in gait cycle duration were not statistically significant. Similar to cadence, gait cycle duration is dependent on walking speed. Increases in both cadence and gait cycle duration reflect increased walking speed (more, longer steps). Increased walking speed with AFO use contradicts previous studies comparing a neuroprosthesis to an AFO. However, the increased walking speed in concert with greater GA with the AFO versus WalkAide suggests that increased walking speed was not linked to increased kinematic gait symmetry and therefore do not reflect improved gait function.

The gait cycle durations for these post-stroke subjects (1.81 to 1.94 sec) were much less than that reported for healthy, age-matched individuals (1.10 sec) during self-selected, level, over ground walking [58]. As seen by Table 20 in which the gait cycle durations measured during level treadmill ambulation were contrasted with previous investigations involving post-stroke subjects, the gait cycle durations measured in the current study (~1.9 sec) were comparable to that reported by Roth et al. (2.2 sec) [22], and nearly 25-30% greater than that reported by others [39, 42, 45]. This again suggests that the current study's subjects may be moderate, not high functioning, post-stroke individuals.

As alluded to above, the slower cadence and prolonged gait cycle duration in the current study may also be attributed, at least in part, to treadmill versus over ground walking. Level treadmill ambulation of post-stroke individuals with training and/or prior treadmill experience typically results in faster cadences than during over ground walking [45]. While three of the current study's subjects had treadmill experience, five subjects had no prior treadmill walking experience since their stroke. The slower cadence demonstrated by the subjects in the current study may have been due to tentativeness and lack of confidence due to the moving belt, or heightened insecurity as the treadmill was elevated on a ramp – despite the fact that subjects were encouraged to grasp the treadmill handrails. Additionally, subjects were not blinded to treadmill speed and nearly all subjects selected the same speed for all treadmill orientations (S2, S4, S5,

S7, S8, S10) and/or for both treatments (S4). Qualitative observation of the current subjects' gait indicated that all but one subject (S7) walked faster over ground than on the treadmill.

As expected for this non-homogeneous post-stroke population, substantial variability in cadence and gait cycle duration (nearly 20% for the WalkAide, 10% for AFO use) was observed. Large variability in cadence (approximately 14% [45], 28% [42], and 26% [18]) and gait cycle duration (approximately 11-14% [39], 41% [22] and 10% [60]) has been reported in previous studies investigating temporal parameters of post-stroke individuals. This variability affects the statistical power, making it more difficult to detect significant differences between treatments (or between limbs or treadmill orientations). No statistically significant differences in cadence or gait cycle duration were found between treatments (or between limbs or treadmill orientations) across all subjects and for the Group A subset. Differences in these parameters may still, however, exist although more subjects are needed to effectively contrast the AFO/WalkAide treatments, affected/unaffected limbs, and treadmill orientations (see Section 4.7).

Table 20: Comparison of cadence and gait cycle duration [mean and (s.d.)] during level walking for various post-stroke investigations; DNS= does not specify

Reference	N	Treatment	Assistive Device	Surface	Cadence (steps/min)	Gait cycle duration (sec)
Current Study	8	WalkAide (4 wks)	none	Treadmill	62.6 (12.6)	1.91 (.38)
		AFO	none	Treadmill	63.5 (7.7)	1.88 (.24)
Brouwer, 2009 [45]	10	none	none	Over ground	87.0 (11.4)	-
				Treadmill	90.8 (13.4)	-
Gok, 2003 [59]	12	AFO, 100%	cane, 100%	Over ground	65	-
Kim, 2003 [42]	20	none	cane, ~25%	Over ground	73.9 (20.7)	-
Ring, 2009 [39]	15	AFO	DNS	Over ground	-	1.48 (0.21)
		Bioness (4 wks)		Over ground	-	1.47 (0.18)
		Bioness (8 wks)		Over ground	-	1.41 (0.16)
Roth, 1997 [22]	25	DNS	cane or walker	Over ground	54.4 (2.4)	2.22 (.92)
von Schroeder, 1995 [18]	49	AFO, ~15%	cane or walker, 25%	Over ground	84.8 (22.4)	1.5 (.06)
Voigt, 2000 [60]	8	KDC 2000A, (similar to ODFS)	DNS	Over ground	86.1 (8.6)	1.39 (0.14)

5.1.2 Stance and Swing Phase Durations

Although no statistically significant differences in gait cycle duration were observed between treatments (or limbs or treadmill orientations), resolving the gait cycle further into stance and swing phase duration may better reflect functional changes in the gait of post-stroke subjects. Differences in stance and swing phase durations between treatments and/or limbs may indicate motor impairment [62], difficulty with limb advancement [19], weakness of the affected limb [18, 63], and instability [18] or poor balance [63]. Specifically, stance duration was found to be an indicator of the severity of motor impairment, with prolonged stance duration of the unaffected limb linked to increased impairment and weakness of the affected limb [62]. Prolonged affected limb swing duration may reflect difficulty in advancing the affected limb [19], possibly due to decreased affected ankle power during push-off in late stance and early swing [17]. Swing phase duration is also related to single limb support (SLS) duration of the contralateral limb; therefore prolonged affected limb swing corresponds to prolonged SLS of the unaffected limb. Shorter SLS duration of the affected limb (i.e. decreased swing duration of unaffected limb) may indicate difficulty in balancing or full weight bearing on the affected leg [18-19, 62]. As such, increasing affected limb SLS duration is a common rehabilitative goal [19].

As shown in Figure 39, the mean stance duration of the unaffected limb (AFO: 1.31-1.435 seconds, WA: 1.4036-1.464 seconds) was prolonged compared to the affected limb (AFO: 1.31-1.4139 seconds, WA: 1.385-1.421 seconds) for both treatments and all treadmill orientations; these differences were statistically significant for level and inclined WalkAide ambulation and decline AFO ambulation. Group A showed these significant results for AFO decline and WalkAide incline, but not WalkAide level ambulation. As only some of these differences in stance duration between limbs were statistically significant, further analysis of additional subjects is needed to confirm these results as population variability and limited sample size may mask significant differences (see Section 5.5). This similarity in

stance duration between limbs may reflect increased gait symmetry, although kinematic symmetry (see Section 5.2.1) during stance was not observed. Prolonged stance duration with the unaffected limb is commonly observed with impaired ambulators, such as post-stroke amputees, and is often caused from trouble with balance and weakness of the affected limb when compared to the unaffected limb [15, 22].

Statistically significant differences in stance duration were also observed between treatments for both limbs. As shown in Figure 39, prolonged stance duration was observed during WalkAide (unaffected: 1.40-1.46, affected: 1.38-1.42 seconds) versus AFO (unaffected: 1.31-1.45, affected: 1.31-1.41 seconds) ambulation. For all subjects, these differences were statistically significant for the affected limb during level walking and the unaffected limb during decline and level walking. For the Group A subset, these differences were only statistically significant during level walking for both limbs. The prolonged stance durations during level WalkAide ambulation suggest that the AFO may address motor impairment and weakness of the affected limb more effectively on level surfaces, although the WalkAide and AFO treatments are comparable in addressing these problems during non-level ambulation. Further analysis of additional subjects, however, may reveal additional differences with treatment (see Section 4.7 regarding limitations due to the small sample size and population nonhomogeneity).

Although stance phase is important for gait analysis, the treatments investigated in this study primarily influence the swing phase of gait; therefore differences in swing duration were expected between limbs, and between treatments. For all subjects, the affected limb swing duration (WalkAide: 0.52-0.54, AFO: 0.50-0.55 seconds, Figure 39) was significantly prolonged with respect to that of the unaffected limb (WalkAide: 0.50-0.51, AFO: 0.50-0.52 seconds) during inclined walking, suggesting potential difficulties with limb advancement during inclined ambulation. . Significantly longer swing phase durations for the unaffected versus affected limb were also observed during declined AFO ambulation for both the full population and the Group A subset.

Similar affected and unaffected limb swing duration during level and declined walking may indicate that limb advancement was easier during declined and level walking with respect to inclined walking. Swing duration during inclined walking requires additional knee and hip flexion [41]. As these post-stroke subjects lacked sufficient affected limb muscle strength for increased knee and hip flexion (e.g. lower MMT scores for the hip and knee on the affected versus unaffected side), the prolonged affected limb swing phase duration during inclined walking was to be expected.

For both the full subject population and Group A subset, the unaffected limb swing duration was significantly longer with the AFO versus WalkAide treatment during level (full population) and inclined (full population and Group A subset) walking. The prolonged unaffected limb swing duration with AFO treatment implies longer affected limb SLS duration with AFO ambulation, suggesting increased affected limb stability and balance with the AFO during inclined ambulation. Greater stability during stance during AFO ambulation is expected since the AFO provides structural support during mid to late stance. In contrast, such stance stability is not provided by the WalkAide, although stimulation often occurs during late stance prior to swing.

The stance and swing phase duration measured during level treadmill ambulation in this study are contrasted with that of previous investigations of post-stroke subjects in Table 21. The stance durations reported in the current study (WalkAide: 1.38, AFO: 1.36 seconds), while nearly double that reported for healthy, age-matched subjects (0.67 seconds, [58]), were comparable to that measured by Kim for the unaffected limb (~1.34 seconds, [42]) during over ground walking with no assistive device (75% subjects) or a cane (25% subjects). The stance duration of the affected and unaffected limbs measured in the current study, however, were greater than that reported in other studies [18, 42, 45; affected limb [42]]. Although many of the post-stroke individuals did not require assistive devices during over ground and treadmill ambulation, their stance durations were almost half (0.65-0.87 seconds) that measured in the current study. Only Roth's subjects [22], all of whom required an assistive device for ambulation, exhibited longer

stance durations (affected: 1.67, unaffected: 1.86 seconds) than those reported in the current study. Contrary to the current study for which the stance duration of the affected and unaffected limbs were similar, these prior investigations all reported shorter stance durations for the affected versus unaffected limb. Shorter stance durations for the affected limb would reflect temporal based asymmetry, associated with gait inefficiency, difficulty with balance control, and risk of musculoskeletal injury and bone loss [43]. Although the current study's subjects had longer stance durations, they demonstrated greater temporal symmetry, which may reflect increased gait efficiency and balance when compared to those in previous studies, but is more likely an artifact of treadmill ambulation [48] and/or use of the handrail support, see Section 5.6.1.4.

The mean swing duration of both the affected and unaffected limbs was longer than the swing duration reported for healthy, age-matched individuals over the age of 51 years (0.43 seconds [58]). However, the mean swing duration of both limbs was similar to that reported in previous studies involving post-stroke subjects. While the stance duration measured in the current study differed from that reported in the literature for post-stroke individuals, the swing durations were more comparable (current study: 0.51-0.53, literature: 0.44-0.65, mean: 0.51 seconds).

The prolonged stance duration and comparable swing durations reported in the current versus literature studies suggest that the prolonged gait cycle duration in the current study was primarily due to increases in stance duration. Although the subjects in the current study exhibited greater motor impairment than many of the post-stroke subjects that required assistive devices in prior investigations, the current subjects' ability to advance their limbs during swing while ambulating with the WalkAide and AFO was comparable.

Table 21: Comparison of stance and swing duration [mean and (s.d.)] during level walking with previous studies of post-stroke ambulation

Reference	N	Treatment	Assistive Device	Surface	Limb	Stance Duration		Swing	
						sec	% cycle	sec	% cycle
Current Study	8	WalkAide (4 wks)	none	Treadmill	Aff	1.38 (.31)	72.2 (3.4)	0.52 (.09)	27.8 (3.4)
					Unaff	1.39 (.32)	72.8 (4.2)	0.51 (.09)	27.2 (4.2)
		AFO	none	Treadmill	Aff	1.36 (.19)	72.0 (4.2)	0.53 (.08)	28.0 (2.7)
					Unaff	1.37 (.20)	73.0 (2.8)	0.52 (.12)	27.0 (2.8)
Brouwer, 2009 [45]	10	none	none	Over ground	Aff	0.82 (.13)	63.3 (3.2)	0.48 (.09)	36.7 (3.2)
					Unaff	0.87 (.16)	66.3 (3.1)	0.44 (.05)	33.7 (3.1)
				Treadmill	Aff	0.65 (.09)	50.8 (8.0)	0.65 (.18)	49.2 (8.0)
					Unaff	0.71 (.14)	54.7 (7.0)	0.59 (.13)	45.3 (7.0)
Kim, 2003 [42]	20	none	cane, ~25%	Over ground	Aff	1.14 (.45)	-	-	-
					Unaff	1.34 (.51)	-	-	-
Roth, 1997 [22]	25	DNS	Cane or walker	Over ground	Aff	1.67 (0.8)	74 (8)	0.59 (.27)	26 (9)
					Unaff	1.86 (.94)	82 (8)	0.41 (.10)	18 (9)
von Schroeder, 1995 [18]	49	AFO, ~15%	cane or walker, 25%	Over ground	Aff	1.06 (0.5)	67.6 (6.6)	0.51 (.23)	33.0 (6.6)
					Unaff	1.14 (0.6)	70.8 (8.6)	0.43 (0.19)	29.2 (8.6)

5.1.3 Double Limb Support Duration

Another temporal parameter commonly examined is double limb support (DLS). Prolonged affected limb DLS with either treatment may be required to control the continuously changing base of support during ambulation and may reveal balance differences between treatments. Goldie et al. [19] suggest that each DLS support phase (initial and second) be investigated individually due to the unilateral deficits that occur with stroke. Examination of both DLS periods can give insight into not only balance, but also weight acceptance and ankle function. During the initial period of DLS (e.g. loading response), the limb contacts the ground initiating stance, accepting body weight, absorbing shock, and facilitating forward progression

(see Section 2.2.1). As such, prolonged initial DLS may indicate difficulty with weight acceptance or weight transfer to that limb. The second period of DLS (e.g. pre-swing) corresponds to weight transfer from the ipsilateral to the contralateral limb in preparation for ipsilateral swing. The decreased load on the ipsilateral limb and activation of the ipsilateral gastrocnemius typically results in rapid plantar flexion. A prolonged latter DLS period for the affected limb may indicate difficulty with affected limb push off or loading response of the unaffected limb.

The initial DLS durations were contrasted between treatments (Figures 40 and 43). Prolonged (statistically significant) initial DLS durations for the affected limb were observed on all treadmill orientations while using the WalkAide versus AFO for all subjects and the Group A subset. This prolonged initial DLS may reflect greater difficulty with affected limb weight acceptance using the WalkAide compared to the AFO which may be attributed to the enhanced structural stability of the AFO.

When comparing the second period of DLS between treatments for all subjects, the affected limb exhibited significantly prolonged second DLS periods for AFO versus WalkAide use during declined and inclined ambulation. (For the Group A subject, treatment differences in the latter period of DLS were statistically significant during declined walking only.) The prolonged periods of latter DLS during non-level walking with the AFO suggests that the WalkAide, not the AFO, allows the affected limb to plantar flex more effectively during pre-swing and ease weight transfer during unaffected limb loading response.

Finally, the total DLS was significantly longer with WalkAide (0.87-0.92 seconds) versus AFO (0.80-0.89 seconds) during level ambulation for both the affected and unaffected limbs, suggesting generally improved weight acceptance and transfer with AFO versus WalkAide use. This result must be interpreted with caution, however, as the WalkAide typically resulted in prolonged stance duration than that observed during AFO ambulation. As such, the total DLS duration was also contrasted between treatments when expressed as a function of gait cycle

percentage. Using this normalized total DLS duration, WalkAide ambulation actually resulted in similar total DLS duration as during AFO ambulation (45 %, see Table 22). The prolonged periods of total DLS during level WalkAide walking may therefore be attributed to the prolonged stance duration rather than functional differences between treatments.

These DLS durations were compared to existing literature involving level ambulation of post-stroke subjects, as seen in Table 22. The normalized total DLS durations for both limbs and treatments in the current study were approximately 40-45% gait cycle, within the range of that reported in the literature (37.4 % [18], 52 % [22]). The initial DLS duration time (0.41-0.46 seconds) reported for the current study was longer for both the affected and unaffected limbs than that reported previously by Goldie et al. (0.23-0.25 seconds, [19]). However, the normalized initial DLS period for these two studies is similar (20-24% vs. 18%). The differences in initial DLS duration time between these studies may be attributed to faster cadence of Goldie's subjects (85.2 steps/min) versus that of the current study (~63 steps/min), as cadence is inversely related to gait cycle duration.

Table 22: Comparison of initial and total DLS durations [mean and (s.d.)] during level walking with previous studies of post-stroke ambulation

Reference	N	Treatment	Assistive Device	Surface	DLS Duration		Initial DLS Duration			
					sec	% cycle	sec		% cycle	
					Aff		Aff	Unaff	Aff	Unaff
Current Study	8	WalkAide (4 wks)	none	Treadmill	0.92 (.27)	45 (6)	0.45 (.21)	0.46 (.13)	21 (5)	23 (3)
		AFO	none	Treadmill	0.87 (.16)	45 (4)	0.41 (.11)	0.46 (.12)	20 (3)	24 (3)
Roth, 1997 [22]	25	DNS	Cane or walker	Over ground	-	52 (17)	-	-	-	-
von Schroeder, 1995 [18]	49	AFO, ~15%	cane or walker, 25%	Over ground	-	37 (11)	-	-	-	-
Goldie, 2001 [19]	42	none	none	Over ground	-	-	0.25 (.15)	.23 (.20)	18 (6)	18 (3)

5.1.4 Stride Time Variability

Prior investigation of post-stroke individuals ambulating in both an AFO and a neuroprosthesis has been limited (see Section 2.5). One parameter previously reported is swing time variability of the unaffected limb. Since swing duration of the unaffected limb is also the SLS duration of the affected limb, the unaffected limb swing time variability can provide insight into gait stability, balance, and fall risk [39, 64 and 65]. Swing time variability is independent of walking speed and thus may indicate changes in a subject's balance and fall risk that may not be identified by walking speed.

To quantify swing time variability, the coefficient of variation (CV) of the unaffected limb swing time was determined. As shown in Figure 44, greater swing time variability was observed during WalkAide [All: 4.45 ± 1.97, Group A: 3.97 ± 1.82] versus AFO [All: 3.43 ± 0.81, Group A: 3.47 ± 0.82] ambulation for all treadmill orientations; however, this trend was not statistically significant. Large variability of the unaffected limb swing time or affected limb SLS duration may indicate affected limb weight bearing difficulty, instability, and balance problems. As such, the trend of increased unaffected limb swing time variability during WalkAide ambulation may reflect greater affected limb stance instability or balance problems, indicative of increased risk of tripping and falling while using the WalkAide compared to the AFO.

The CV has been used previously to quantify swing time variability. Ring et al. [39] contrasted unaffected limb swing time variability for 15 post-stroke individuals using both a neuroprosthesis (NESS L300, Bioness) and an AFO during level, over ground walking. While no statistically significant differences between treatments after 4 weeks of neuroprosthetic adaptation were reported, a significant decrease in swing time variability with the neuroprosthesis versus AFO was observed after 8 weeks of neuroprosthetic acclimation.

While the current study also incorporated a 4 week neuroprosthesis acclimation period, the acclimation protocol varied between these two studies, as did the timing of the AFO gait

analysis. The initial adaptation period in the Ring's study involved only partial use of the neuroprosthesis in conjunction with the AFO, while subjects in the current study achieved full-time use of the WalkAide after 2 weeks. Ring et al. only tested the AFO at the gait session following the initial neuroprosthesis adaptation period. However, overall function of the subject could have increased during this time, decreasing the swing time variability in both the AFO and WalkAide. In the current study, half of the subjects were tested in the AFO following testing in the WalkAide. Therefore, a possible carry-over effect due to WalkAide treatment may have contributed to the decreased AFO variability that was not observed in Ring's study.

Overall, both studies indicate that a four week adaptation period with a neuroprosthesis may not be sufficient to significantly improve swing time variability when compared to an AFO. However, Ring et al. found a significant decrease in swing time variability with the neuroprosthesis versus AFO after 8 weeks of neuroprosthetic acclimation. As such, future studies should include at least 8 weeks of neuroprosthesis acclimation to assess differences in swing time variability between treatments; additional acclimation to the AFO is not necessary.

5.1.5 Summary of Temporal Parameter Findings

Temporal parameters are the most common parameters used to assess the lower extremity motor function of post-stroke individuals due to their ease of measure, reproducibility, and correlation to overall function. Temporal parameters, such as cadence and swing duration, were compared between treatments to determine if the use of a neuroprosthesis influences temporal parameters and/or temporal symmetry on both level and non-level surfaces when compared to an AFO.

Although trends of greater cadence during AFO versus WalkAide use (all treadmill orientations) and prolonged gait cycle duration during AFO versus WalkAide use (inclined walking) were observed, no statistically significant differences in either parameter were detected.

The unaffected limb exhibited significantly longer stance duration than the affected limb (both treatments), consistent with previous research. Longer unaffected limb stance durations suggest temporal asymmetry with both treatments. The stance duration was significantly prolonged during WalkAide versus AFO ambulation (level walking, both affected and unaffected limbs, declined walking for the unaffected limb only). These results suggest that the AFO may address motor impairment and weakness of the affected limb more effectively than the WalkAide during level ambulation.

As the treatment objective of both AFO and WalkAide treatment is to assist with foot clearance during swing, differences in swing duration between treatments were expected. Significantly prolonged swing duration was observed for the affected versus unaffected limb (inclined walking) for both treatments. This temporal swing asymmetry suggests that inclined walking adversely affects limb advancement for both treatments; results investigating temporal symmetry based on swing duration (SR) can be found in Section 5.2.1.1. Significantly prolonged unaffected limb swing duration (e.g. prolonged affected limb SLS) was observed during AFO versus WalkAide ambulation (level and inclined orientations). Prolonged affected limb SLS reflects increased balance and limb stability, suggesting that the AFO improves balance and limb stability when compared with the WalkAide during level and inclined ambulation. This result is most likely due to the extra structural support provided by the AFO during mid to late stance which is not present with the WalkAide.

The initial DLS duration of the affected limb was significantly longer during WalkAide versus AFO ambulation (all treadmill orientations). The shorter initial DLS duration of the affected limb during AFO use suggests improved affected limb weight acceptance, which may be attributed to the enhanced structural support of the AFO. Significantly prolonged second DLS durations were observed during AFO versus WalkAide ambulation (non-level: inclined and declined), reflecting potential difficulty with affected limb plantar flexion during pre-swing. As such, the WalkAide may better facilitate affected limb plantar flexion during pre-swing.

Greater swing time variability was observed with the WalkAide versus AFO (all treadmill orientations), although these treatment differences were not statistically significant.

5.2 GAIT SYMMETRY AND KINEMATIC PARAMETERS

Full gait analysis and investigation of lower extremity joint kinematics have been used to assess functional rehabilitative gains for individuals post-stroke; specifically, symmetry in weight bearing and weight transfer between the affected and unaffected limbs [45- 47], as well as temporal symmetry of stance and swing duration, have been investigated [30, 43, 45]. However, while temporal asymmetry may reflect differences in motor function, muscle strength, balance, and ankle function of the affected versus unaffected limbs, investigation of kinematic symmetry (or asymmetry) is necessary to fully quantify gait changes post-stroke.

5.2.1 Temporal Based Gait Symmetry

The investigation of temporal gait symmetry has most commonly been conducted using the symmetry ratio (SR). This measure, suggested by Patterson et al., may indicate gait inefficiency, difficulty with balance control, risk of musculoskeletal injury, and bone loss [43].

5.2.1.1 Symmetry Ratio

SR's for stance and swing durations for the current study were calculated using Equation 1 (Section 2.6.2.2). Across all subjects, the SR for stance duration during level ambulation was 0.99 with both the WalkAide and the AFO (Figure 45), while SR for stance duration during non-level walking ranged from 0.97 to 1.00. Analysis of the SR for stance duration of the Group A subset revealed similar results. The SR for swing duration for all subjects was also similar between treatments [AFO: 1.03 ± 0.15, WA: 1.05 ± 0.15] during level treadmill walking and non-level (0.97-1.10) walking for both treatments. Similar ranges (0.97-1.06) for swing SR's were observed for the Group A subset. No statistically significant differences in stance or swing

duration SR's were observed between treatments (AFO versus WalkAide) on any treadmill orientation, suggesting similarities in gait efficiency, balance control and injury risk between treatments (Hypothesis 1a). The lack of statistically significant differences in SR between treatments may also be attributed to the small sample size (n=8) and large inter-subject variability (0.15), as well as temporal symmetry artifact due to treadmill ambulation [48].

Patterson et al. compared the SR's for stance and swing duration for 161 post-stroke subjects and 81 age- matched controls. The SR for stance duration was 1.09 ± 0.10 for the post-stroke subjects and 1.02 ± 0.02 for healthy control subjects. For swing, the SR was 1.24 ± 0.34 for the post-stroke subjects and 1.02 ± 0.02 for healthy control subjects. The SR's for both stance and swing phase duration differed significantly between the post-stroke and age-matched healthy subjects. Note that the "affected" limb for the healthy control subjects was randomly assigned; therefore, an SR range from 1.02 ± 0.02 could also be reported as 0.98 ± 0.02.

The SR for stance duration was comparable for the post-stroke subjects in the current study was most similar to Patterson's healthy control (not post-stroke) subjects. This suggests that both the WalkAide and AFO allowed for "normal" symmetry of the stance phase duration during both level and non-level ambulation. Since the SR for stance duration of the current study's subjects was more similar to Patterson's normal versus post-stroke subjects, the current study's subjects may be considered higher functioning (i.e. greater gait efficiency and balance) than the post-stroke subjects in Patterson's study. However, Patterson's subjects were assessed during over ground walking; the moving belt of the treadmill in the current study may have forced symmetry that might not otherwise be observed during over ground walking. As such, comparisons between the study subjects are not conclusive.

The swing SR for the post-stroke subjects in the current study was also most similar o that for Patterson's healthy subjects, again suggesting near normal swing duration SRs for the current study's subjects. Near normal swing duration SR's suggest greater gait efficiency and balance, although this is inconclusive do to the potential influence of the treadmill on temporal

symmetry. The large variability in SR swing duration in the current study is similar to that measured for Patterson's post- stroke subjects, suggesting that some asymmetric swing durations may exist with WalkAide and/or AFO treatments. Due to the lack of trends or statistically significant differences in swing duration SR between treatments, neither treatment increased swing duration symmetry more effectively than the other. Further study of additional subjects is needed to confirm this.

5.2.1.2 Gait Asymmetry Index

In contrast to the aforementioned temporal symmetry measure, Ring et al. [39] defined a temporal *a*symmetry measure, the gait asymmetry index, to contrast swing phase duration for AFO versus neuroprosthesis treatments.

Previous research has shown that large swing time asymmetry indices reflect uneven weight distribution and potential increased risk of falling. Swing asymmetry may be attributed to insufficient power generated for swing by affected limb push-off and/or increased time required for affected foot placement and weight acceptance [64].

Gait asymmetry indices for swing are summarized in Figure 46. The swing gait asymmetry indices for all subjects were greater during WalkAide (0.053-0.065) ambulation on all treadmill orientations when compared to the AFO gait asymmetry indices (0.044-0.052); these differences, however, were not statistically significant (Hypothesis 1a). Similar results were observed for the Group A subset, again with no statistically significant differences in gait asymmetry index between treatments. The trend in increased swing asymmetry during WalkAide ambulation suggests increased asymmetry of weight distribution and increased risk of falling while using the WalkAide compared to the AFO on all treadmill orientations. However, these differences in the swing gait asymmetry index between treatments (~0.01) are too small to be statistically significant, and in fact, are less that the gait asymmetry indices reported in previously in the literature for post-stroke individuals (AFO: 0.20, neuroprosthesis: 0.17-0.20, [39]).

Ring et al. contrasted the gait asymmetry index for swing duration for 15 post-stroke individuals during AFO and neuroprosthesis (NESS L300) over ground ambulation after 4 and 8 weeks of neuroprosthesis adaptation. Gait asymmetry indices of 0.20 ± 0.09 and 0.19 ± 0.09 were reported during AFO and neuroprosthesis ambulation, respectively (4 week neuroprosthesis acclimation). Longer neuroprosthesis acclimation resulted in swing asymmetry indices of 0.17 ± 0.08. While no statistically significant differences in swing asymmetry indices were observed between treatments after 4 weeks of neuroprosthesis acclimation, significant differences between treatments, favoring the neuroprosthesis, were observed after 8 weeks of neuroprosthesis use.

The gait asymmetry indices for swing measured for the current study were substantially less than those reported by Ring et al., as expected due to the greater swing symmetry (less asymmetry) demonstrated by the subjects in the current study. The greater symmetry or decreased asymmetry suggest better balance and equal weight bearing (i.e. similar SLS durations) between limbs demonstrated by the current subjects, regardless of treatment. The enhanced temporal symmetry may also be attributed to the treadmill itself; previous studies have suggested that the consistent speed of the treadmill belt experienced by both limbs encourages temporal and spatial symmetry [48].

5.2.2 Kinematic Asymmetry

As indicated above, temporal symmetry (or asymmetry) measures are more appropriate for over ground walking as the moving belt of a treadmill forces the affected limb to keep pace with the unaffected limb during treadmill ambulation, enhancing temporal symmetry [48]. In addition, temporal symmetry measures assess overall motion and fail to capture the complex movement of the various limb segments, providing limited information regarding the behavior of each joint [48]. These shortcomings can be addressed by kinematic measures of symmetry.

As stated previously (Section 3.5.2), while the SR and SI can be used to investigate kinematic symmetry or asymmetry, both measures are insensitive to sign changes in joint motion,

differences in ROM that may occur between subjects or joints, and lack the resolution to observe small inter-limb differences. As such, a new measure of gait asymmetry, GA, was defined.

GA was assessed for each functional phase of stance and swing. As this research investigates treatment for drop foot, ankle and knee joint motion were of primary interest, particularly during swing as drop foot causes problems with foot clearance during swing. In addition, as the WalkAide may stimulate the anterior tibialis prior to swing and continue stimulation after heel strike, the PS and LR transition periods were also investigated.

As shown in Figures 47 and 48, increases in sagittal plane knee GA during WalkAide versus AFO ambulation were observed during IS, while increases in knee GA during AFO versus WalkAide ambulation were observed during MS and TS for all treadmill orientations. Increased knee GA indicate either increased or decreased affected versus unaffected limb knee flexion. Throughout swing, decreased knee flexion may limit foot clearance and lead to trips/falls; increased knee flexion may increase energy consumption. While Figures 47 and 48 summarize the knee GA magnitude or absolute value, all subjects (individual data reported in Appendix B) exhibited negative knee GA throughout swing for both treatments. This negative knee GA reflects decreased affected versus unaffected limb knee flexion during swing. Decreased affected limb knee flexion during IS with WalkAide use may lead to less affected versus unaffected limb toe clearance if not compensated by ankle dorsiflexion; missed WalkAide stimulations may result in both decreased affected versus unaffected limb knee flexion and ankle dorsiflexion, potentially increasing the risk of stumbles or falls. Decreased knee GA (i.e. increased affected limb knee flexion for this population) with the WalkAide during MS and TS might also indicate that knee flexion occurred later in swing when compared to the unaffected limb and AFO treatment. Delayed affected limb knee flexion, combined with missed stimulations, may again contribute to increased potential risk of stumbles or falls. Although these trends in knee GA during swing were noted, no statistically significant differences in knee GA with treatment were found. As the WalkAide and AFO primarily influence the ankle, this lack of statistically significant differences

in *knee* GA between treatments was not unexpected, although treatment differences in both knee and ankle GA are tested via Hypothesis 1a.

As expected, the sagittal plane ankle GA (Figures 49-50, 20-60%) typically exceeded that reported for the knee (20-40%). In addition, greater ankle GA was observed during swing for level and inclined ambulation with the AFO (mean: 25-58%) versus WalkAide (mean: 10-27%), particularly during mid and terminal swing, although these differences were not statistically significant. As such specific conclusions regarding differences in ankle GA during MS and TS between treatments (Hypothesis 1a) are not possible. When examined on an individual basis (Appendix C), the sign of ankle GA during MS and TS was consistently positive, reflecting increased affected versus unaffected limb dorsiflexion. While greater affected versus unaffected limb dorsiflexion may increase energy consumption, such potential increases were likely modest as affected limb knee flexion (see Figure C1 and C2) did not also increase. The greater ankle GA observed during mid and terminal swing with AFO use suggests that the AFO may have been less effective than the WalkAide in addressing foot drop, despite increased affected versus unaffected ankle dorsiflexion and greater affected versus unaffected limb toe clearance. This increased affected limb dorsiflexion and toe clearance may increase energy consumption and risk of muscle fatigue during AFO ambulation.

In addition to trends in increases in both minimum and mean sagittal plane ankle GA during AFO versus WalkAide ambulation in MS and TS, the minimum and mean ankle GA was significantly higher during IS for AFO ambulation (level and inclined walking). This increased kinematic asymmetry with AFO versus WalkAide use supports Hypothesis 1a. Increased ankle GA during IS may signify either decreased (positive GA) or increased (negative GA) ankle plantar flexion motion of the affected limb relative to the unaffected limb. Examination of individual ankle GA data during IS (Appendix C) demonstrated positive ankle GA values for nearly all subjects for both treatments, indicating decreased affected versus unaffected limb plantar flexion during IS. The greater ankle GA during IS with AFO use therefore suggests that

the AFO resulted in a greater reduction in affected limb plantar flexion during level and incline walking when compared to the WalkAide. Decreased affected limb plantar flexion after push off may reflect decreased ankle power, as needed for forward progression of the limb during swing phase. This potential decreased affected limb ankle push-off with AFO use is further supported by the shorter latter DLS or PS periods for the affected versus unaffected limb.

Although not related to the specific research hypotheses, the effect of treadmill orientation on ankle GA for both treatments was also assessed. No statistically significant differences in mean and minimum sagittal plane ankle GA during swing were found between treadmill orientations during WalkAide ambulation. However, statistically significant differences in mean and minimum ankle GA during IS were observed between treadmill orientations during AFO ambulation, both between level/inclined and declined/inclined treadmill orientations for minimum ankle GA, as well as between declined/inclined treadmill orientations for mean ankle GA. Since 7 of 8 ankle GA values were positive when assessed on an individual basis, greater ankle GA as the treadmill orientation progressed from declined to inclined indicates increasingly reduced levels of affected versus unaffected limb ankle plantar flexion during AFO ambulation from declined to inclined ambulation.

Further investigation of sagittal plane ankle GA indicated that these differences in minimum and mean GA during swing were not due to changes in unaffected ankle ROM between treatments or amongst treadmill orientation (e.g. GA denominator, Figure 51), but were due to differences in ankle motion between limbs with treatment (e.g. GA numerator, Figure 55 and 56). Mean passive ankle ROM, measured at the beginning of the gait session without the AFO, was similar during the AFO (affected: 36°,unaffected: 37°) and WalkAide (affected: 38°, unaffected: 44°) gait sessions, as was mean dorsiflexion strength (AFO: 2.25, WalkAide: 2.33) and ankle spasticity (AFO: 2.4, WalkAide: 2). Therefore the restricted active ankle ROM of the affected versus unaffected limb observed during AFO ambulation (Figure 54) may be attributed to the rigidity of the AFO, specifically the plantar flexion stops and AFO stiffness, not the functionality

of the affected ankle. As with temporal asymmetry, differences in affected versus unaffected ankle motion have been correlated with balance [56] and energy efficiency [57]. The lack of statistically significant differences in ankle GA during WalkAide ambulation may result in greater balance and energy efficiency when compared with the AFO for all treadmill orientations.

These measures of sagittal plane knee and ankle GA were contrasted with that calculated from data reported in a pilot study of a single post-stroke subject ambulating over ground with both a neuroprosthesis (NESS L300) and an AFO [66], see Table 23. Note that van Swigchem's subject was tested after 2 weeks and 1.5 years of neuroprosthesis use. Table 23 demonstrates that, if kinematic data are available, GA can be calculated for prior studies. The GA values for van Swigchem's subject were comparable to that of the current study. This single subject data also indicate that GA may change with time.

Table 23: Sagittal plane knee and ankle GA

			GA @ 2 wks [66]	GA @ 1.5y [66]	GA @ 4 wks (current study)
Knee	AFO	IS max	11	75	30
		PS min	22	0	16
	neuroprosthesis	IS max	4	2.5	36
		PS min	22	0	16
Ankle	AFO	IS min	40	17	50
		MS avg	20	0	29
	neuroprosthesis	IS min	20	3	26
		MS avg	0	33	13

5.2.3. Maximum Ankle Angular Velocity During IS

WalkAide stimulation of the anterior tibialis facilitates ankle dorsiflexion. The effectiveness of this stimulation in addressing drop foot and toe clearance during swing can be assessed by contrasting the ankle dorsiflexion rate or angular velocity during IS. Greater peak ankle angular velocity corresponding to rapid ankle dorsiflexion is desired to achieve toe clearance quickly and prevent stumbles or falls. As shown in Figure 57 the peak dorsiflexion

velocity during IS was significantly greater during WalkAide (62.4-68.3 °/sec) than during AFO (30.2-34.3 °/sec) ambulation, both for all subjects and the Group A subset. The greater dorsiflexion velocity with the WalkAide indicates that WalkAide stimulation results in more rapid dorsiflexion than that provided with the passive AFO.

5.2.4 Summary of Gait Symmetry Findings

Temporal and kinematic symmetry can provide insight into motor function, muscle strength, balance and ankle function. Hypothesis 1a predicted that *both* temporal and kinematic symmetry will improve during ambulation with a neuroprosthesis when compared with ambulation with an AFO on both level and non-level surfaces. Investigation of temporal gait symmetry revealed no statistically significant differences in stance SR, swing SR or gait asymmetry index with treatment, contrary to Hypothesis 1a. However, this temporal symmetry might be attributed to treadmill ambulation.

Kinematic asymmetry at the knee and ankle were compared between treatments to further test Hypothesis 1a and assess whether the WalkAide improved kinematic symmetry during both level and non-level walking. Although greater sagittal plane knee GA during swing was observed during WalkAide ambulation on all treadmill orientations when compared with the AFO, no statistically significant differences in knee GA were observed between treatments. Sagittal plane ankle GA was increased with the AFO versus the WalkAide during all the phases of swing, although these treatment differences were statistically significant during IS only. These increased in kinematic asymmetry with AFO use (or increased kinematic symmetry with WalkAide use) support Hypothesis 1a. The increased ankle GA reflects decreased affected limb plantar flexion with the AFO during IS, suggesting potentially decreased ankle power during PS.

5.3 TOE CLEARANCE

Toe clearance is important for evaluating the treatment efficacy of the neuroprosthesis and AFO in addressing foot drop (Hypothesis 1b). Increased toe clearance is indicative of greater efficacy and decreased risk of falling [67]. However, greater affected versus unaffected limb toe clearance may lead to increased energy costs and may reflect a fear of falling. The optimal toe clearance is not known, although values of 12.6 mm [67] and 10.3 mm [69] have been reported for young, healthy subjects.

The mean toe clearance was greater for the affected versus unaffected limb [WalkAide: affected: 29.7-32.0, unaffected: 25.9-29.6 mm; AFO: affected: 33.0-36.1, unaffected: 27.3-30.6 mm, Figure 58]. These differences in toe clearance between limbs were statistically significant for either treatment regardless of treadmill orientation. While sufficient toe clearance is necessary to prevent falls, increased affected versus unaffected limb toe clearance may reflect over compensation and potentially wasted energy and inefficient gait.

Toe clearance was contrasted between treatments to test Hypothesis 1b which stated that increased affected limb toe clearance would result with WalkAide versus AFO ambulation for both level and non-level surfaces. As shown in Figure 58 the toe clearance of the affected limb was significantly greater with AFO than during WalkAide ambulation for all subjects on level and inclined treadmill orientations, refuting this research hypothesis. The increased affected limb toe clearance during AFO ambulation reflects decreased fall risk due to toe/foot drag, although the increased affected versus unaffected limb toe clearance may also indicate some overcompensation for drop foot.

The difference between the affected and unaffected limb toe clearance was evaluated to investigate this potential overcompensation and elevated energy cost. Smaller differences reflect less wasted energy expenditure. For all subjects, the difference in toe clearance between limbs was less during WalkAide versus AFO ambulation (Figure 59), although these differences were

not statistically significant; for the Group A subset, this trend was only observed during declined and inclined walking. These results suggest that the WalkAide provides adequate toe clearance compared to the AFO – supporting, but not confirming the first research hypothesis which predicted greater efficacy with neuroprosthesis use. Further investigation with more subjects is needed to fully test this hypothesis.

In addition to examining differences in toe clearance between treatments, the potential effects of treadmill orientation on toe clearance were investigated. However, no statistically significant differences in toe clearance magnitude or relative change were observed between treadmill orientations. Similarities in toe clearance between treadmill orientations suggest that each treatment provides sufficient affected limb foot clearance when compared to the unaffected limb foot clearance, despite changes in treadmill orientation.

No prior studies investigating neuroprostheses to treat foot drop in post-stroke subjects have reported toe clearance. However, toe clearance during level, over ground walking has been investigated for young, healthy and elderly adults. Winter et al. examined toe clearance in 11 healthy young adults and found an average toe clearance of 12.6 ± 4.5 mm during level over ground walking [68], about half that measured for the post-stroke subjects in the current study. In a second study, Winter et al. reported mean toe clearance in healthy elderly subjects as 15.0 mm [68], similar to that found for young adults in his previous study. Khandoker et al. also investigated toe clearance of both young and elderly adults, reporting median toe clearances of 10.3 and 10.1 mm, respectively during over ground walking [69]. These toe clearance for healthy young and elderly subjects were 30-50% that observed for the post-stroke subjects in the current study. The greater toe clearances in the current study may be attributed to the stroke, with subjects over compensating due to foot drop and a fear of falling. Ambulation on a treadmill, as opposed to over ground walking, may have also encouraged subjects to increase their toe clearance to minimize stumbles. Methodological artifact may also have elevated these toe clearance measure. Toe clearance was based on a marker placed on the shoe; this marker

placement, however, likely affected toe clearance only by several millimeters. Small errors due to treadmill plane definition may also have affected these toe clearance measures. The treadmill plane was estimated by three treadmill markers; errors in the distance between the center of these markers and the moving belt of the treadmill may have resulted in slight (< 1 mm) over-estimation of toe clearance.

Comparison of toe clearance between treatments was used to test Hypothesis 1b, which stated that use of the neuroprosthesis would improve treatment efficacy (i.e. increase toe clearance) on level and non-level surfaces when compared to the AFO. Significantly greater affected versus unaffected limb toe clearance was observed (all treadmill orientations) for both treatments. Affected limb toe clearance was significantly increased during AFO versus WalkAide ambulation (level and inclined walking), reflecting a decreased risk of falling due to toe drag during AFO use. As such, these results do not support Hypothesis 1b. However, the increased affected limb toe clearance may also reflect potential overcompensation to foot drop and potentially increased energy expenditure. Smaller differences (not statistically significant) in affected versus unaffected toe clearance were observed during WalkAide ambulation when compared with the AFO (all treadmill orientations) suggesting that adequate toe clearance is still provided by the WalkAide.

5.4 WALKAIDE STIMULATION RELIABILITY AND TIMING

Currently there are three commercially available neuroprostheses: the Odstock Dropped Foot Stimulator, the NESS L300 and the WalkAide. Both the Odstock Dropped Foot Stimulator and the NESS L300 utilize a heel switch placed in the shoe of the affected limb to define the stimulation periods during gait. The WalkAide uses an accelerometer on the affected limb to measure the 2D tibial angle or tilt. The clinician programs the WalkAide using this tilt data, setting the stimulation initiation and termination thresholds. Additional programming parameters include the minimum “wait” time between stimulations, as well as the minimum and maximum

stimulation duration, so as to prevent anterior tibialis fatigue. Anterior tibialis stimulation is initiated when the tilt angle exceeds the stimulation initiation threshold; stimulation is terminated when the tilt angle falls below the stimulation termination threshold. If the tilt angle exceeds the stimulation initiation threshold during the wait time, stimulation will not occur.

As clinicians typically program the WalkAide using these tibial tilt angles during level walking, stimulation errors may occur during non-level walking as lower extremity limb kinematics have been shown to vary on inclined and declined surfaces [41]. Stimulation errors may result in missed or extraneous stimulations, adversely affecting the stimulation reliability, and/or changes in the stimulation timing (i.e. stimulation initiation and termination) – thereby increasing the risk of falls during non-level walking.

5.4.1 WalkAide Stimulation Reliability for Tilt Sensor-based Programming

The stimulation reliability (StR) of the WalkAide is important in assuring the safety of the individuals using the neuroprosthesis. Missed stimulations, indicated by StR values less than one, may result in little to no ankle dorsiflexion during swing, leading to insufficient toe clearance and increased risk of falling. Extraneous stimulations, indicated by StR values greater than one, may result in stimulations occurring at random times throughout the gait cycle.

As shown in Figure 60, four (S5, S8, S10, S12) of the eight subjects demonstrated near optimal StR values of one for all treadmill orientations. Only one subject (S7) demonstrated StR values of 1.5 or greater, reflecting extraneous stimulations, for all treadmill orientations. This subject walked significantly slower on the treadmill than over ground. Since the subject's WalkAide programming was set during faster over ground walking, stimulation programming included a shorter wait time; during the slow treadmill ambulation, the WalkAide stimulated twice during some gait cycles. The remaining subjects (S2, S4, S6) demonstrated extra stimulations (StR>1) for level and inclined treadmill orientations, and missed stimulations

(StR<1) for the declined treadmill orientation. Missed stimulations during declined walking were also reported by several subjects (S4, S6, and S8) during the acclimation period.

Hypothesis 2a of this study stated that the StR would be affected by treadmill orientation, namely that missed or extraneous stimulations would occur during non-level walking (e.g. StR < 1 or StR > 1during non-level walking). No statistically significant differences in tilt sensor-based StR were found between treadmill orientations, refuting Hypothesis 2a. However, several subjects (S2, S4, S6, S7) demonstrated extraneous and missed stimulations during non-level walking – providing partial support of this second research hypothesis. Extraneous and missed stimulations for subjects on declined/inclined walking surfaces suggest that, depending on the individual's environment, the PT/clinicians should include non-level walking during the WalkAide programming sessions. It is worth noting that three subjects (S8, S10, S12) who visited the PT more frequently (at least 3 times) exhibited consistent optimal StR, suggesting that supplemental programming sessions with the clinician may improve WalkAide stimulation during both level and non-level ambulation.

The extraneous (StR>1) and missed stimulations (StR<1) may be attributed to changes in baseline tilt due to treadmill orientation, accelerometer noise, alignment errors and/or movement artifact, as well as inappropriate wait times between stimulations. As seen in Figure 66, the extra peaks in tilt sensor data due to movement artifact and/or alignment errors (Figure 67) may cause extraneous stimulations if the wait time is too small (less than 0.20 sec). Extraneous tilt sensor peaks may be more common during inclined walking if the subject circumducts to assist toe clearance. Declined walking introduced a positive tilt bias (Figure 68), thereby requiring an increased tilt threshold for stimulation termination.

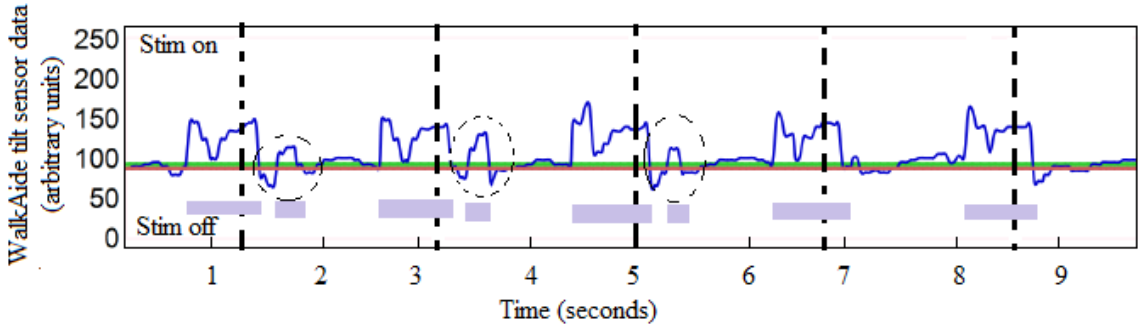

Figure 66: Sample WalkAide tilt sensor data (blue), stimulation on (green) and off (red) thresholds illustrating potential extraneous stimulations (circled) due to insufficient wait time between gait cycles.

The heel strike of each respective gait cycle is indicated by a dashed vertical line, while stimulation periods are indicated by the gray box.

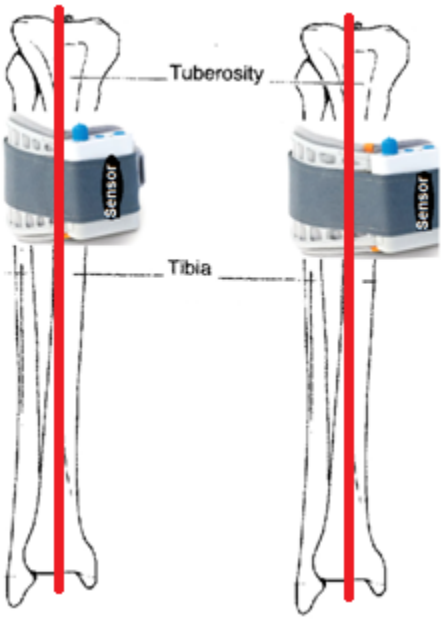

Figure 67: Sample WalkAide stimulator and tilt sensor positioning.

The red line represents alignment of the device with the tibial crest; Correct placement (left) and malalignment (right)

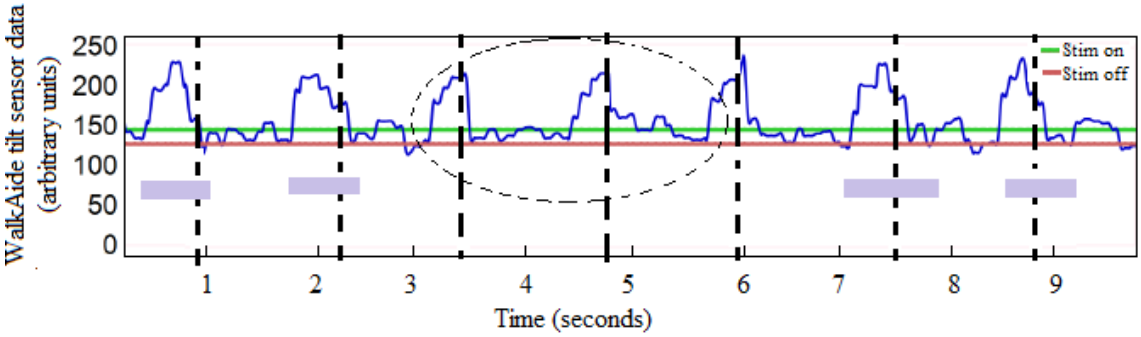

Figure 68: Sample WalkAide tilt data (blue) during declined walking illustrating the positive tilt bias, causing potential missed stimulation (circled) due to too low a stimulation termination ("off") threshold.

The heel strike of each respective gait cycle is indicated by a dashed vertical line, while stimulation periods are indicated by the gray box.

Extraneous stimulations may lead to instability due to the lack of ankle control or confusion due to unexpected stimulations. Extra stimulations during late stance may also prevent or reduce affected limb plantar flexion or push-off during PS. Unnecessary stimulation during mid-stance may result in greater affected versus unaffected limb dorsiflexion and decreased knee stability during SLS [14, 70]. These extraneous stimulations during stance may also prevent the desired stimulation during swing due to the wait time between successive stimulations.

Missed stimulations may lead to decreased affected limb toe clearance and potential increased risk of falls. No stumbles or falls resulted from missed stimulations during gait analysis since the subjects could either provide some active dorsiflexion or extra knee flexion to clear the floor, and they were using the treadmill handrails for support. Since walking trials were only 2 minutes long, the additional muscle activity needed to accommodate the missed stimulations without toe drag likely did not induce muscle fatigue.

During community ambulation, the cognitive attention needed for concomitant tasks and potential knee or hip muscle fatigue may contribute to stumbles or falls due to missed stimulations. Future studies to investigate potential cognitive effects on gait may be warranted. Cognitive demands might be increased by having the subject perform simultaneous tasks such as counting backwards during ambulation. If subjects are unable to accommodate missed stimulations with the tilt sensor programming while dual tasking, a common experience during community ambulation, the heel sensor may be a safer option for community ambulators during level and non-level walking.

5.4.2 WalkAide Stimulation Reliability for Heel Sensor-based Programming

Heel sensor data were also reviewed to determine whether the heel sensor (as used for the Odstock and NESS neuroprostheses) was more robust than the tilt sensor for stimulation control during non-level walking (Hypotheses 3a and 3b). Heel sensor-based stimulations were simulated by reviewing 3-4 WalkAide level walking trials and identifying the stimulation

initiation and termination thresholds that minimized potential extraneous stimulations due to heel sensor noise or movement artifacts (as done clinically during tilt sensor programming). The wait time and the maximum/minimum stimulation durations were the same as those set during clinical tilt-sensor based programming. As WalkAide heel sensor data were not available for subjects S7 and S8, theoretical heel sensor-based StR was based on heel loading data from the F-scan. This alternative sensor may also influence the StR values as the thickness, size, placement, range (F-scan: 0-25 psi, WalkAide: 0-150 arbitrary units), sensitivity, resolution (F-scan: 1 psi, WalkAide: 1 arbitrary unit), and sampling rate (F-scan: 50Hz, WalkAide: 25Hz) of the F-scan and WalkAide sensors differ.

StR for theoretical heel sensor-based stimulation is shown in Figure 61. Two subjects (S4, S5) exhibited missed (StR < 1) and/or extraneous (StR > 1) theoretical stimulations for declined and inclined ambulation, while the remaining subjects exhibited optimal StR values (StR = 1) during theoretical heel sensor-based stimulation. The StR values of the theoretical heel sensor-based programming were unaffected by treadmill orientation; combined subject data resulted in StR values of approximately one for all treadmill orientations. These results suggest that the heel sensor programming is not influenced by treadmill orientation, and may result in more optimal StR for than for tilt sensor-based stimulation, supporting Hypothesis 3a.

These theoretical heel sensor-based StR results, however, are dependent on the selected stimulation initiation and termination thresholds, as well as the assumed wait time and stimulation duration restrictions. The stimulation initiation and termination thresholds selected for the heel sensor-based stimulations were selected to minimize extra stimulations; as such, these StR values represent optimal programming. During clinical WalkAide programming (using the tilt or heel sensor), stimulation thresholds are similarly optimized during level over ground walking, not post hoc review of level treadmill ambulation.

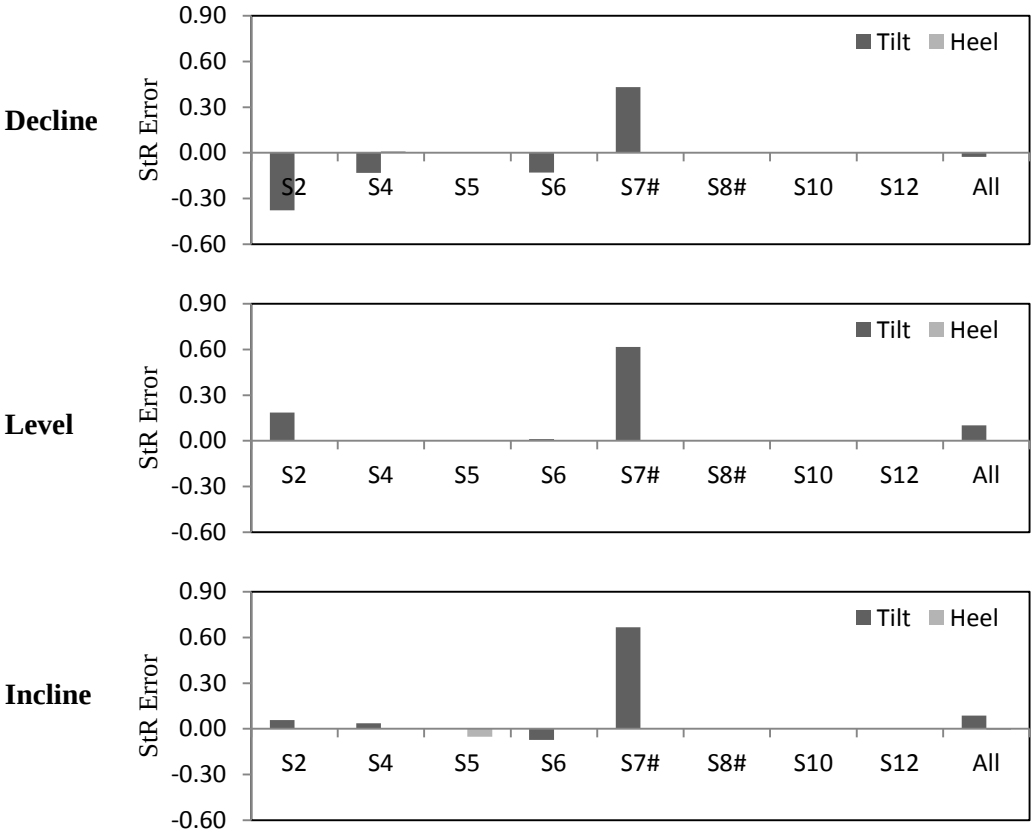

Figure 69: StR error for tilt and theoretical heel sensor-based stimulations for decline (top), level (middle), and incline (bottom) walking.

5.4.3 WalkAide Stimulation Reliability for Tilt versus Heel Sensor-based Programming

To test the research Hypothesis 3a, that StR would be improved using the heel versus tilt sensor during non-level walking, the StR results of Figures 60 and 61 were compared. Specifically, the StR error (e.g. |SR-1|*100%) was contrasted for each sensor, as shown in Figure 69. With the exception of subject S5 for inclined walking, theoretical heel sensor-based StR errors were less than that for the tilt sensor for all treadmill orientations. For subject S5, heel sensor-based StR errors was negative (e.g. StR < 1) during non-level walking, indicating missed stimulations; these missed stimulations were not observed during tilt sensor-based stimulation. These missed stimulations may be attributed to the combination of increased heel pressure due to foot and/or heel sensor movement and the wait time and minimum/maximum stimulation durations. If the stimulation begins and must be sustained for the minimum stimulation duration

followed by no stimulation during the specified wait time, the subsequent stimulation as indicated by the stimulation initiation threshold might be missed. These heel sensor-based StR errors may be minimized by more effectively positioning (i.e. by clinician or PT instead of research staff) and securing the heel sensor (i.e. taped directly to the shoe insole rather than merely secured to the shoe's heel counter).

Although differences in StR errors between tilt versus theoretical heel-sensor based stimulation were observed in individual subjects, overall StR errors across all subjects did not differ significantly between sensors for any treadmill orientation. This suggests that the theoretical heel sensor-based stimulation was not more reliable than tilt sensor-based stimulation during non-level walking, contrary to Hypothesis 3a. This lack of statistically significant differences in StR between sensors, however, may be due to the small sample size. Seventy-five per cent (6/8) subjects exhibited optimal (unity) theoretical heel sensor-based StR values and only 50% (4/8) subjects demonstrated optimal tilt sensor-based StR values during non-level walking. As such, one might argue that the heel sensor may still be considered a better sensor for more reliable stimulation control during non-level walking, partially supporting Hypothesis 3a.

There are advantages and disadvantages associated with the WalkAide heel and tilt sensors. In contrast to the WalkAide heel sensor, the tilt sensor (integrated within the WalkAide stimulation unit) is positioned with respect to the tibia as the WalkAide cuff and unit are secured on the affected limb. As such, no sensor is placed inside the shoe and individuals are free to wear any shoe (even heels and sandals) and WalkAide donning is more facile – although consistent tilt sensor alignment along the tibial crest may be a concern. Use of the heel sensor requires that the shoe has an adequate heel box, and placement of this sensor can be difficult for hemiparetic individuals with upper limb impairment. Improved StR with the heel sensor may offset these shoe placement limitations. Several subjects complained about missed stimulations during non-level walking while using the tilt sensor-based programming.

5.4.4 WalkAide Tilt Sensor-based Stimulation Initiation and Termination Timing

Hypothesis 2b stated that, due to changes in tibial angle during non-level walking [40], tilt sensor-based StR and stimulation timing would vary during non-level versus level walking.

The primary goal of WalkAide stimulation of the anterior tibialis is to provide ankle dorsiflexion during swing to assist with foot clearance for individuals with foot drop. Stimulation may also extend into loading response to minimize foot slap (see Section 2.2.1). As seen by Figure 63, tilt sensor-based stimulation initiation (StI) occurred at approximately -20% swing for most subjects (S2, S4, S7, S8, S10, and S12), indicating that StI occurred during late stance, more specifically during PS. In contrast, for subject S5, StI occurred during early swing (~5-10% swing, during IS); for subject S6, StI was seemingly early, occurring at approximately -30 to -40% swing (during mid or terminal stance).

In general, StI was delayed as the treadmill orientation progressed from declined to inclined, occurring later in stance, closer to swing phase itself. Post-hoc testing revealed statistically significant differences in StI between declined/inclined treadmill orientations; no statistically significant differences in StI were observed between level and non-level surfaces, as posed in Hypothesis 2b. StI occurred earlier in the gait cycle during declined walking (-15% swing) when compared to inclined (-5% swing) walking. The earlier stimulation initiation during declined walking may result in decreased plantar flexion during push-off, muscle activity that assists forward progression of the affected limb during swing. However, such plantar flexion may be less necessary for declined walking [41] since the limb does not need to be elevated as high for swing (i.e. less knee flexion during declined walking [41]). The earlier dorsiflexor stimulation and reduced push-off during PS during declined walking may, in fact, be advantageous. Delayed dorsiflexion stimulation until IS during declined ambulation may result in decreased affected limb dorsiflexion and toe clearance during swing.

For most subjects (S2, S4, S6, S7, S8, and S10; Figure 63), tilt sensor-based StT occurred during the first 20% of the subsequent gait cycle (i.e. 120% gait cycle), indicating that StT typically occurred during LR of the following cycle. For subjects S6 and S7, ST occurred as late as 130% gait cycle for inclined walking. StT was also quite late for subjects S5 and S12 (130 to 150% gait cycle). For these subjects and the inclined walking trials of subjects S6 and S7, StT most likely occurred during the subsequent mid or terminal stance period. Although differences in StT between treadmill orientations were observed for some subjects (S4, S5, S6, and S7), no statistically significant differences in StT timing were observed between level and non-level walking trials, contrary to Hypothesis 2b.

Changes in StI (but not StT) timing during non-level walking reflect changes in the stimulation duration during non-level walking. StI occurred earlier in the gait cycle, prior to swing, during declined vs. level ambulation. Since the StT timing did not change, the earlier StI reflected prolonged stimulation duration during decline walking, potentially leading to muscle fatigue if declined walking trials are extended. Muscle fatigue, combined with the missed stimulations observed during declined walking, may lead to stumbles or falls if the subject can no longer accommodate the missed stimulation due to fatigue. For inclined walking, the StI occurred later in the gait cycle, closer to swing, thereby reflecting shorter stimulation periods. Potential muscle fatigue during inclined walking is therefore unlikely.

During clinical WalkAide programming with the tilt sensor, the stimulation initiation and termination tilt thresholds are specified, as well as the minimum wait time between stimulation periods and the minimum and maximum stimulation durations. The lack of changes in StT timing during non-level walking may be due, at least in part, to these wait times and minimum/stimulations durations. Delayed StT may reduce dorsiflexion during terminal swing or loading response, contributing to potential toe drag and stumbles or foot slap and first ankle rocker inhibition, respectively; however this was not observed for this study. Limiting the first

ankle rocker may adversely affect affected limb forward progression and shock absorption during LR.

In addition to changes in the StI timing itself, the variability of the StI and StT timing was also examined for tilt sensor-based stimulation. Large variability (2-30% gait cycle, mean 9.62% gait cycle) in both stimulation timing parameters was observed, although this variability was substantially reduced (2-18% gait cycle, mean 6.7% gait cycle) after eliminating gait cycles with extraneous stimulations (Figure 65). This remnant variability in StI and StT timing may be attributed to tilt sensor errors (e.g. movement artifact due to lower limb tremors or circumduction, sensor alignment errors) and/or programming parameters (wait time, stimulation minimum or maximum duration). Note that the StI and StT variability was reduced for three subjects (S8, S10, and S12) who met with the clinician one week prior to gait analysis for WalkAide programming adjustment. It is recommended that future studies require that subjects meet more frequently (at least 3 times) with the clinician for WalkAide adjustment and meet with the clinician one week prior to gait analysis for final adjustments.

5.4.5 WalkAide Heel Sensor-based Stimulation Initiation and Termination Timing

The StI, StT and the variability in these parameters were also investigated for theoretical heel sensor-based stimulation, using both the WalkAide heel sensor and F-scan plantar pressure data (Figure 64).

For the theoretical heel sensor-based stimulations, the StI occurred at approximately -20% swing; statistically significant differences in heel sensor-based StI were not observed with treadmill orientation. Since heel sensor-based StI occurred around -20% swing for inclined walking, plantar flexion during late stance needed during inclined walking may be prevented by the early dorsiflexion activity during heel sensor-based stimulations, resulting in prolonged swing.

Heel sensor-based StT timing occurred slightly earlier in stance (approximately +10% stance or 110% gait cycle) than that observed during tilt sensor-based (approximately +20% stance or 120% gait cycle) stimulation. Only one subject (S8) demonstrated theoretical heel sensor-based StT during swing (95% gait cycle, -5% stance), prior to the subsequent gait cycle. StT typically extended into stance to provide ankle support during LR and prevent foot slap. Earlier StT during heel sensor-based stimulation may limit the stimulation's effect on ankle stability during LR (specifically for subject S8). As with StI timing, no statistically significant differences in heel sensor-based StT were observed between treadmill orientations.

5.4.6 WalkAide Tilt versus Heel Sensor-based Stimulation Initiation and Termination Timing

As predicted by Hypothesis 3b, more consistent StI timing was observed among subjects with the heel versus tilt sensor, with only one subject (S5; versus two with the tilt sensor) demonstrating StI earlier than -20% swing. Unlike the tilt sensor-based stimulations, the StI timing for heel sensor-based programming was not influenced by treadmill orientation, potentially making the heel sensor a more predictable and reliable sensor for non-level ambulation (which is not observed during programming), supporting Hypothesis 3b. However, the consistent StI timing with the heel sensor may adversely affect inclined ambulation. StI earlier in stance would initiate dorsiflexion sooner, possibly preventing the extra plantar flexion needed during inclined ambulation.

As for tilt sensor-based stimulation, theoretical heel sensor-based StT occurred during the subsequent stance period; however, StT occurred slightly earlier with the heel (+10% stance) versus the tilt (+20% stance) sensor. Stimulation typically continues into the following stance duration to provide ankle support during LR and prevent foot slap; therefore earlier StT during heel versus tilt sensor-based stimulation may limit the heel sensor-based stimulation's effect on ankle stability during LR when compared with the tilt sensor-based stimulation. Additionally, the consistent StI and earlier StT timing with heel sensor-based programming corresponds to a

decreased stimulation duration, possibly minimizing the likelihood of tibialis muscle fatigue during walking. Therefore, these results suggest that the tilt sensor may support improved weight transfer and balance throughout LR when compared to the heel sensor, but may lead to muscle fatigue during prolonged walking.

To test Hypothesis 3b, the variability in the StI and StT timing was compared between sensors. The variability in theoretical heel sensor-based stimulation timing was substantially less (1-10% gait cycle) than that observed with tilt sensor-based timing (2-18% gait cycle), Figure 65, supporting Hypothesis 3b. This reduced variability is most likely due to the greater consistency of the heel sensor data; the careful review of heel sensor data during multiple gait cycles for selection of the stimulation initiation and termination thresholds may also have reduced this theoretical timing variability.

Based on the decreased variability in StI/StT timing (and increased StR) of heel sensor versus tilt sensor--based stimulation, the heel sensor may be considered more reliable for non-level walking, supporting Hypotheses 3b (and Hypothesis 3a). However, the differences in tilt sensor StI with treadmill orientation may actually improve the efficacy of tilt sensor-control, increasing the stimulation period during declined walking. Further testing is recommended to determine if the changes in stimulation timing observed with the tilt sensor during non-level walking are more beneficial than the consistency of the heel sensor stimulation timing. Future gait analysis, inclusive of kinetic data, may help determine if the delayed StI observed during tilt sensor-based stimulation facilitates greater push-off (i.e. increased ankle moment and/or ankle power) during PS while ambulating on inclined versus level surfaces.

5.4.6 Summary of WalkAide Stimulation Findings

Tilt sensor-based stimulation reliability and timing on level versus non-level surfaces were compared to test Hypotheses 2a and 2b. No statistically significant differences in StR were observed between level and non-level walking trials for tilt-sensor based programming, refuting

Hypothesis 2a. Several subjects (4 out of 8), however, exhibited missed or extraneous stimulations during non-level walking with tilt sensor-based programming, providing partial support of Hypothesis 2a. Statistically significant differences in StI timing were observed between level and non-level walking trials for tilt sensor-based programming. The StI occurred earlier in the gait cycle during declined versus level walking; StI occurred later in the gait cycle during inclined versus level walking. These results support Hypothesis 2b. No trends or statistically significant differences in StT timing were observed between level and non-level walking for tilt sensor-based programming.

Stimulation reliability and timing based on tilt sensor- versus heel sensor-based programming were contrasted to test Hypotheses 3a and 3b. No statistically significant differences in SR were observed between tilt versus heel sensor-based programming during non-level walking, contrary to Hypothesis 3a. Missed or extraneous stimulations were observed for 6 of 8 subjects with tilt sensor-based stimulation; missed or extraneous stimulations were only observed for 4 of 8 subjects with heel sensor-based programming during non-level walking trials. These differences in stimulation reliability support, but do not confirm, Hypothesis 3a. StI or StT timing variability was decreased for heel sensor- versus tilt-sensor-based stimulation during non-level walking; however these differences between sensors were not statistically significant and therefore no definitive conclusions about Hypothesis 3b can be made. Based on the StR and StI/StT variability results, the heel sensor may provide more consistent stimulation when compared with the tilt sensor during non-level walking, potentially supporting Hypotheses 3a and 3b.

5.5 POWER ANALYSIS

As the post-stroke population in this study was relatively non-homogeneous, the various measured parameters were not normally distributed and non-parametric tests (Wilcoxon rank sum, Friedman) were necessary for statistical analysis. Non-parametric tests do not assume

normal distribution, are more robust, and can accommodate the variability in the post-stroke population; however, these non-parametric tests are less powerful than parametric tests. As such, larger sample sizes are often required [55]. To determine the number of subjects necessary to achieve statistical differences ($p>0.05$) with sufficient power ($P>0.80$), as well as to determine the power of the various parameters of the current study, *a priori* and post hoc power analyses were conducted.

A priori power analyses were conducted to determine the subject population needed to detect statistically significance differences, assuming small (0.20), medium (0.50) and large (0.80) effect sizes [55] – see Table 15. *A priori* power analyses for individual, non-paired measures (i.e. GA and toe clearance) indicate that 15, 35, and 208 subjects are necessary to detect statistically significant differences with small, medium, and large effects. Due to the robust nature of non-parametric tests, only large differences between means, accompanied by small standard deviations (i.e. large effect size, $d>1.0$) result in statistically significant differences for measures such as GA and toe clearance of only eight subjects. The use of repeated measures in non-parametric statistical analyses facilitate identification of statistically significant difference with a small sample size, as supported by the *a priori* power analyses of paired, repeated measures analysis (1, 2, and 7 subjects needed for small, medium, and large effect sizes, respectively).

Multiple *a priori* and post hoc power analyses based on the actual effect sizes of the current study were conducted both to determine the number of subjects necessary to detect statistical significance and the actual statistical power – see Table 16. *A priori* analyses of the affected limb temporal parameters indicate that eight subjects are sufficient to observe statistically significant differences ($p>0.05$) with sufficient power ($P>0.80$) in affected limb stance and total DLS durations for declined and level walking. Despite an adequate sample size and sufficient power, these temporal parameters did not demonstrate statistically significant differences between treatments during declined walking, due to both the large parameter variability and the temporal symmetry due to treadmill ambulation. For inclined swing duration

and level initial DLS, nine subjects are necessary to observe statistically significant differences in these parameters between treatments with sufficient power; statistically significant differences in these parameters between treatments, however, were observed for the 8 subjects in the current study. The low power in these parameters may be attributed to the relatively large variability in the temporal parameters and the associated small effect sizes (d=0.18). Finally, power analysis of the swing and second DLS duration during level walking indicate that more than 1000 subjects may be needed to detect statistically significant differences between treatments for these parameters. The large number of subjects required for statistically significant treatment differences in these parameters may be attributed to the high variability, but is most likely due to the similarity in means (< 0.01 seconds difference in means for both parameters). Therefore, even with the addition of more subjects, statistically significant differences in level swing and second DLS between WalkAide and AFO treatments are unlikely.

To determine the power and number of subjects necessary to observe statistically significant differences in temporal symmetry measures (Hypothesis 1a), *a priori* and post hoc power analyses were conducted, see Table 17. For stance and swing SR, as well as the gait asymmetry index, more than 100 subjects are required to observe statistically significant differences between treatments during both level and non-level walking. The lack of statistically significant differences in temporal symmetry measures between treatments may be attributed to the small to moderate effect sizes ($0.02 \leq d \leq 0.33$) and low power (0.06-0.21) due to the small sample size (n=8) and large inter-subject variability (SR: 0.15, gait asymmetry index: 0.60). Additionally, the similar temporal symmetry may have been an artifact of treadmill ambulation [48].

A priori and post hoc power analyses were also conducted for the GA (non-repeated) measures of the knee and ankle to determine the number of subjects need to test Hypothesis 1a (kinematic symmetry) and to quantify the power of the current results, respectively. As seen in Table 18, more than 100 subjects are needed to detect statistically significant differences in knee

GA during between treatments (Hypothesis 1a). These requisite large sample sizes suggest that differences in knee GA between treatments may not be observed, even with an expanded subject population. The post hoc power analyses indicate that the power of the knee GA results is very low ($P<0.10$); as such, the probability for a false negative is high [55].

As the WalkAide and AFO treatments for drop foot directly affect the ankle, it is not surprising that fewer subjects ($N\geq10$) are needed to detect treatment differences in ankle GA. These smaller requisite sample populations may be attributed to the greater effect size in ankle (0.16-1.39) versus knee (0.03-0.36) GA and the moderate power (0.11-0.71 for level and inclined ambulation). Statistically significant treatment differences in minimum and mean ankle GA during IS on level and inclined treadmill orientations were observed for the current study, although the associated power (mean GA: ~0.65, minimum GA: ~0.66) was moderate. Two to four additional subjects may be necessary to achieve power of 0.80. Statistically significant differences between treatments were not observed for ankle GA during MS and TS, consistent with the *a priori* power analyses which indicate that more than 51 subjects are required for medium effect sizes (d=0.40-0.57). With the high probability of false negative due to low power (0.11-0.18) and medium effect sizes (d=0.41-0.57), the small sample size (n=8) and the large inter-subject variability likely prevented detection of statistically significant differences in mean dorsiflexion during MS and TS.

A *priori* and post hoc power analyses for toe clearance (Table 19) indicate that only 2-4 subjects are required to observe statistically significant differences in affected limb toe clearance between treatments for level and non-level walking (Hypothesis 1b). These sample size projections and study power are consistent with the aforementioned hypotheses testing, which showed statistically significant differences in affected limb toe clearance between treatments during level and inclined walking. Larger subject populations ($N\geq24$) are required to detect statistically significant treatment differences in the affected limb toe clearance during declined walking. The low power (0.05-0.29) and moderate effect size ($0.02\leq d\leq0.11$) suggest that there is

a high probability of a false negative for these measures and that more subjects may reveal differences that could not be detected with eight subjects.

Power analyses were also conducted to investigate differences in toe clearance between the affected and unaffected limbs (i.e. toe clearance symmetry) with treatment. Fewer (38-85) subjects, although still more than that in the current study, are required to detect statistically significant differences in toe clearance symmetry between treatments. These requisite enlarged sample populations may be attributed to the large variability in measured toe clearance for the 8 subjects in the current study. The low power ($P<0.23$) and medium to large effect sizes ($d=0.44$-0.68) again suggest that statistically significant differences might be observed if the subject population were increased ($N>38$).

In summary, more than 100 subjects are required to observe statistically significant differences with small to medium effect sizes for temporal symmetry measures (SR and gait asymmetry index). Eight to more than 100 subjects are required to detect statistically significant treatment differences in ankle and knee GA, respectively. Finally, 38-85 subjects are required to detect significant differences in affected limb toe clearance between treatments.

5.6 STUDY LIMITATIONS

There are a number of limitations that must be taken into consideration when interpreting the results of this study. These limitations result from assumptions and/or errors associated with study protocol procedures, sensors, data acquisition, and/or signal processing.

5.6.1 Study Procedures

5.6.1.1 Event Detection

All kinematic data were processed and reviewed as a function of gait cycle. The respective gait cycle events, heel strike and toe off, were identified based on the velocity of the

heel and toe markers since the attempted use of foot switches generated inconsistent and inaccurate event detection (see Section 3.3.2). These heel and toe markers were placed on the shoes themselves, not directly on the calcaneus and metatarsal heads, as subjects were shod during all walking trials. As such, although the subject wore the same shoes for both AFO and WalkAide ambulation, heel and toe marker position errors may have occurred. Such position errors, in turn, affect heel and toe marker velocity calculations. For example, positioning the toe marker on the shoe surface over the second metatarsal phalangeal joint, as done in the current study, likely resulted in earlier toe off estimation (earlier swing initiation, potentially prolonging swing duration) than would have been estimated by positioning the toe marker at the tip of the shoe. In addition to position/velocity errors due to marker placement on the shoe itself, heel position/velocity is also affected by the relative heel compliance (a compliant heel may delay heel strike); the effects of heel compliance are likely minimal, however, as the same shoes were worn for both gait analysis sessions.

5.6.1.2 Shod Walking

To facilitate comparison between AFO and WalkAide ambulation, the same shoes were worn for both gait analysis sessions. However, accommodation of an AFO often requires that shoes be 1-2 sizes larger than otherwise worn. The larger shoes can be laced more tightly during WalkAide trials, but may result in kinematic errors with respect to the toe marker. In addition, the extra shoe length may require increased toe clearance during WalkAide ambulation to minimize toe drag and potential stumbling.

The placement of the heel and toe markers on the shoes may also influence the estimated lower extremity joint kinematics and the GA measures at the knee and ankle. A study examining the difference in gait parameters of normal children ambulating in shoes compared to barefoot walking found that all differences in shod kinematic data for the lower extremity joints remained within 5° of that for barefoot ambulation [71].

5.6.1.3 KAD Alignment

Another factor potentially influencing joint kinematics was the placement of the knee alignment devices (KAD's) used to correct the knee axis of rotation. If the KAD is placed slightly anterior to the femoral epicondyle, additional knee flexion is introduced; posteriorly placed KAD's introduce a knee extension bias and may result in knee hyperextension artifacts. Rotation of the KAD's about the femoral epicondyles in the transverse plane may also introduce both coronal and sagittal plane knee kinematic errors [72]. KAD positioning errors also affect the pelvis angle [72], and in turn, the hip, knee and ankle angles. Due to the sensitivity of joint kinematics to KAD placement, extra care was taken with KAD placement during static trials.

5.6.1.4 Treadmill Walking

An additional procedural limitation was the inclusion of treadmill versus over ground ambulation. While over ground ambulation may better approximate household and community ambulation, inclusion of multiple trials of level and non-level walking over ground would likely have introduced subject fatigue (and/or reduced study power if less than 30 gait cycles per orientation were included). The moving belt of the treadmill encourages the affected limb to keep pace with the unaffected limb, minimizing potential temporal asymmetry. To facilitate level and non-level walking, the treadmill was positioned on a wooden ramp; the increased height may have heightened a potential fear of falling, resulting in decreased walking speed and potentially increased knee or hip flexion and toe clearance.

While treadmill ambulation can be justified to minimize potential subject fatigue, these subjects were not blinded to treadmill speed. Most subjects opted to keep the same belt speed for all treadmill orientations (S2, S4, S5, S7, S8, S10) for a given treatment – thereby minimizing temporal differences between treadmill orientations. Only one subject (S4) opted to keep the same belt speed for both treatments.

5.6.1.5 WalkAide Programming

The small number of visits with the study affiliated PT or clinician in charge of WalkAide programming (minimum of 3 visits, 2 for initial programming and a third 1 week following initial programming) required of all study subjects may have adversely affected StR and consistency of stimulation timing. At the beginning of this study, subjects were only required to see the PT one week after the initial fitting. Additional visits with the PT were only scheduled upon the subject's request; only one of these early subjects (S5) revisited the PT. As the study progressed, it was highly recommended that the subjects revisit the PT the week prior to the gait session; the last four subjects were compliant with this recommendation and revisited the PT before the gait session. These additional PT visits the week prior to gait analysis likely resulted in improved WalkAide programming and the resultant StR and consistency of stimulation timing. Three of the last four subjects demonstrated optimal StR's (i.e. StR=1) for all treadmill orientations, confirming that these additional PT visits were beneficial. Therefore the StR and variability in stimulation timing for the first four study subjects were likely worst case scenarios.

5.6.2 Sensors

5.6.2.1 Skin Movement

Joint kinematic data are also influenced by marker placement and skin movement. For all subjects except S4 and S6, the ASIS markers were referenced to markers on the lateral pelvis for more consistent viewing. While the lateral pelvis markers were placed so as to minimize skin movement, these markers were not placed on bony prominences and may be subject to increased movement errors. These potential movement artifacts also affect the subsequent pelvic orientation calculations and hip, knee and ankle joint kinematics as per the PlugInGait model. Such skin movement errors, however, are likely less than 4.2^{o} in the sagittal plane [73].

5.6.2.2 WalkAide Heel Sensor

Errors in theoretical heel sensor-based stimulation timing may have been introduced due the WalkAide heel sensor itself. This heel sensor (20 mm diameter) was smaller than the heel, and its position was not prescribed by the study PT. The relative thickness of the sensor (~0.5 mm) likely introduced plantar stress concentrations, resulting in potentially higher heel loading that was more easily detected. Sensor placement (and movement) errors may have affected theoretical heel sensor-based stimulation reliability and timing (maximum timing error estimated at 0.20 seconds based on F-scan heel sensor box sensitivity analysis for a single subject). Further stimulation reliability and timing errors may have been introduced via the selected heel initiation and termination stimulation thresholds, although these errors are likely minimal as the thresholds were based upon review of several (3-4) level walking trials so as to minimize potential extraneous and/or missed stimulations. Heel sensor-based stimulation reliability and timing data based on clinical programming (stimulation initiation/termination thresholds, heel not tilt sensor-based wait time, minimum/maximum stimulation duration) is recommended to minimize potential theoretical heel sensor-based programming errors.

5.6.2.3 F-scan Insoles

In the current study, theoretical heel sensor-based stimulation was based on F-scan data for subjects S7 and S8. Since the F-scan insole covered the full plantar surface, a 20 mm by 20 mm box was used to approximate the WalkAide heel sensor (same box location for all treadmill trials). This heel box was positioned to encompass the region of peak heel loading (due to potential stress concentration due to concurrent placement on mal-functioning WalkAide heel sensor, 0.15 mm thick). While potential movement of the F-scan insole was likely minimal, potential movement of the mal-functioning WalkAide heel sensor may have occurred; such movement errors, however, were likely minimal as the heel box loading remained inclusive of

peak forces. Both the timing and magnitude of the heel sensor loading was also influenced by the location of the F-scan heel sensor box, with a maximum error estimate of 0.20 seconds and 0.268 psi (only 1.4% of the maximum F-scan heel sensor amplitude) based on F-scan sensitivity analysis results for a single subject.

5.6.3 Data Acquisition

5.6.3.1 Synchronization

The WalkAide tilt sensor-based stimulation timing parameters, StI and StT, were based on the anterior tibialis surface EMG signal (1800 Hz). As the WalkAide does not provide analog output, direct data acquisition of WalkAide stimulation data was not possible. Due to nerve conduction (54 m/s [74-75] and muscle conduction velocity (5 m/s [76]), stimulation timing delays of 19.7-29.5 msec assuming 0.33-0.5 m travel) may have been introduced to EMG-based stimulation timing estimates. Additional stimulation timing errors may be introduced due to the low pass filtering (zero-phase digital filter) and threshold detection algorithm. These potential stimulation timing errors, however, were likely less than that which would have occurred if WalkAide stimulation data (acquired at 25 Hz) had been used.

5.6.3.2 Sampling Rate

Theoretical heel sensor-based stimulation timing estimates were likely affected by the modest WalkAide sampling rate (25 Hz, 40 msec sampling interval). This sampling rate may have introduced theoretical stimulation timing errors of ± 40 msec. For subjects S7 and S8 for whom the theoretical heel sensor-based stimulation was based on F-scan data, potential stimulation timing errors were likely somewhat less as F-scan data were acquired at 50 Hz (sampling interval of 20 msec).

5.6.4 Signal Processing

In addition to potential errors in heel strike and toe off event detection due to marker placement, additional event detection timing and kinematic errors may have been introduced due to digital filtering (Woltring filter) prior to event detection. Such filtering, although strongly recommended prior to numerical differentiation so as to calculate heel and toe velocity, may have affected the location of velocity minima and maxima used in the event detection algorithm. The motion data were sampled at 120 Hz, so potential errors of 1-2 frames correspond to timing errors less than 17 msec.

5.6.5 Proposed Study Modifications

5.6.5.1 Enhanced Clinician Involvement

In the current study, only a single follow-up clinician visit was required for WalkAide programming adjustment; additional visits were encouraged based on subject perceived need. A minimum of 3 visits with the clinician to review WalkAide programming is recommended, with one of these sessions to occur one week prior to scheduled gait analysis to optimize programming and minimize stimulation variability.

In addition, it is recommended that the clinician attend the WalkAide gait analysis session to confirm WalkAide electrode/tilt sensor placement and programming during over ground ambulation. The clinician might also program the WalkAide for heel sensor-based stimulation (setting the heel sensor stimulation initiation/termination thresholds, wait time, and minimum/maximum stimulation duration). This clinical programming would ensure that theoretical heel sensor-based stimulation was more clinically appropriate, facilitating clinical comparison of the two WalkAide sensor stimulation control options. Subsequent acclimation to clinical heel sensor-based stimulation and a second WalkAide gait analysis session might also be incorporated.

5.6.5.2 Over Ground Versus Treadmill Ambulation

Over ground walking may reveal temporal differences masked by treadmill ambulation. Over ground walking trials would also facilitate comparison of self-selected walking speed with the two treatment options, as well as potential acquisition of force plate data for kinetic analysis. Such over ground walking trials might augment treadmill walking trials, as treadmill walking maximizes the number of gait cycles acquired without introducing fatigue. The self-selected walking speed determined during over ground walking can be used to select a proper treadmill walking speed for each subject for each respective treatment.

The use of an instrumented split-belt treadmill (e.g. Bertec) for gait analysis maximizes the number of gait cycles (and study power), minimizes potential fatigue, and enables kinetic data acquisition to assess joint moment and power. For any treadmill gait analyses, subjects should be blinded to treadmill speed.

5.6.5.3 Subject Recruitment and Sample Population Size

The final recommendation for future studies is to increase the study sample size. Based on power analyses, a minimum of 10 subjects is necessary to detect potentially significant differences in GA between treatments. Studies investigating toe clearance treatment effects should include at least 38 subjects. Similarly, studies contrasting temporal parameters between treatments during treadmill walking should include a minimum of 25 post-stroke subjects.

5.7 FUTURE STUDY

Based on the results of this study, two hypotheses are suggested for potential future study:

1) Ankle power during PS increases during WalkAide versus AFO ambulation on both level and inclined surfaces.

Such a study requires kinetic data and might include over ground walking with force plates (risk of subject fatigue and limited power due to restricted gait cycles) or incorporate a kinetically-instrumented treadmill. Kinetic analysis might also reveal differences in gait symmetry during PS (and LR) that were not apparent during kinematic analysis. Note that as the need for active push-off is reduced for declined walking, declined walking trials need not be included.

2) Clinically programmed WalkAide heel sensor- versus tilt sensor-based stimulation results in improved stimulation reliability and more consistent timing during both level and non-level walking.

   Clinical heel sensor-based programming requires setting the specific sensor stimulation initiation/termination thresholds, the wait time, and the minimum/maximum stimulation duration. Alternatively, the stimulation reliability and timing of the clinically programmed WalkAide (tilt sensor) and Bioness (heel sensor), randomly selected, might be contrasted after 8 weeks of neuroprosthesis acclimation.

CHAPTER 6: CONCLUSION

A common gait impairment for stroke survivors is the inability to dorsiflex the ankle during swing causing the foot to drag along the floor (i.e. foot drop); approximately 20% of the stroke survivors suffer from foot drop. This impairment limits mobility, increases instability, and increases the individual's risk of tripping and falling.

Foot drop is traditionally treated with an ankle-foot-orthosis (AFO), a solid or articulated plastic brace which holds the ankle in a neutral position during swing, preventing the foot from dragging along the ground. An alternative treatment is the WalkAide, a neuroprosthesis which electrically stimulates the peroneal nerve to activate the dorsiflexors during swing. Without structural constraints, the neuroprosthesis may increase ankle range of motion and improve gait symmetry and efficacy on both level and non-level surfaces when compared to an AFO, enhancing a person's mobility and safety inside and outside the home.

This study used gait analysis to quantify the differences in temporal (SR, gait SI) and kinematic (GA) gait symmetry and efficacy (toe clearance) of eight post-stroke individuals suffering from foot drop during level and non-level ambulation using a WalkAide and an AFO. The neuroprosthesis stimulation reliability and timing were also contrasted for two programming options (clinically programmed tilt sensor, theoretical heel sensor programming).

Research Hypothesis 1: The use of a neuroprosthesis for post-stroke individuals with drop foot will improve temporal and kinematic gait symmetry, as well as treatment efficacy on both level and non-level surfaces compared to an AFO. No statistically significant differences in temporal symmetry (SR or SI) were observed between treatments, partially refuting this hypothesis. Additionally, no differences in knee GA were observed. Trends (not statistically significant) of greater ankle GA during MS and TS with the AFO versus the WalkAide were observed. Statistically significant differences in ankle GA during IS between treatments were observed, supporting this hypothesis. Greater (not statistically significant) mean and minimum

ankle GA during IS was found for AFO ambulation (all treadmill orientations) when compared to the WalkAide. Significantly greater affected limb toe clearance was observed during level and inclined AFO ambulation when compared with the WalkAide, contrary to this hypothesis. However, greater toe clearance symmetry was observed during WalkAide ambulation, partially supporting this hypothesis.

Research Hypothesis 2: Non-level walking will adversely affect the WalkAide tilt sensor-based stimulation reliability and timing. No statistically significant differences in tilt sensor-based stimulation reliability were observed between level and non-level walking trials. However, four of the eight subjects demonstrated missed/extraneous stimulations ($SR \neq 1$) during non-level walking, partially supporting this hypothesis; only three of eight subjects demonstrated missed/extraneous stimulations during level walking. StI timing occurred significantly closer to swing as the treadmill processed from declined to inclined orientations; no statistically significant differences in StT timing were observed between level and non-level walking. Although the observed changes in StI with level versus non-level ambulation support this hypothesis, these changes may be beneficial as StI closer to swing during inclined ambulation may allow for greater ankle plantar flexion during PS.

Research Hypothesis 3: heel sensor versus tilt sensor stimulation control will improve stimulation reliability and exhibit more consistent stimulation timing during non-level ambulation. No statistically significant differences stimulation reliability or timing were observed between tilt and theoretical heel sensor-based stimulation programming. However, more subjects (6/8) demonstrated optimal StR with theoretical heel sensor-based programming, supporting but not confirming this hypothesis. Stimulation variability decreased (not statistically significant) with the heel sensor-based stimulation, again supporting but not confirming this hypothesis.

Based on the current study, post-stroke individuals may exhibit increased kinematic ankle symmetry and possibly increased gait efficacy (toe clearance symmetry) during level and non-level community ambulation with a WalkAide instead of an AFO. For post-stroke individuals who are household ambulators only and/or demonstrate an inconsistent gait pattern, an AFO is likely sufficient to minimize foot drop. Heel sensor-based programming may provide more reliable and consistent dorsiflexion stimulation than tilt sensor-based programming. Additional clinical programming sessions may be required to allow increased plantar flexion needed during inclined walking. Community ambulators who commonly encounter inclines may benefit from tilt sensor-based stimulation.

The current study also introduced a new measure to assess kinematic asymmetry (GA). Unlike previous measures of gait symmetry, SR or SI, the new GA measure also accounts for differences in kinematic motion sign (e.g. dorsiflexion versus plantar flexion). Since this measure normalizes the difference between limbs by the affected limb joint ROM, it facilitates comparison between subjects and joints. This new GA measure provides a potentially valuable option for future comparisons of kinematic asymmetry due to different treatments and/or pathologies.

REFERENCES

[1] *Heart Disease and Stroke Statistical Update: 2009.* (2010). Retrieved February 12, 2010, from http://www.americanheart.org.

[2] Wade, D. T., Wood, V. A., Heller, A., Maggs, J., & Langton Hewer, R. (1987) Walking after stroke. measurement and recovery over the first 3 months. *Scandinavian Journal of Rehabilitation Medicine, 19 (1),* 25-30.

[3] Hyndman, D., Ashburn, A., & Stack, E. (2002) Fall events among people with stroke living in the community: Circumstances of falls and characteristics of fallers. *Archives of Physical Medicine and Rehabilitation, 83(2),*165-170.

[4] *The WalkAide Manual (2008).* Innovative Neurotronics. Print.

[5] *Stroke Sourcebook* (1993). Joyce Brennfleck Shannon. 1st edition.

[6] *Post-stroke Rehabilitation: Fact Sheet* (2010). Retrieved November 15, 2010, from http://www.ninds.nih.gov/disorders/stroke/poststrokerehab.htm

[7] *Stroke rehabilitation information.* (2007). Retrieved November 15, 2010, from http://www.ninds.nih.gov/disorders/stroke/stroke.htm

[8] *Impact of stroke*. (2010). Retrieved November 15, 2010, from http://americanheart.org/presenter.jhtml?identifier=1033

[9] Jørgensen, H. S., Nakayama, H., Raaschou, H. O., & Olsen, T. S. (1995). Recovery of walking function in stroke patients: The copenhagen stroke study. *Archives of Physical Medicine and Rehabilitation, 76*(1), 27-32.

[10] Keenan, M. A., Perry, J., & Jordan, C. (1984). Factors affecting balance and ambulation following stroke. *Clinical Orthopaedics and Related Research, (182)*(182), 165-171.

[11] Perry, J., Garrett, M., Gronley, J. K., & Mulroy, S. J. (1995). Classification of walking handicap in the stroke population. *Stroke; a Journal of Cerebral Circulation, 26*(6), 982-989.

[12] Perry, J., & Burnfield, J. M. (2010). *Gait analysis :Normal and pathological function* (2nd ed.). Thorofare, NJ: Slack.

[13] Hausdorff, J. M., & Alexander, N. B. (2005). *Gait disorders :Evaluation and management*. Boca Raton: Taylor & Francis.

[14] Rose, J., & Gamble, J. G. (2006). *Human walking* (3rd ed.). Philadelphia: Lippincott Williams & Wilkins.

[15] Woolley, S. M. (2001). Characteristics of gait in hemiplegia. *Topics in Stroke Rehabilitation, 7*(4), 1-18.

[16] Wall, J. C., & Ashburn, A. (1979). Assessment of gait disability in hemiplegics. hemiplegic gait. *Scandinavian Journal of Rehabilitation Medicine, 11*(3), 95-103.

[17] Olney, S. J., & Richards, C. (1996). Hemiparetic gait following stroke. part I: Characteristics. *Gait & Posture, 4*(2), 136-148.

[18] von Schroeder, H. P., Coutts, R. D., Lyden, P. D., Billings, E.,Jr, & Nickel, V. L. (1995). Gait parameters following stroke: A practical assessment. *Journal of Rehabilitation Research and Development, 32*(1), 25-31.

[19] Goldie, P. A., Matyas, T. A., & Evans, O. M. (1996). Deficit and change in gait velocity during rehabilitation after stroke. *Archives of Physical Medicine and Rehabilitation, 77*(10), 1074-1082.

[20] Bohannon, R. W. (1992). Walking after stroke: Comfortable versus maximum safe speed. *International Journal of Rehabilitation , 15*(3), 246-248.

[21] Brandstater, M. E., de Bruin, H., Gowland, C., & Clark, B. M. (1983). Hemiplegic gait: Analysis of temporal variables. *Archives of Physical Medicine and Rehabilitation, 64*(12), 583-587.

[22] Roth, E. J., Merbitz, C., Mroczek, K., Dugan, S. A., & Suh, W. W. (1997). Hemiplegic gait. relationships between walking speed and other temporal parameters. *American Journal of Physical Medicine & Rehabilitation, 76*(2), 128-133.
[23] Olney, S. J., Griffin, M. P., Monga, T. N., & McBride, I. D. (1991). Work and power in gait of stroke patients. *Archives of Physical Medicine and Rehabilitation, 72*(5), 309-314.
[24] Platts, M. M., Rafferty, D., & Paul, L. (2006). Metabolic cost of over ground gait in younger stroke patients and healthy controls. *Medicine and Science in Sports and Exercise, 38*(6), 1041-1046.
[25] Michael, K. M., Allen, J. K., & Macko, R. F. (2005). Reduced ambulatory activity after stroke: The role of balance, gait, and cardiovascular fitness. *Archives of Physical Medicine and Rehabilitation, 86*(8), 1552-1556.
[26] Patterson, K. K., Parafianowicz, I., Danells, C. J., Closson, V., Verrier, M. C., Staines, W. R., McIlroy, W. E. (2008). Gait asymmetry in community-ambulating stroke survivors. *Archives of Physical Medicine and Rehabilitation, 89*(2), 304-310.
[27] Richards, C., & Knutsson, E. (1974). Evaluation of abnormal gait patterns by intermittent-light photography and electromyography. *Scandinavian Journal of Rehabilitation Medicine Supplement, 3*, 61-68.
[28] Knutsson, E., & Richards, C. (1979). Different types of disturbed motor control in gait of hemiparetic patients. *Brain : A Journal of Neurology, 102*(2), 405-430.
[29] Burdett, R. G., Borello-France, D., Blatchly, C., & Potter, C. (1988). Gait comparison of subjects with hemiplegia walking unbraced, with ankle foot orthosis, and with air-stirrup® brace. *Physical Therapy, 68*(8), 1197-1203.
[30] Waters, R. L., McNeal, D., & Perry, J. (1975). Experimental correction of footdrop by electrical stimulation of the peroneal nerve. *The Journal of Bone and Joint Surgery, 57*(8), 1047-1054.
[31] Bowker, P. (1993). *Biomechanical basis of orthotic management*. Oxford England; Boston: Butterworth-Heinemann.
[32] Laufer, Y., Hausdorff, J. M., & Ring, H. (2009). Effects of a foot drop neuroprosthesis on functional abilities, social participation, and gait velocity. *American Journal of Physical Medicine & Rehabilitation, 88*(1), 14-20.
[33] Liberson, W. T., Holmquest, H. J., Scott, D., & Dow, M. (1961). Functional electrotherapy in stimulation of the peroneal nerve synchronized with the swing phase of gait in hemiparetic patients. *Archives of Physical Medicine and Rehabilitation, 42*, 101-105.
[34] Sheffler, L. R., & Chae, J. (2007). Neuromuscular electrical stimulation in neurorehabilitation. *Muscle & Nerve, 35*(5), 562-590.
[35] Shimada, Y., Matsunaga, T., Misawa, A., Ando, S., Itoi, E., & Konishi, N. (2006). Clinical application of peroneal nerve stimulator system using percutaneous intramuscular electrodes for correction of foot drop in hemiplegic patients. *Neuromudulation: Technology at the Neural Interface, 9*(4), 320-327.
[36] Taylor, P. N., Burridge, J. H., Dunkerley, A. L., Wood, D. E., Norton, J. A., Singleton, C., & Swain, I. D. (1999). Clinical use of the odstock dropped foot stimulator: Its effect on the speed and effort of walking. *Archives of Physical Medicine and Rehabilitation, 80*(12), 1577-1583.
[37] Burridge, J., Taylor, P., Hagan, S., Wood, D., & Swain, I. (1997). The effects of common peroneal stimulation on the effort and speed of walking: A randomized controlled trial with chronic hemiplegic patients. *Clinical Rehabilitation, 11*(3), 201-210.
[38] Hausdorff, J. M., & Ring, H. (2008). Effects of a new radio frequency-controlled neuroprosthesis on gait symmetry and rhythmicity in patients with chronic hemiparesis. *American Journal of Physical Medicine & Rehabilitation, 87*(1), 4-13.

[39] Ring, H., Treger, I., Gruendlinger, L., & Hausdorff, J. M. (2009). Neuroprosthesis for footdrop compared with an ankle-foot orthosis: Effects on postural control during walking. *Journal of Stroke and Cerebrovascular Diseases, 18*(1), 41-47.
[40] Sheffler, L. R., Hennessey, M. T., Naples, G. G., & Chae, J. (2006). Peroneal nerve stimulation versus an ankle foot orthosis for correction of footdrop in stroke: Impact on functional ambulation. *Neurorehabilitation and Neural Repair, 20*(3), 355-360.
[41] McIntosh, A. S., Beatty, K. T., Dwan, L. N., & Vickers, D. R. (2006). Gait dynamics on an inclined walkway. *Journal of Biomechanics, 39*(13), 2491-2502.
[42] Kim, C. M., & Eng, J. J. (2003). Symmetry in vertical ground reaction force is accompanied by symmetry in temporal but not distance variables of gait in persons with stroke. *Gait & Posture, 18*(1), 23-28.
[43] Patterson, K. K., Gage, W. H., Brooks, D., Black, S. E., & McIlroy, W. E. (2010). Evaluation of gait symmetry after stroke: A comparison of current methods and recommendations for standardization. *Gait & Posture, 31*(2), 241-246.
[44] Sadeghi, H., Allard, P., Prince, F., & Labelle, H. (2000). Symmetry and limb dominance in able-bodied gait: A review. *Gait & Posture, 12*(1), 34-45.
[45] Brouwer, B., Parvataneni, K., & Olney, S. J. (2009). A comparison of gait biomechanics and metabolic requirements of overground and treadmill walking in people with stroke. *Clinical Biomechanics, 24*(9), 729-734.
[46] Robinson, R. O., Herzog, W., & Nigg, B. M. (1987). Use of force platform variables to quantify the effects of chiropractic manipulation on gait symmetry. *Journal of Manipulative and Physiological Therapeutics, 10*(4), 172-176.
[47] Herzog, W., Nigg, B. M., Read, L. J., & Olsson, E. (1989). Asymmetries in ground reaction force patterns in normal human gait. *Medicine and Science in Sports and Exercise, 21*(1), 110-114.
[48] Shorter, K. A., Polk, J. D., Rosengren, K. S., & Hsiao-Wecksler, E. T. (2008). A new approach to detecting asymmetries in gait. *Clinical Biomechanics, 23*(4), 459-467.
[49] Duncan, P. W., Propst, M., & Nelson, S. G. (1983). Reliability of the Fugl-Meyer assessment of sensorimotor recovery following cerebrovascular accident. *Physical Therapy, 63*(10), 1606-1610.
[50] Clarkson, H. M., & Gilewich, G. B. (1989). *Musculoskeletal assessment : Joint range of motion and manual muscle strength*. Baltimore: Williams & Wilkins.
[51] *Vicon User Manual* (2002). Vicon Motion Systems. Print
[52] Devore, J. L. (1982). *Probability and statistics for engineering and the sciences* (1st ed.). Belmont, CA, USA: Thomson Brooks/Cole.
[53] Rice, W.R (1989). Analyzing tables of statistical tests. *Evolution, 43*, 223–5.
[54] Chan, Y. H. (2003). Biostatistics 101: Data presentation. *Singapore Medical Journal, 44*(6), 280-285.
[55] Bhavna, P., Dunne, M., & Armstrong, R. (2010). Sample Size Estimation and Statistical Power Analyses. *Optometry Today*.
[56] Mecagni, C., Smith, J. P., Roberts, K. E., & O'Sullivan, S. B. (2000). Balance and ankle range of motion in community-dwelling women aged 64 to 87 years: A correlational study. *Physical Therapy, 80*(10), 1004-1011.
[57] Ballaz, L., Plamondon, S., & Lemay, M. (2010). Ankle range of motion is key to gait efficiency in adolescents with cerebral palsy. *Clinical Biomechanics, 25*(9), 944-948.
[58] Blanc, Y., Balmer, C., Landis, T., & Vingerhoets, F. (1999). Temporal parameters and patterns of the foot roll over during walking: Normative data for healthy adults. *Gait & Posture, 10*(2), 97-108.
[59] Gok, H., Kucukdeveci, A., Altinkaynak, H., Yavuzer, G., & Ergin, S. (2003). Effects of ankle-foot orthoses on hemiparetic gait. *Clinical Rehabilitation, 17*(2), 137-139

[60] Voigt, M., & Sinkjaer, T. (2000). Kinematic and kinetic analysis of the walking pattern in hemiplegic patients with foot-drop using a peroneal nerve stimulator. *Clinical Biomechanics, 15*(5), 340-351.
[61] Daly, J. J., Roenigk, K., Holcomb, J., Rogers, J. M., Butler, K., Gansen, J., Ruff, R. L. (2006). A randomized controlled trial of functional neuromuscular stimulation in chronic stroke subjects. *Stroke; A Journal of Cerebral Circulation, 37*(1), 172-178.
[62] Hsu, A. L., Tang, P. F., & Jan, M. H. (2003). Analysis of impairments influencing gait velocity and asymmetry of hemiplegic patients after mild to moderate stroke. *Archives of Physical Medicine and Rehabilitation, 84*(8), 1185-1193.
[63] Chen, G., Patten, C., Kothari, D. H., & Zajac, F. E. (2005). Gait deviations associated with post-stroke hemiparesis: Improvement during treadmill walking using weight support, speed, support stiffness, and handrail hold. *Gait & Posture, 22*(1), 57-62.
[64] Hausdorff, J. M. (2005). Gait variability: Methods, modeling and meaning. *Journal of Neuroengineering and Rehabilitation, 2*, 19.
[65] Yogev, G., Plotnik, M., Peretz, C., Giladi, N., & Hausdorff, J. M. (2007). Gait asymmetry in patients with parkinson's disease and elderly fallers: When does the bilateral coordination of gait require attention? *Experimental Brain Research, 177*(3), 336-346.
[66] van Swigchem, R., Weerdesteyn, V., van Duijnhoven, H. J., den Boer, J., Beems, T., & Geurts, A. C. (2011). Near-normal gait pattern with peroneal electrical stimulation as a neuroprosthesis in the chronic phase of stroke: A case report. *Archives of Physical Medicine and Rehabilitation, 92*(2), 320-324.
[67] Winter, D. A. (1992). Foot trajectory in human gait: A precise and multifactorial motor control task. *Physical Therapy, 72*(1), 45-53; discussion 54-6.
[68] Winter, D. A., Patla, A. E., Frank, J. S., & Walt, S. E. (1990). Biomechanical walking pattern changes in the fit and healthy elderly. *Physical Therapy, 70*(6), 340-347.
[69] Khandoker, A. H., Lynch, K., Karmakar, C. K., Begg, R. K., & Palaniswami, M. (2010). Toe clearance and velocity profiles of young and elderly during walking on sloped surfaces. *Journal of Neuroengineering and Rehabilitation, 7*, 18.
[70] Lehmann, J. F., Warren, C. G., & DeLateur, B. J. (1970). A biomechanical evaluation of knee stability in below knee braces. *Archives of Physical Medicine and Rehabilitation, 51*(12), 688-695.
[71] Oeffinger, D., Brauch, B., Cranfill, S., Hisle, C., Wynn, C., Hicks, R., & Augsburger, S. (1999). Comparison of gait with and without shoes in children. *Gait & Posture, 9*(2), 95-100.
[72] Schwartz, M. H., Trost, J. P., & Wervey, R. A. (2004). Measurement and management of errors in quantitative gait data. *Gait & Posture, 20*(2), 196-203.
[73] Benoit, D. L., Ramsey, D. K., Lamontagne, M., Xu, L., Wretenberg, P., & Renstrom, P. (2006). Effect of skin movement artifact on knee kinematics during gait and cutting motions measured in vivo. *Gait & Posture, 24*(2), 152-164.
[74] Singh, N., Behse, F., & Buchthal, F. (1974). Electrophysiological study of peroneal palsy. *Journal of Neurology, Neurosurgery & Psychiatry, 37(11)*), 1202-213.
[75] Connelly, D. M., Rice, C. L., Roos, M. R., & Vandervoort, A. A. (1999). Motor unit firing rates and contractile properties in tibialis anterior of young and old men. *Journal of Applied Physiology, 87*(2), 843-852.
[76] Ganong, W. F. (1991) *Review of Medical Physiology*. Norwalk, CT: Appleton & Lange.

APPENDIX A: Fugl-Meyer Assessment Tasks (adapted from [50])

LOWER EXTREMITY (supine)	I. Reflex activity—tested in supine position. Achilles _____ Patellar _____	0—No reflex activity 2—Reflex activity	4	
Supine	II. a. Flexor Synergy [Task 1] Hip flexion _____ Knee flexion _____ Ankle dorsiflexion _____ [Task 2] b. Extensor synergy—(motion is resisted) Hip extension _____ Adduction _____ Knee extension _____ Ankle plantarflexion _____	a. 0—Cannot be performed 1—Partial motion 2—Full motion b. 0—No motion 1—Weak motion 2—Almost full strength compared to normal	6 8	
SITTING (knees free of chair)	III. Movement Combining Synergies [Task 3] a. Knee flexion beyond 90° _____ b. Ankle dorsiflexion _____	a. 0—No active motion 1—From slightly extended position knee can be flexed but not beyond 90° b. 0—No active flexion 1—Incomplete active flexion 2—Normal dorsiflexion	4	
STANDING	IV. Movement Out of Synergy Hip at 0° [Task 4] a. Knee flexion _____ b. Ankle dorsiflexion _____	a. 0—Knee cannot flex without hip flexion 1—Knee begins flexion without hip flexion, but doesn't get to 90°, or hip flexes during motion 2—Full motion as described b. 0—No active motion 1—Partial motion 2—Full motion	4	
SITTING	V. Normal Reflexes Knee flexors _____ Patellar _____ Achilles _____	0—2 of the 3 are markedly hyperactive 1—One reflex is hyperactive or 2 reflexes are lively 2—No more than 1 reflex lively	2	
(SUPINE)	VI. Coordination/Speed [Task 5] Heel to opposite knee (5 repetitions in rapid succession) a. Tremor _____ b. Dysmetria _____ c. Speed _____	a. 0—Marked tremor 1—Slight tremor 2—No tremor b. 0—Pronounced or unsystematic 1—Slight or systematic 2—No dysmetria c. 0—Six seconds slower than unaffected side 1—Two to 5 seconds slower 2—Less than 2 seconds difference	6	
		TOTAL MAXIMUM LOWER EXTREMITY SCORE	34	

Task 1: While supine, the subject is asked to flex his/her affected hip, knee, and ankle in one complete motion

Task 2: While supine, the subject is asked to extend his/her affected hip, knee, and ankle in one complete motion while staff adds resistance via the forefoot

Task 3: While sitting faced forward in a chair, the subject is asked to flex his/her affected knee and ankle in one complete motion

Task 4:While standing, the subject is asked to flex his/her affected knee and ankle, without flexing his/her hip, in one complete motion

Task 5:While supine, the subject is asked to bring his/her non-affected ankle to his/her affected knee 5 times without stopping. The subject is then asked to repeat this task, but bring his/her affected ankle to his/her unaffected knee.

APPENDIX B: GA for all phases of the gait cycle (swing and stance)

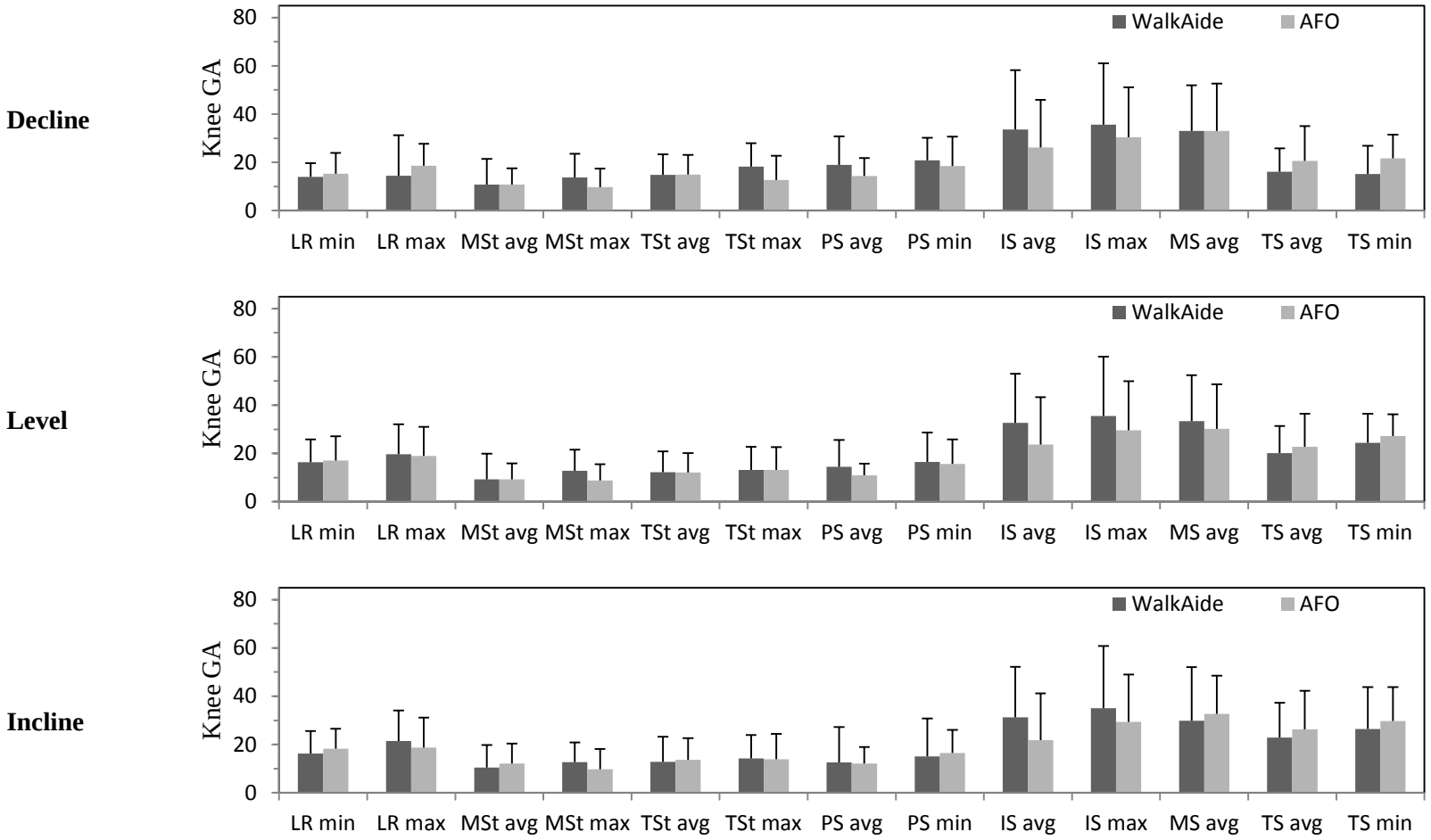

Figure B1: GA of knee motion for all functional gait phases of all subjects during decline (top), level (middle), and inclined (bottom) treadmill walking.

+, ++ denotes a statistically significant difference (0.05 and 0.01 levels, respectively)

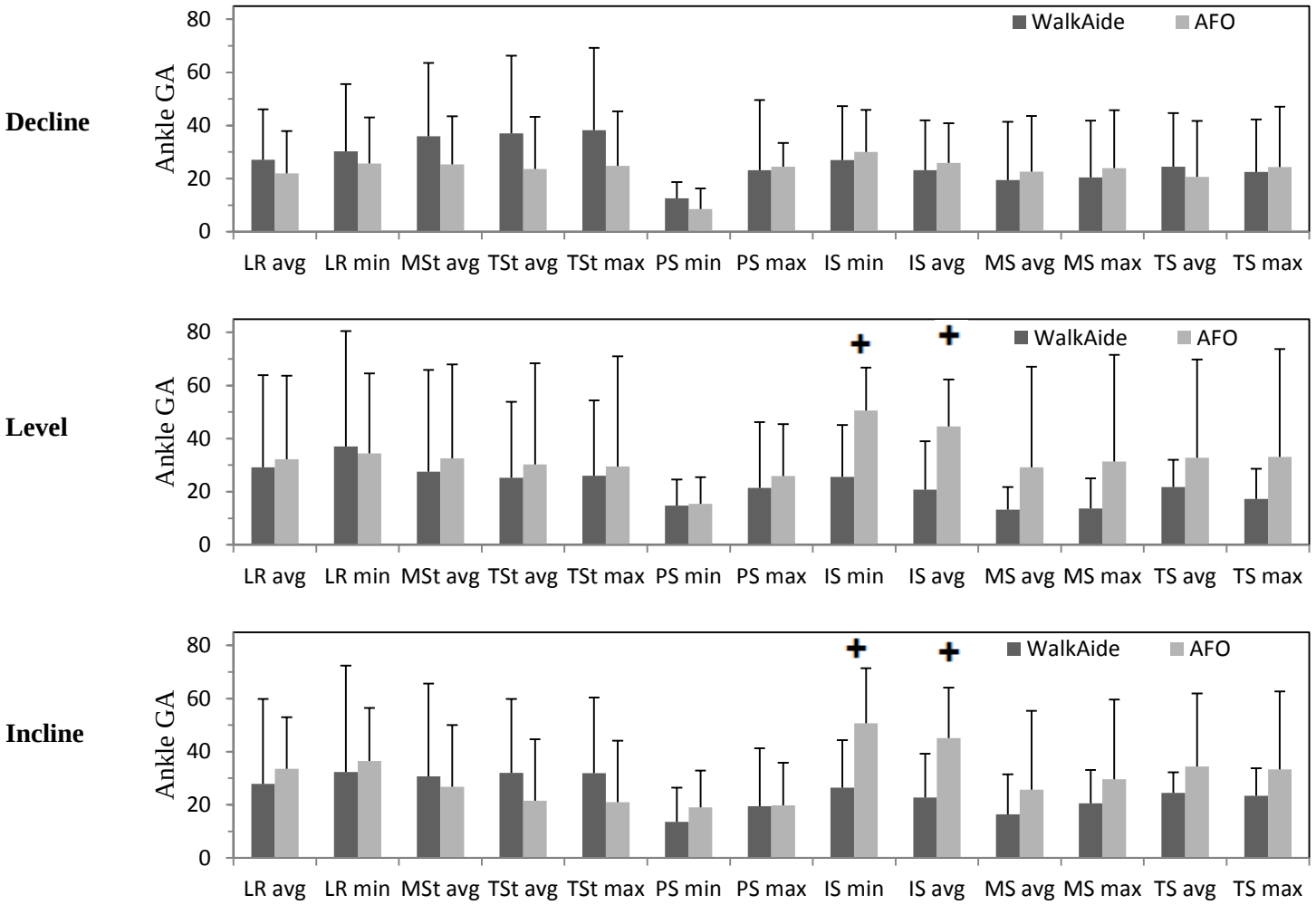

Figure B2: GA of ankle motion for all functional gait phases of all subjects during decline (top), level (middle), and inclined (bottom) treadmill walking.

+, ++ denotes a statistically significant difference (0.05 and 0.01 levels, respectively)

APPENDIX C: Individual ankle GA during IS, MS, and TS

MS mean knee motion:

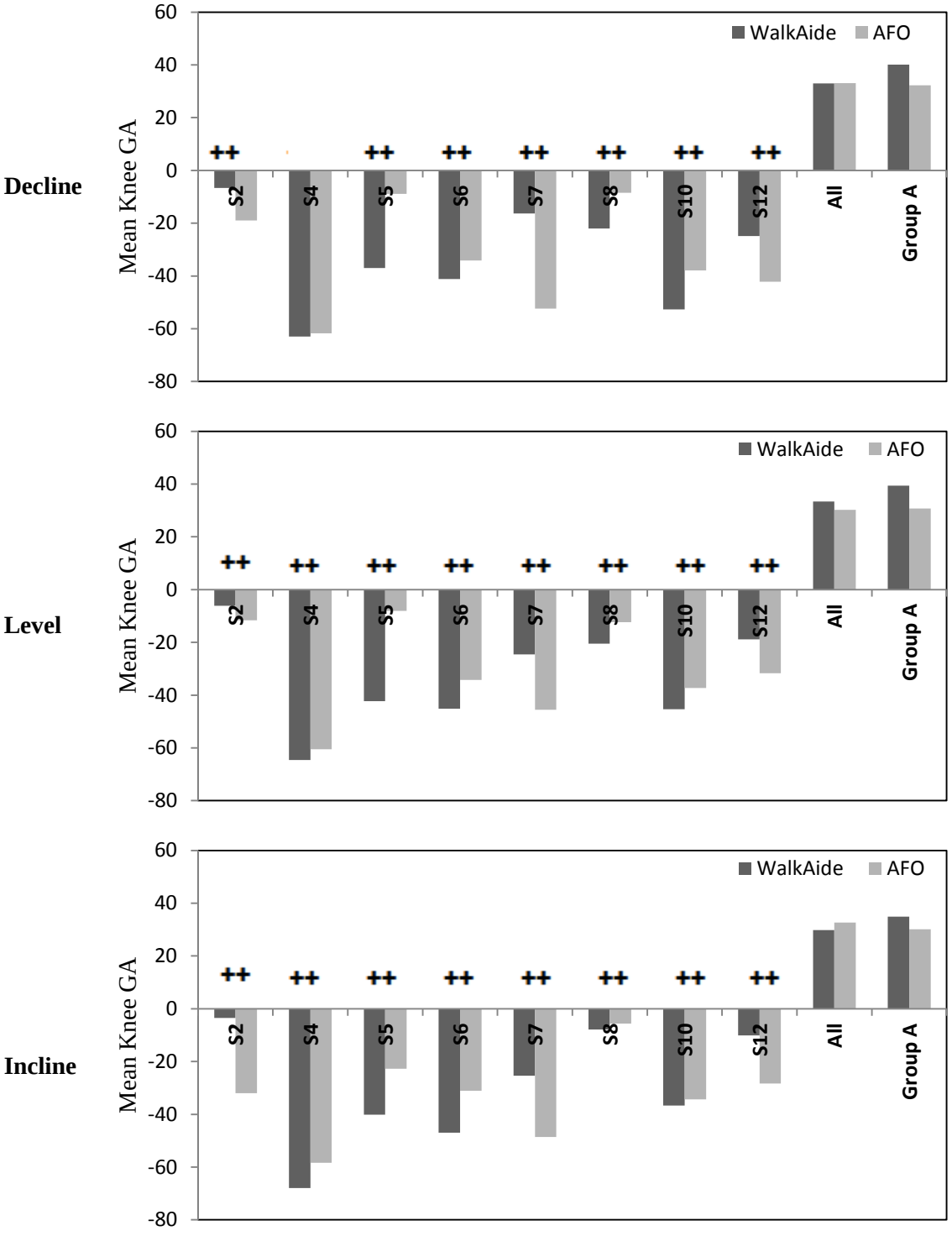

Figure C1: Individual GA for mean knee motion during MS contrasting treatments for declined (top), level (middle), and inclined (bottom) walking.
+, ++ denotes a statistically significant difference (0.05 and 0.01 levels, respectively)

TS mean knee motion:

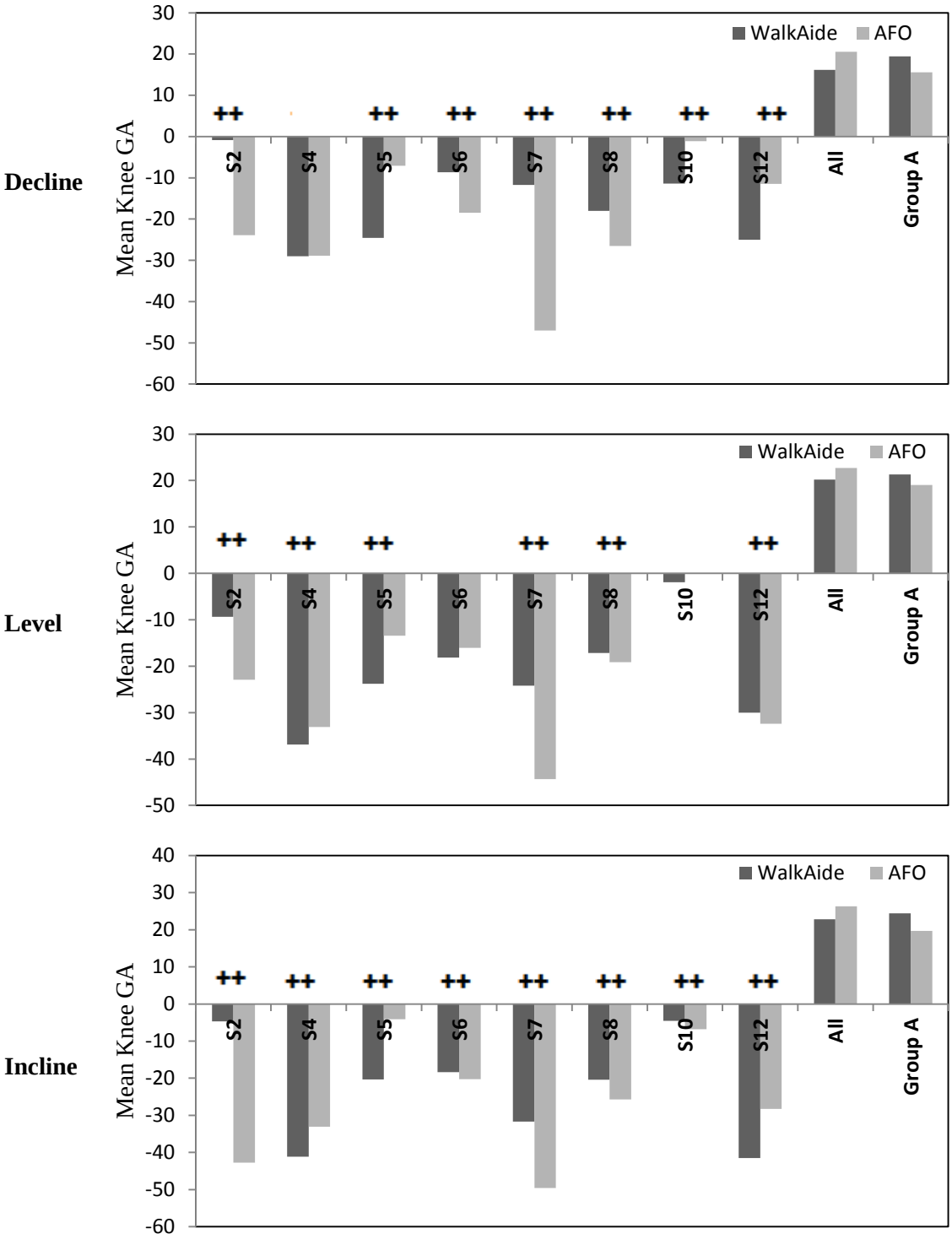

Figure C2: Individual GA for mean knee motion during TS contrasting treatments for declined (top), level (middle), and inclined (bottom) walking.
+, ++ denotes a statistically significant difference (0.05 and 0.01 levels, respectively)

IS minimum and mean ankle motion:

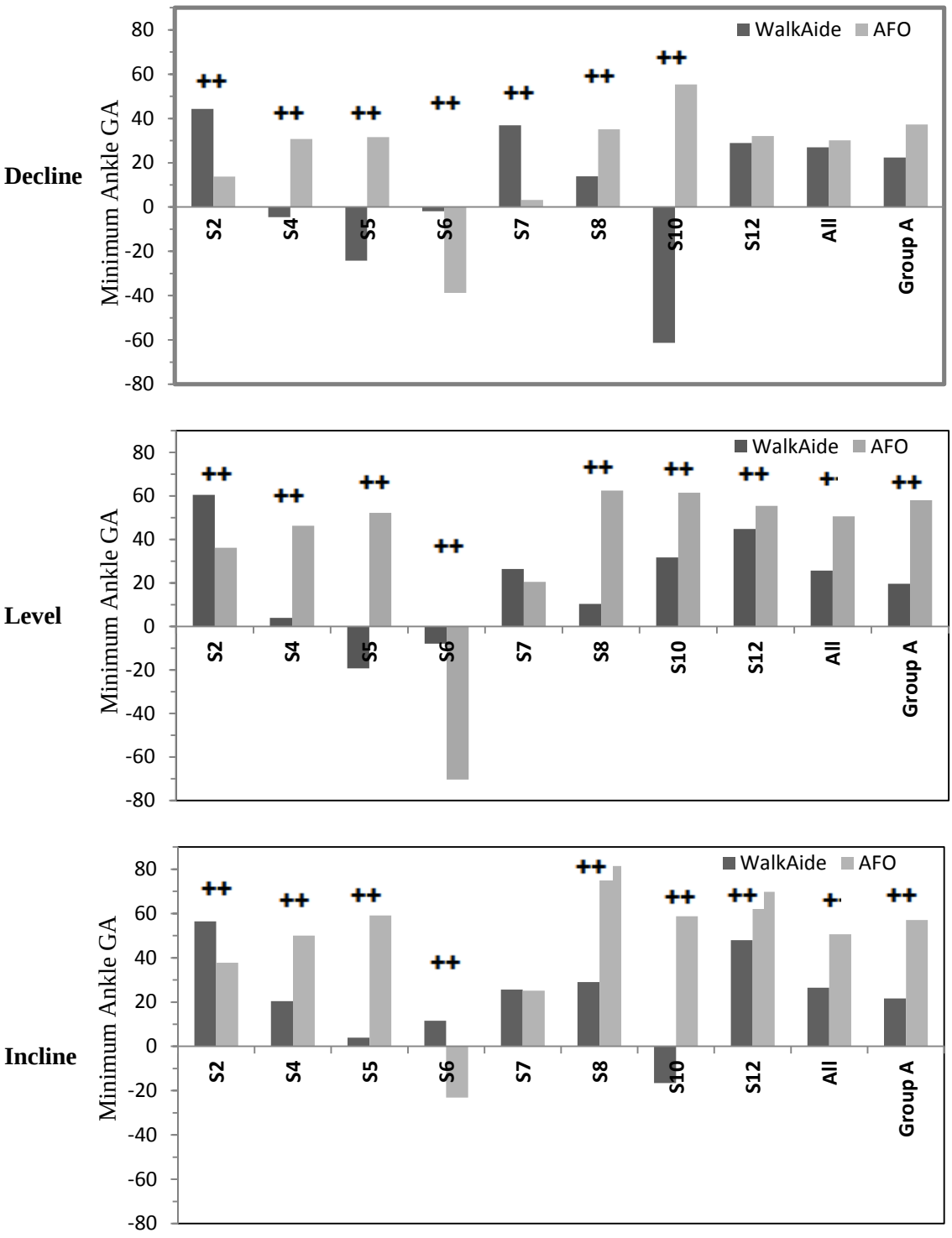

Figure C3: Individual GA for minimum ankle motion during IS contrasting treatments for declined (top), level (middle), and inclined (bottom) walking.

+, ++ denotes a statistically significant difference (0.05 and 0.01 levels, respectively)

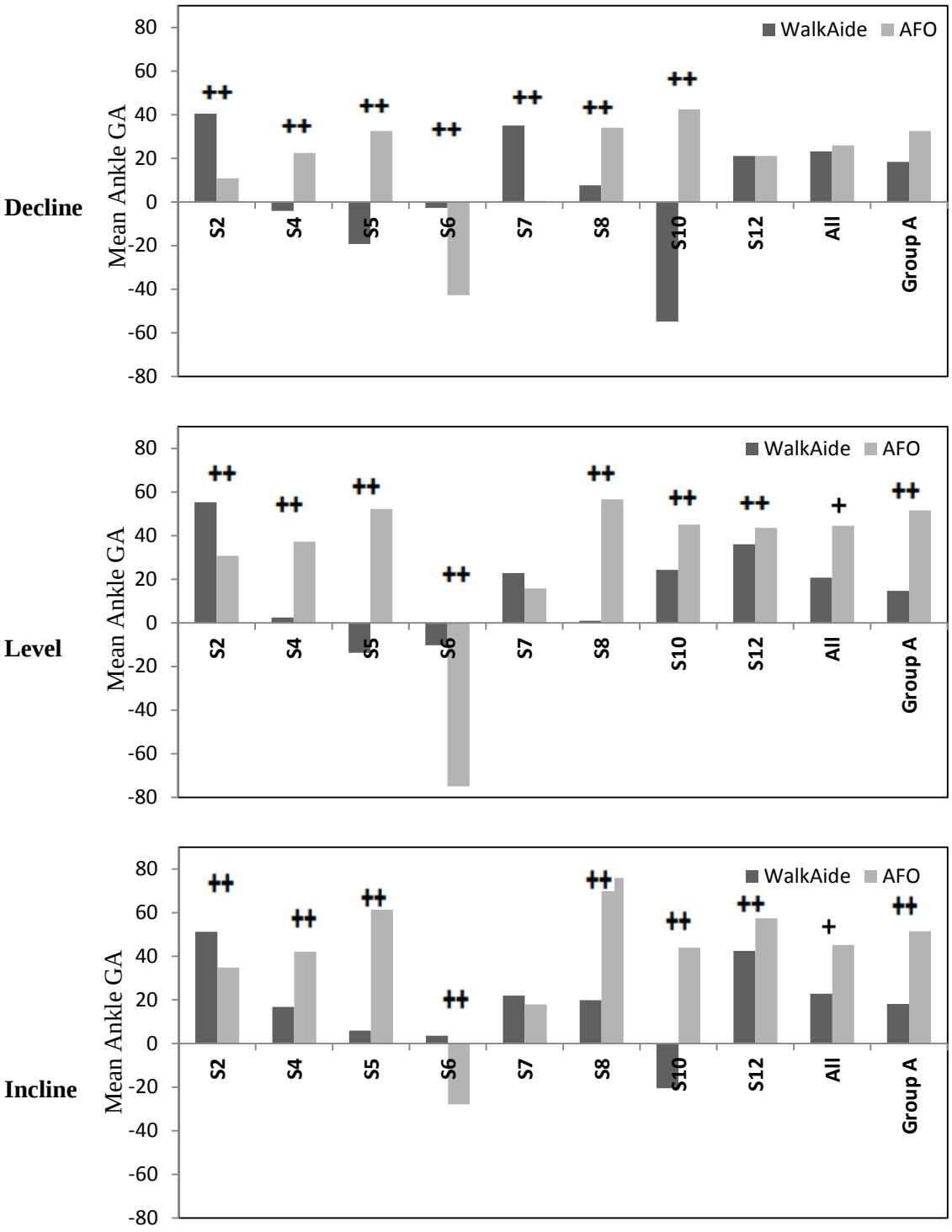

Figure C4: Individual GA for mean ankle motion during IS contrasting treatments for declined (top), level (middle), and inclined (bottom) walking.
+, ++ denotes a statistically significant difference (0.05 and 0.01 levels, respectively)

MS mean and maximum ankle motion:

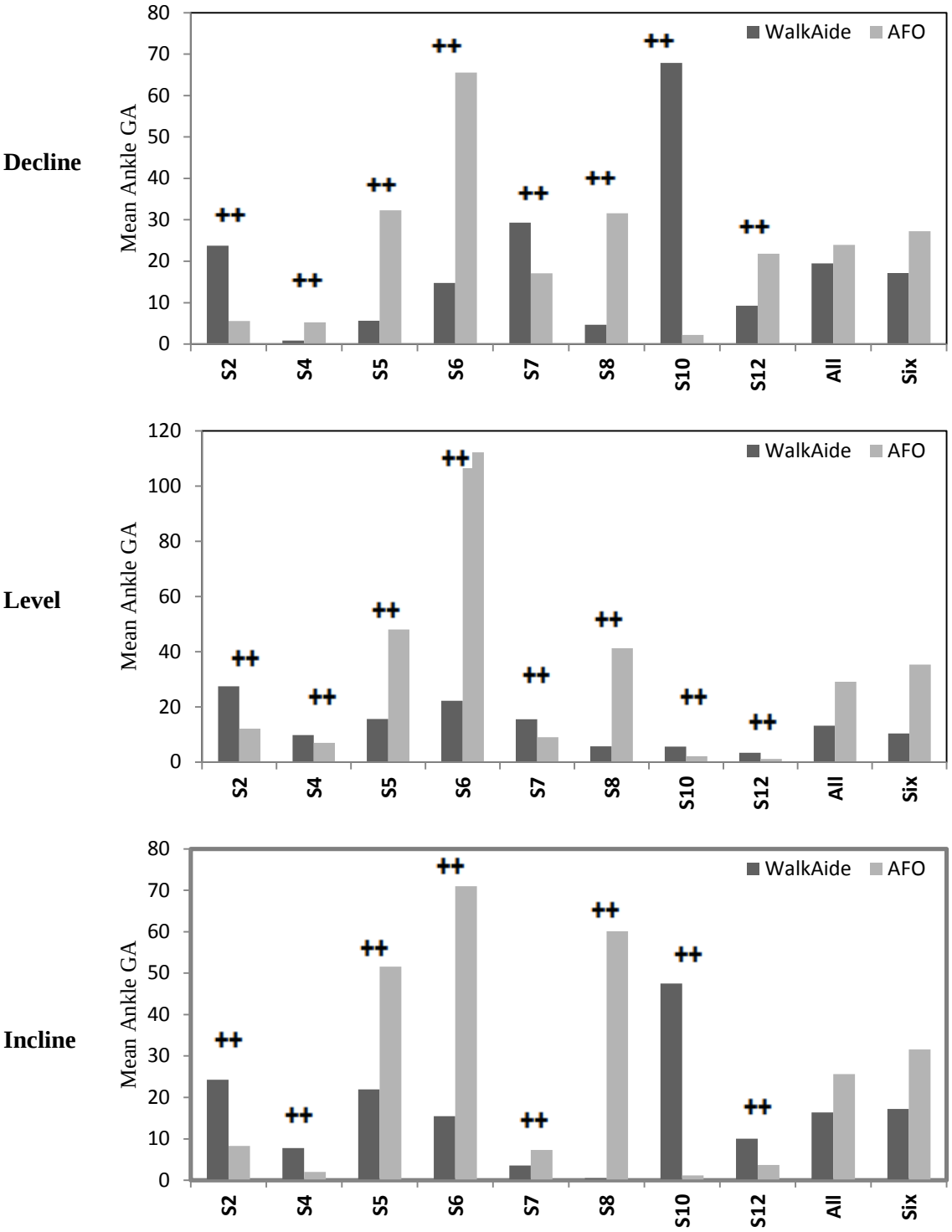

Figure C5: Individual GA for mean ankle motion during MS contrasting treatments for declined (top), level (middle), and inclined (bottom) walking.
+, ++ denotes a statistically significant difference (0.05 and 0.01 levels, respectively)

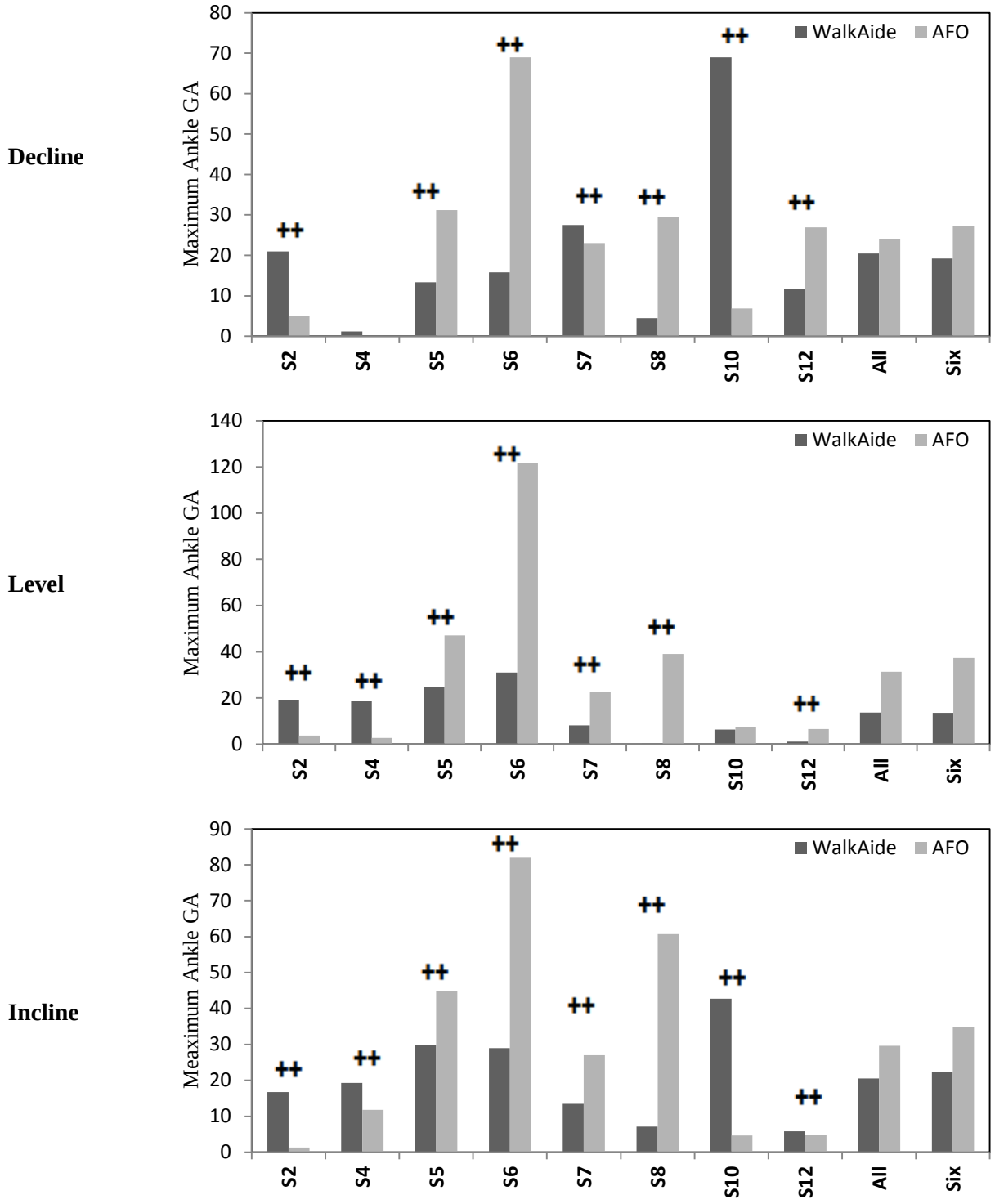

Figure C6: Individual GA for maximum ankle motion during MS contrasting treatments for declined (top), level (middle), and inclined (bottom) walking.

+, ++ denotes a statistically significant difference (0.05 and 0.01 levels, respectively)

TS mean and maximum ankle motion:

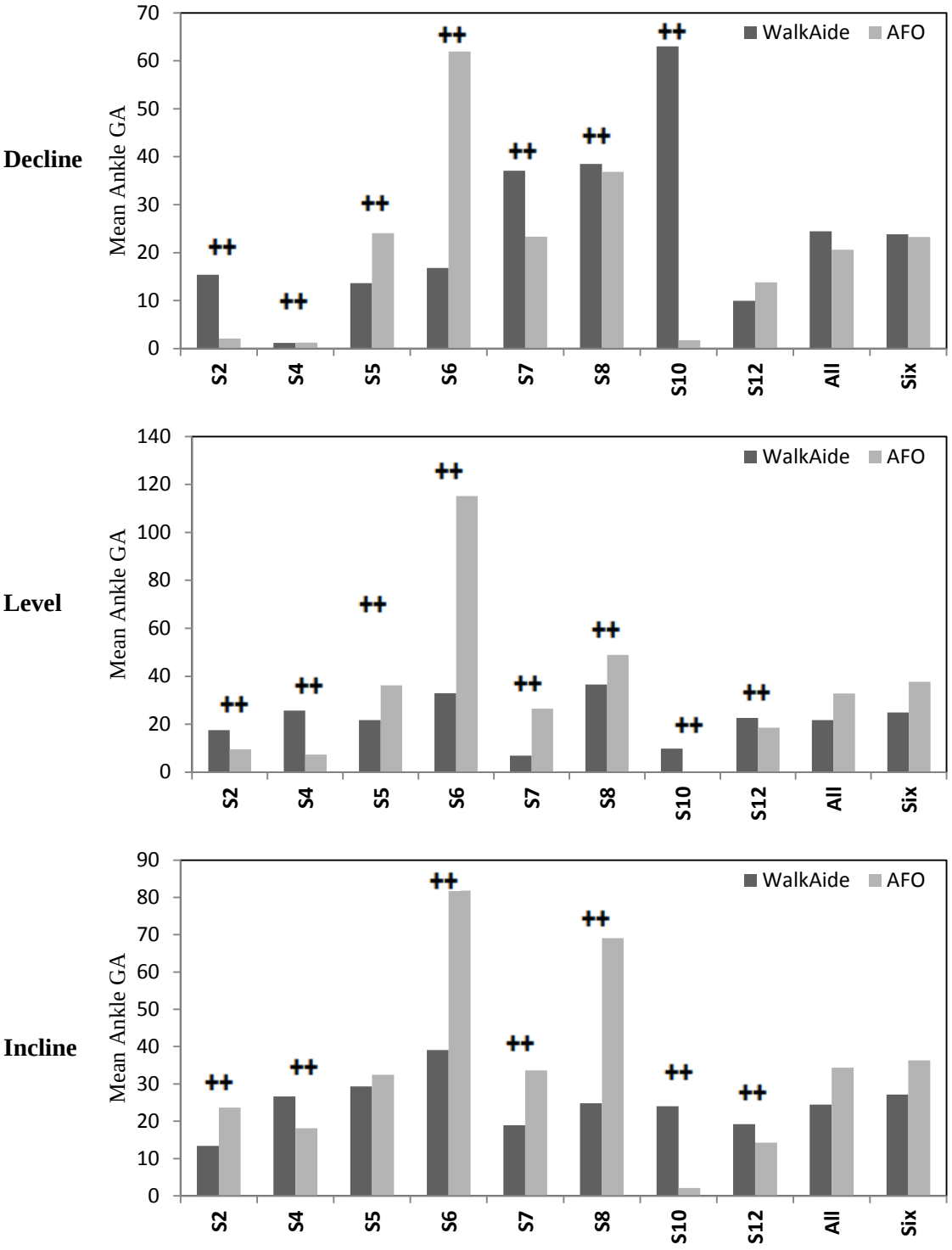

Figure C7: Individual GA for mean ankle motion during TS contrasting treatments for declined (top), level (middle), and inclined (bottom) walking.
+, ++ denotes a statistically significant difference (0.05 and 0.01 levels, respectively)

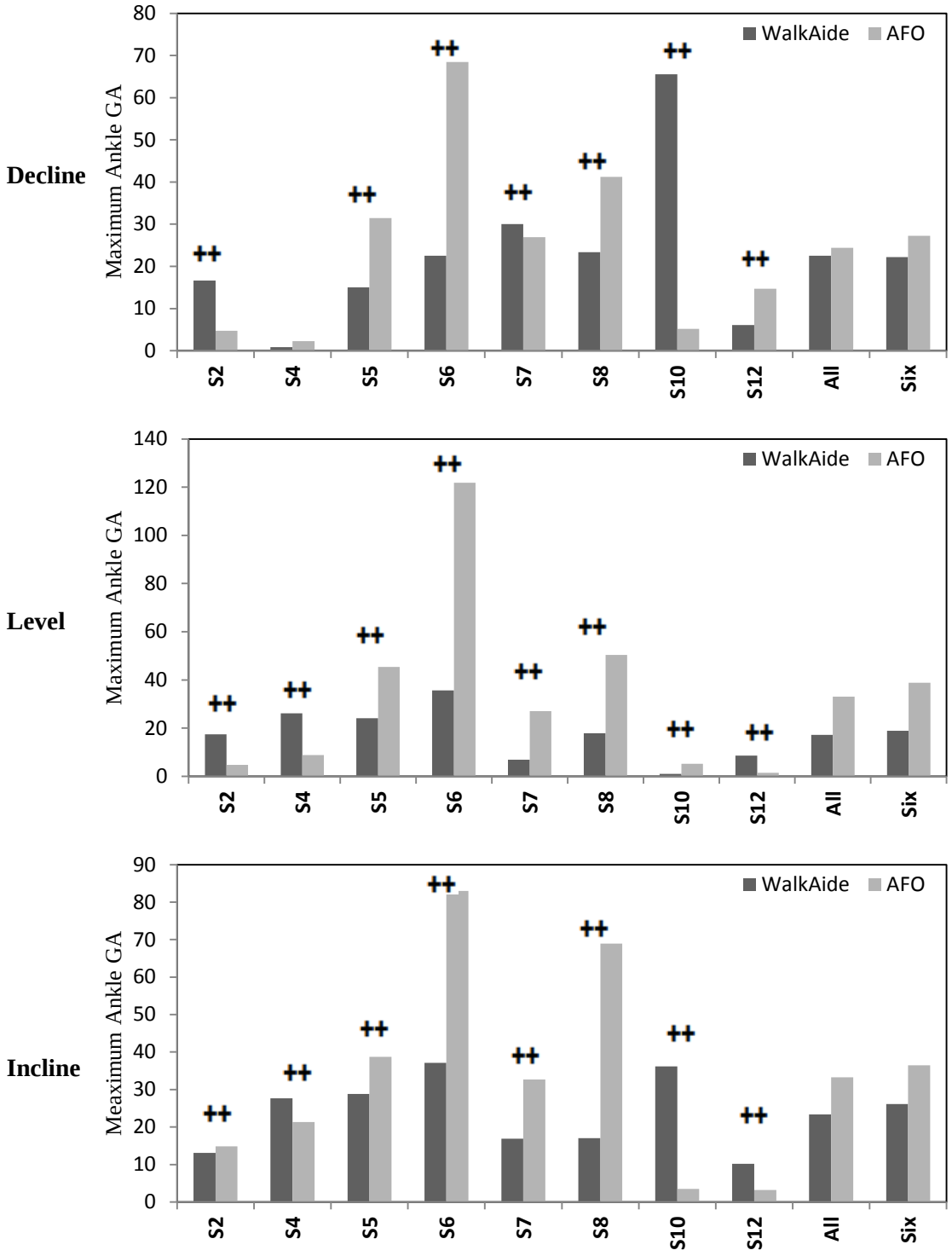

Figure C8: Individual GA for maximum ankle motion during TS contrasting treatments for declined (top), level (middle), and inclined (bottom) walking.

+, ++ denotes a statistically significant difference (0.05 and 0.01 levels, respectively

APPENDIX D: F-scan insole versus WalkAide heel sensor

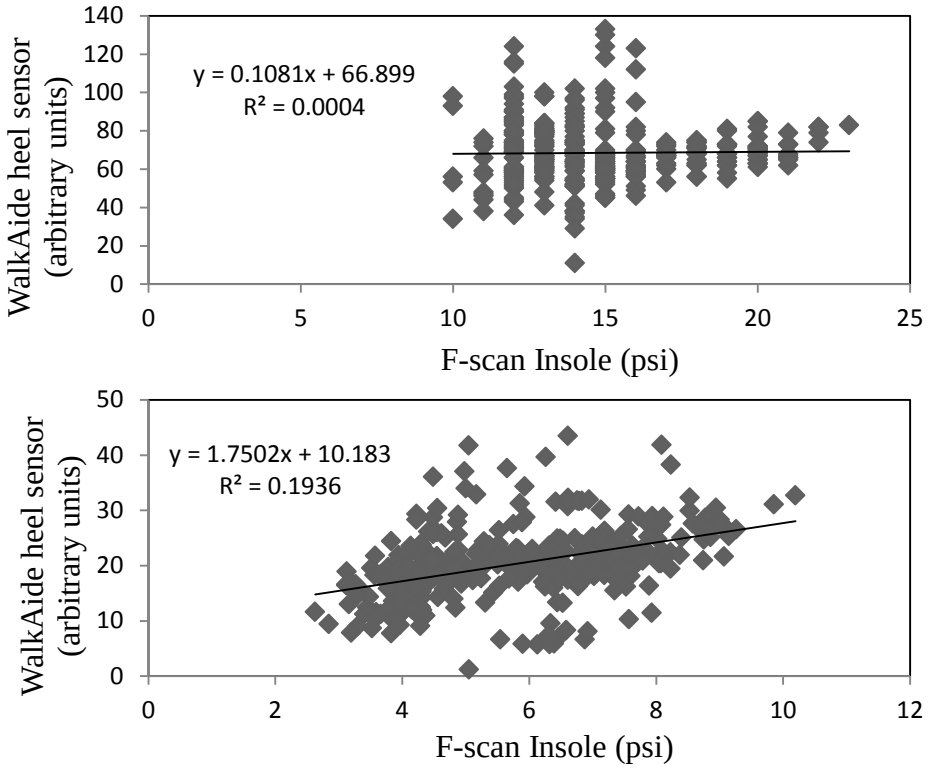

Figure D1: Maximum (top) and mean (bottom) F-scan heel box plantar pressure versus WalkAide heel sensor loading data.

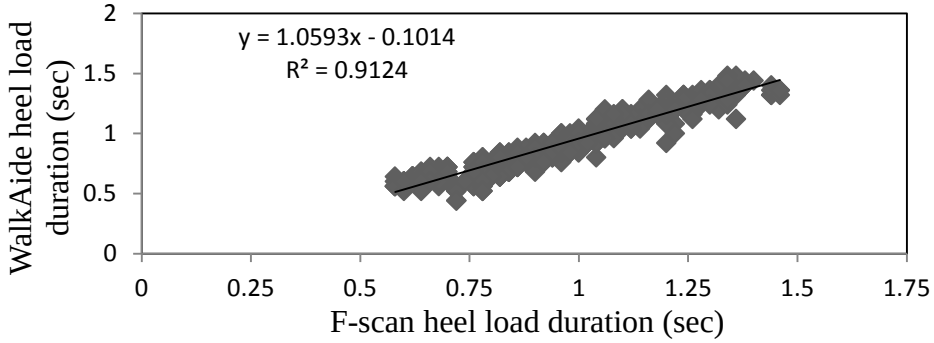

Figure D2: F-scan heel load duration versus WalkAide heel sensor load duration.

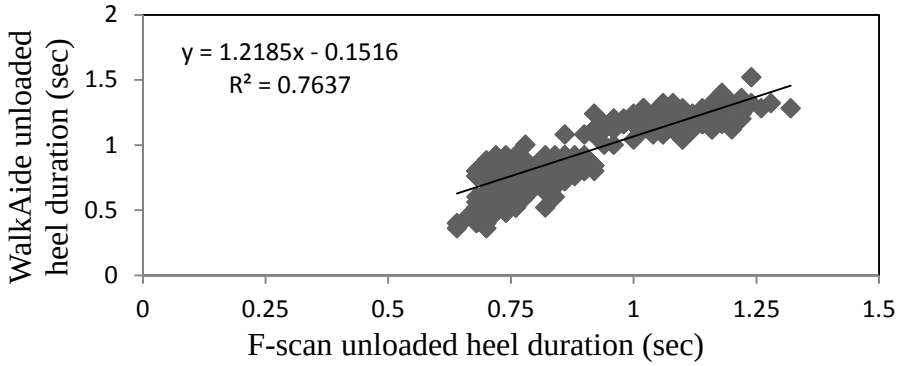

Figure D3: F-scan unloaded heel duration versus WalkAide heel sensor unloading duration.