**THE EFFECTS OF LIMB LOAD ON MUSCLE ACTIVITY DURING ISOLATED
REFLEX TESTING AND WALKING POST-STROKE**

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Abstract

Amy West Hardee: The Effects of Limb Load on Muscle Activity During Isolated
Reflex Testing and Walking Post-Stroke
(Under the direction of Michael Lewek and Richard Goldberg)

Gait impairments are common after stroke, and are most often characterized by difficulty flexing the leg during swing phase. Limb load, or the weight born through the leg, can influence extensor muscle excitability and may be responsible for this 'stiff-legged gait pattern'. We hypothesized that there would be a direct relationship between limb load mediated knee extensor excitability during an isolated setting and during late stance/early swing of gait. Nine subjects had electromyography and kinematics recorded during walking with 0, 20, and 40% body weight support. I also collected quadriceps reflex excitability with and without an imposed limb load in an isolated setting. No significant relationship was observed between limb load and quadriceps muscle activity during reflex testing or walking. Limb load did not influence quadriceps activity suggesting that limb load sensors, which are believed to reside in the plantarflexor muscles, may act locally rather than globally throughout the limb.

This thesis is dedicated to my husband, Thomas, whose love and positivity kept me motivated these past two years, to my family for being my constant cheerleaders, and to Lucy for always being there as my sounding board and ready to talk through my "so what could this mean?" kind of questions.

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And finally, I would like to thank all the subjects who volunteered to participate in this study.

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List of Abbreviations and Symbols

ANOVA	Analysis of Variance
BWS	Body Weight Support
BWSTT	Body Weight Support Treadmill Training
CNS	Central Nervous System
COM	Center of Mass
CPG	Central Pattern Generator
EMG	Electromyography
IRB	Institutional Review Board
MG	Medial Gastrocnemius
MVIC	Maximum Voluntary Isometric Contraction
Nm	Newton meter
RF	Rectus Femoris
SCI	Spinal Cord Injury
TA	Tibialis Anterior
VL	Vastus Lateralis
VM	Vastus Medialis

1.0   Introduction and Background

The purpose of this study is to examine the effects of limb load on knee extensor excitability during gait in patients who have experienced a stroke. We are also interested in observing muscle excitability of the knee extensors during an isolated reflex testing situation.

1.1   Stroke

Stroke is the third leading cause of death in America following diseases of the heart and cancer. Out of the 795,000 Americans that have experienced a new or recurring stroke each year, about 137,000 die from complications. For those that survive, however, stroke is one of the leading causes of long term disability in adults. Dealing with the initial care and subsequent long-term disability required an estimated direct and indirect cost of $68.9 billion in 2009 (AHA 2009).

The primary causes of stroke include the blockage or bursting of an artery, which carries oxygen and nutrients to the brain. When the artery is disrupted, part of the brain no longer receives blood and oxygen and begins to die. The effects of a stroke depend on the location of the obstruction and the extent of the brain tissue affected.   Each hemisphere has unique functions, including control of the contralateral side of the body. As a result, motor effects depend on the side and location of the lesion. For example, if a stroke occurs in the brain's left side, the patient may experience paralysis of the right side of the body and the left side of the face, memory loss, or speech and language problems. If the

stroke occurs on the right side of the brain, the patient may experience paralysis of the left side of the body and right side of the face, memory loss, and vision problems.

The length of time to recover from a stroke depends on its severity. 15%-30% of those experiencing a stroke are permanently disabled (AHA 2009). According to the National Heart, Blood, and Lung Institute's Framingham Heart Study, among ischemic stroke survivors who were older than 65 years of age, 50% had hemiparesis, 30% were unable to walk without some assistance, 26% were dependent in ADL (activities of daily living), and 19% had aphasia (AHA 2009).

1.2   Gait

1.2.1 The 'Normal Gait' Cycle

The human gait cycle consists of two major phases: stance and swing. The stance phase begins when the foot makes contact with the ground (termed 'heel strike') and ends when the same limb leaves the ground (termed 'toe off'). During normal gait, heel strike is the point at which the heel touches the ground, and toe off occurs when the big toe leaves the ground (Figure 1). During swing, the foot is completely airborne and during stance at least part of the foot is contacting the ground. There are two periods of double support during each gait cycle, in which both limbs are in contact with the ground simultaneously.

Figure 1: The Gait Cycle
http://me.queensu.ca/people/deluzio/GaitAnalysis.php
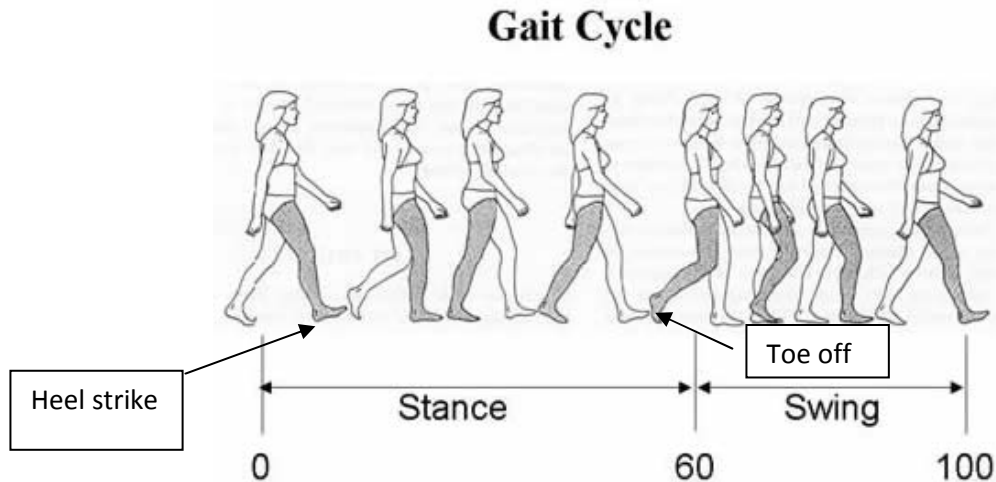
During normal gait, the knee must flex and extend to allow for shock absorption during stance and foot clearance during swing. Specifically, the knee flexes in preswing to approximately 35° and then to 60° in mid-swing (Wheeless 2007). The vasti become active in the terminal swing phase and remain active until the early midstance phase (Figure 2). Normally, the <u>absence</u> of quadriceps activity in the preswing phase is important to allow the knee to flex 35° before toe-off (Wheeless 2007). Rapid flexion of the knee prior to toe-off has been correlated with the amount of knee flexion achieved during swing phase (Goldberg, Ounpuu et al. 2003). The rectus femoris (RF), a hip flexor and knee extensor, has a burst of activity during the preswing and initial swing phases. As a hip flexor, the RF increases knee flexion during swing, which is largely a passive movement, but as a knee extensor, can impair the ability to adequately flex the knee. Activity of the hamstrings begins in the midswing phase and ends

in the loading response phase and therefore has little influence on knee flexion during swing.

Figure 2:   Quadriceps Anatomy
http://msjensen.cehd.umn.edu/webanatomy/muscular/musc_thigh_1_m.htm
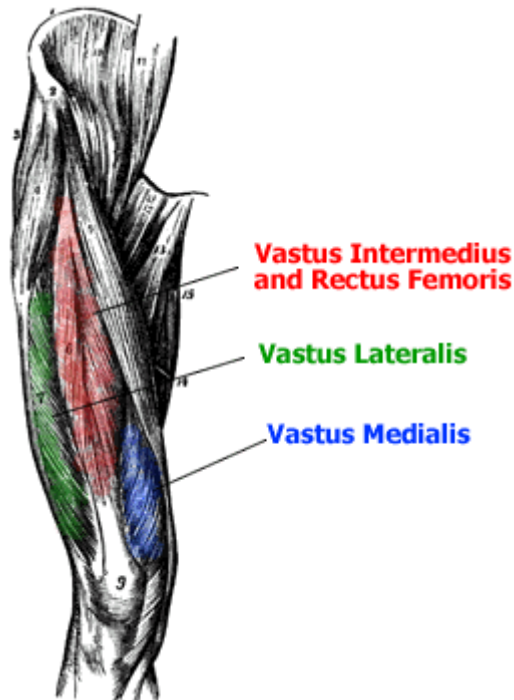
1.2.2 The Effect of Stroke on Gait

After stroke, muscle activity patterns and thus kinematics during walking are often altered. This can result in decreased gait speed, postural instability, reduced energy efficiency, and kinematic abnormalities (Lewek, Hornby et al. 2007).

One of the most common forms of abnormal gait is termed "spastic paretic stiff-legged gait". Stiff-legged gait is marked by a lack of normal knee flexion during the swing phase of walking (Figures 3 and 4) and is a common disability

associated with stroke and other diagnoses that affect the brain or spinal cord.

Reduced knee flexion in swing can contribute to toe drag and creates a very

large moment of inertia that significantly increases the amount of mechanical

energy required to walk (Kerrigan, Gronley et al. 1991). Patients with stiff-legged

gait typically try to compensate for a lack of knee flexion with hip hiking,

circumduction, or toe-drag, which require the patient to exert more energy, and

may lead to additional biomechanical injuries.

Figure 3: Knee Flexion Angle
http://me.queensu.ca/people/deluzio/BiomedicalDataAnalysisTechniques.php
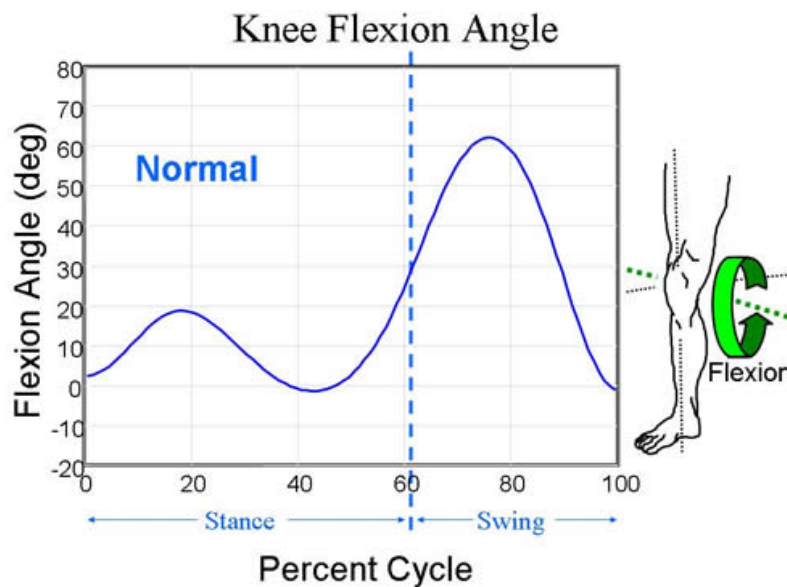
**Figure 4: Paretic vs. Non-Paretic Knee Flexion Angle
(Lewek, Hornby et al. 2007)**
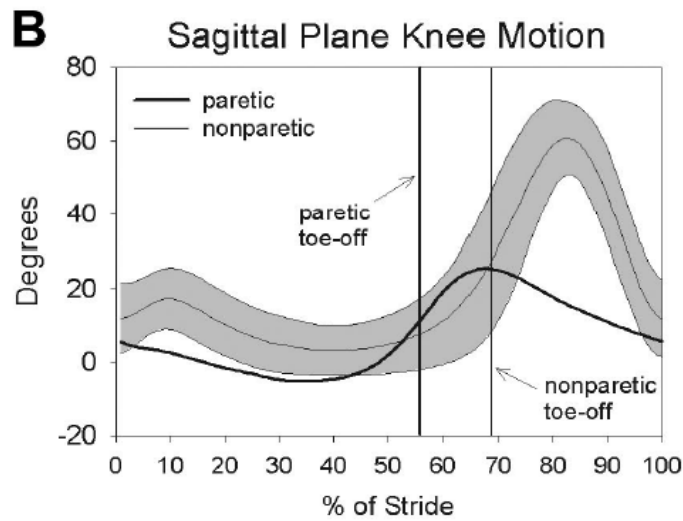
1.3 Treatment of Stiff-Legged Gait

1.3.1 Muscle Studies

Although hip and ankle impairments can contribute to the presence of stiff-legged gait, research on this condition has focused on the impact of knee musculature. Treatment of stiff-legged gait is primarily aimed at improving knee flexion during swing. Often times, the focus for reduced knee flexion is the spasticity of the quadriceps during late stance/early swing. Various studies have shown the influence of knee extensor (rectus femoris, vastus intermedius, vastus lateralis, and vastus medialis) and flexor (hamstrings) muscle activity on knee flexion during swing phase. In 1997, Kerrigan et al hypothesized that hip flexor action or torque may be especially important in allowing the knee to flex (Casey Kerrigan, R et al. 1998). They found that on average, increasing hip flexor torque resulted in a significant increase in peak knee flexion. They concluded that increasing the hip flexor torque was the clinical equivalent of increasing hip flexor action during swing. These findings confirmed Kerrigan's earlier 1991 hypothesis

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that "spastic paretic stiff-legged gait can be attributed to multiple impairments besides spastic quadriceps activity" (Kerrigan, Gronley et al. 1991). In 2001, Kerrigan et al hypothesized that subjects with spastic paretic stiff-legged gait have altered kinetics not only about the knee, but also about the hip and ankle (Kerrigan, Karvosky et al. 2001). After collecting joint kinetic data of 20 subjects with spastic paretic stiff-legged gait and comparing the data with that of 20 control subjects the authors observed increases in peak external hip flexion torque in stance, hip-power generation in loading response, knee-extension torque in midstance, ankle-dorsiflexion torque, and ankle-power absorption in stance (Kerrigan, Karvosky et al. 2001). These findings supported the likelihood of multiple mechanisms being responsible for reduced knee flexion in swing. Also in 2001, Kerrigan et al found that when subjects were asked to walk on their toes compared with normal heel-toe walking, there was a 17 degree reduction in peak knee flexion. This suggested that ankle function is related to knee flexion in swing. These previous studies have suggested that there are kinetic mechanisms taking place throughout the lower limb that can lead to stiff-legged gait that are not only confined to the knee, but could be related to the ankle and hip flexors as well.

1.3.2 Body Weight Support

Treadmill training in conjunction with body weight support (BWSTT) has become a common rehabilitation strategy to restore locomotor function in individuals post-stroke. BWSTT involves the use of a harness system to support a percentage of a patient's body weight to unload the lower extremities while the patient is walking on the treadmill (Finch, Barbeau et al. 1991; Barbeau and

Visintin 2003). From a motor learning perspective, BWSTT may be successful because complete gait cycles can be practiced with many repetitions (massed practice) instead of focusing on single elements. While there is some suggestion that BWSTT is better than training without body weight support, the underlying mechanism is not clear (Finch, Barbeau et al. 1991; Barbeau and Visintin 2003). The clinical rationale of this study is to elucidate the mechanism underlying the benefit of BWSTT.

Previous studies have suggested that after training with a body weight support system, subjects displayed greater increases in terminal stance hip extension angle and hip flexion power (Visintin, Barbeau et al. 1998; Hesse, Konrad et al. 1999; Barbeau and Visintin 2003; Moseley, Stark et al. 2005). One study recruited 100 subjects following stroke and assigned 50 to receive locomotor treadmill training with body weight support and 50 subjects performed this training without body weight support. The researchers took measures related to walking speed and endurance, functional balance, and motor recovery. After 6 weeks of training, the body weight support group scored significantly higher in all clinical outcomes. They also were able to show a significant improvement with BWSTT in those stroke subjects who were more severely impaired and those who were older (65-85 years) (Barbeau and Visintin 2003). In 1997, Hesse et al, had subjects who had a stroke walk on the treadmill with 15%, 30%, 45%, and 60% body weight support at constant walking speed. Their results showed that with body weight support, the amount of time that both limbs were in contact with the ground decreased, the relative single stance period of

the affected limb increased, and the activity of the gluteus increased. Their findings suggested that body weight support did not lead to an inefficient gait cycle (Hesse, Konrad et al. 1999).   In 1998, Visintin et al demonstrated that participants who trained with body weight support scored significantly higher than a control group (without body weight support) for functional balance, motor recovery, overground walking speed, and overground walking endurance. They also found that 3 months after training, the body weight support group continued to have significantly higher scores for overground walking speed and motor recovery (Visintin, Barbeau et al. 1998). This suggested that body weight support treadmill training could have promising long term effects for stroke patients. While these studies demonstrated that subjects benefitted from body weight support during training, the mechanism underlying the benefit is still unclear. A Cochrane review published in 2005 to assess the effectiveness of treadmill training and body weight support in the treatment of walking after stroke, however, showed inconclusive data to support the use of BWSTT (Moseley, Stark et al. 2005). So while some studies have shown positive effects after BWSTT, some have not. Understanding the mechanism underlying muscle activity responses to BWSTT may help elucidate who would respond well to such training and who would not.

1.4   The Effects of Limb Load on Gait

1.4.1 The Proprioceptive System
In the human, the proprioceptive system is made up of a complex blend of voluntary and automatic control. In order for the system to function, the brain must have rapid, precise information regarding the position of the body and each

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limb in space, the length and rate of contraction of each muscle, and the tension which each muscle is applying at its point of insertion (Lin 1990). Muscle spindles, golgi tendon organs, and sensory receptors in joint, subcutaneous tissue, and skin are the primary sensory inputs to the system (Kandel 2000).

Muscles are innervated by alpha and gamma motor neurons. The alpha motor neurons are large lower motor neurons of the brainstem and spinal cord. These neurons innervate extrafusal muscle fibers of skeletal muscle (Hammer 2007). Their cell bodies are found in the central nervous system but are considered to be part of the peripheral nervous system because their axons extend into the body's periphery to innervate skeletal muscles. Alpha motor neurons and the muscle fibers they innervate are called the motor unit. The gamma motor neurons innervate the intrafusal fibers within the muscle spindles (Hammer 2007). These motor neurons set the tension of the spindle cell and control the sensitivity of the spindle receptors.

There are several types of sensory (afferent) fibers. Muscle spindles are innervated by group Ia and II fibers. The Ia fibers, which provide important information about the rate of change of a muscle's length, are the largest and fastest fibers (Hammer 2007). They rapidly adapt because as soon as the muscle stops changing length, the Ia stops changing its firing frequency and adapts to the new length. The group II fibers, provide information about the muscle's current length and are constantly updating the CNS as the length of the muscle is changed (Hammer 2007).

The Ib afferents arise from the golgi tendon organs and simultaneously provide information about the tension in the muscle and the rate of change of tension. During many circumstances, these afferents selectively inhibit the alpha motor neurons of the agonist muscles while facilitating those of the antagonist muscles (Hammer 2007). Golgi tendon organs can be important signals for regulating the step cycle. Electrical stimulation of the afferents from these receptors prolongs the stance phase and often delays the onset of swing until the stimulus has terminated (Kandel 2000). With both the Ia and Ib afferents active during stance, the golgi tendon organs provide a measure of the load carried by the leg. The excitatory action of the golgi tendon organs on extensor motor neurons during walking is opposite to their inhibitory action when locomotor activity is not being generated (Kandel 2000). The consequence of this is that the swing phase will not be initiated until the leg is unloaded and the forces exerted by extensor muscles are low. These golgi tendon organs also contribute greatly to the generation of burst activity in extensor motor neurons (Kandel 2000).

1.4.2 Movements Produced by the Nervous System

There are many types of movements produced by the nervous system. These range from very simple movements, such as reflexes, to complex movements such as rhythmic motor patterns. Reflexes, such as a knee jerk reaction, are involuntary and only need a stimulus strong enough to activate the relevant sensory input pathway. The simplest of the reflexes is the monosynaptic reflex which occurs at a single level of the spinal cord. For example, a patellar tendon tap stretches the muscle spindles causing the Ia afferents to send a

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signal into the spinal cord, synapsing monosynaptically with a quadriceps alpha motor neuron, which produces a quick unsustained contraction of the quadriceps (Kandel 2000). Through reciprocal inhibition, the antagonist muscle, in this case the hamstrings, would be inhibited.

Rhythmic patterns, such as walking, however, are much more complex. Various work and studies have shown that rhythmic motor patterns are produced by neural networks called central pattern generators.

1.4.3 Central Pattern Generators

Central pattern generators (CPGs) are neural networks in the spinal cord that can produce rhythmic patterned outputs such as walking, breathing, flying, or swimming without rhythmic sensory or central input. The basic pattern produced by a CPG is usually modified by sensory information from peripheral receptors or signals from other regions of the central nervous system (Kuo 2002). The generation of rhythmic motor activity by CPGs depends on the cellular properties of individual nerve cells in the network, the properties of the synaptic junctions between neurons, and the pattern of interconnections between neurons (Kuo 2002).

The most common CPGs produce a complex temporal pattern of activation of different groups of motor neurons. There are many different mechanisms that can affect the sequencing of motor patterns. The simplest mechanism is mutual inhibition in which interneurons that fire out of phase with each other are usually reciprocally coupled by inhibitory connections (Marder and Bucher 2001). Mutual excitation is also another important mechanism which allows synchronous firing

of a group of neurons. These are important when it is necessary to rapidly generate a high intensity burst within a group of neurons.

Marder and Calabrese showed that the locust nervous system, when isolated from the animal, could produce rhythmic output resembling that observed during flight, which was some initial evidence that rhythmic motor patterns are centrally generated (Calabrese 1996). Later work showed that in a wide variety of animals, nervous systems isolated from sensory feedback could produce rhythmic outputs resembling those observed during rhythmic motor pattern production.

1.4.4 Previous Studies Examining the Neurophysiological Effect of Loading in Cats

Somatosensory input from the receptors of muscle and skin, input from the vestibular apparatus, and visual input are important types of sensory information that are used to regulate stepping. Many studies have demonstrated that somatosensory input from the limbs regulate the step cycle in spinal and decerebrate cats. When these cats are placed on a treadmill, they can use the somatosensory input to the CPG to attain a rate of stepping that matches the speed of the treadmill. Specifically, afferent input regulates the duration of the stance phase. It has been shown that when the stepping rate increases, the stance duration decreases. This suggested that some form of sensory input signals the end of stance and the initiation of swing.

Many studies have been done to analyze the proprioreceptors in the walking systems of cats. Reflex pathways have been identified that regulate the timing of the transition from stance to swing and control the magnitude of ongoing motor

neuronal activity. In 1978, Grillner and Rossignol had cats walk on a treadmill

with their hind legs. They held one limb and allowed the other to continue to

walk. They found that if the held limb was slowly brought backwards to a

particular point, the limb flexes and continues walking. The hip position at which

the leg lifts off during this reaction was very close to the hip angle at the initiation

of the swing phase during walking (Grillner and Rossignol 1978). They

concluded that the fixed limb tended to initiate lift-off during the midstance or the

midswing of the contralateral limb cycle and that hip position and the

contralateral step cycle phase are two important factors in determining the

initiation of swing in one leg. In 1980, Duysens and Pearson performed an

experiment in which they rigidly fixed the left hindlimb in one position after

denervating almost all the leg muscles except the ankle extensor and ankle

flexor. Rhythmic alternating contractions of the isolated ankle flexor and

extensor occurred in the fixed hindleg during periods of walking in the other three

intact limbs (Duysens and Pearson 1980). They then stretched the triceps surae

so to increase the force of the active contractions to beyond 4 kg and found that

these rhythmic contractions disappeared. The contractions in the ankle flexor

and extensor only returned after the force in the stretched triceps fell below 4 kg.

They also found that inhibition of the locomotor rhythm could be produced by

clamping the ankle joint in a flexed position so as to stretch the ankle extensor.

This suggested that the triceps surae proprioceptors signaling the presence of

loading in the hindlimb extensor muscles inhibit the generation of hindlimb

flexion. They concluded that in normal walking this mechanism could be of major

importance during stance to prevent the initiation of the swing phase when hindlimb extension is needed to fully support the animal. Therefore, the unloading of leg extensor muscles may be necessary in the initiation of swing.

In 1995, Pearson found that during locomotor activity, the influence of feedback from the golgi tendon organs in extensor muscles onto extensor motor neurons is reversed from inhibition to excitation and that the excitatory action of tendon organs during stance ensures that stance is maintained while extensor muscles are loaded. It was concluded that extensor activity may be regulated according to the load carried by the leg (Pearson 1995).

There were various studies that also showed that extensor group I afferents have access to central pattern generators and may be of importance in the reflex regulation of stepping. These experiments showed that utilizing natural stimulation of muscle receptors demonstrate that the group I input to the rhythm generators arises mainly from golgi tendon organ Ib afferents. Therefore, an increased load on limb extensors during the stance phase would enhance and prolong extensor activity while simultaneously delaying the transition to the swing phase of the step cycle (Kriellaars, Brownstone et al. 1994; Conway, Hultborn et al. 1987). Similarly, in 1996 Hiebert et al. tested the hypothesis that muscle spindle afferents signaling the length of hind-leg flexor muscles are involved in terminating extensor activity and initiating flexion during walking. They found that stretch of the flexor muscles during the stance phase reset locomotor rhythm to flexion ipsilaterally and extension contralaterally. They concluded that as the flexor muscles lengthened during the stance phase of

gait, their spindle afferents acted to inhibit the spinal center generating extensor activity and facilitating the initiation of swing (Hiebert, Whelan et al. 1996).

Finally, in 2001, Lam and Pierson examined the influence of proprioceptive input from hip flexor muscles on hip flexor activity during the swing phase of walking in the decerebrate cat (Lam and Pearson 2001). When the hip was manually assisted into flexion, there was a reduction in hip flexor burst activity. When hip flexion was manually resisted or blocked during swing, the duration and amplitude of hip flexor activity increased. Because of these results, they concluded that during swing, feedback from hip flexor muscle afferents, particular those from the sartorius muscles enhanced flexor activity (Lam and Pearson 2001). They also found that if they delayed the onset of flexor activity in the contralateral hindlimb, blocking hip flexion often resulted in the prolongation of ipsilateral flexor activity for long periods of time.

These studies have all suggested that the spinal cord in the cat contains a central pattern generator capable of generating alternating flexion and extension activity for stereotypical limb movements such as walking. These CPGs can operate without any peripheral sensory feedback or supraspinal input. But most importantly, it has also been shown that apart from hip position, unloading of the extensor muscles in the limb may also be an important sensory signal involved in initiating the stance to swing transition. The findings of these studies have led scientists to study similar patterns in humans.

1.4.5 Previous Studies Examining the Neurophysiological Effect of Loading in Humans

In 2000 Pang and Yang chose to test whether the same sensory information that controlled the phases of stepping in the spinal cat also controlled stance and swing phase in a human infant. They studied the forces exerted and the hip angle of the lower limbs in 22 infants while walking supported on a treadmill. They manipulated the hip position and the amount of load experienced by the right limb during stepping. It was found that when the hip was flexed and the load on the limb was high, the stance phase was prolonged and the swing phase was delayed (Pang and Yang 2000). When the hip was extended and the load was low, the stance phase was shortened and swing phase occurred prematurely. Therefore there was an inverse relationship between the hip position and load at the time of swing initiation. This suggested that the same proprioceptive information that is important to cat locomotion, may be critical in humans as well. In 2002, Pang and Yang did a similar study to see if the same factors regulated the initiation of the swing phase in other directions of stepping (Pang and Yang 2002). This time they created the disturbances on the infants by placing a piece of cardboard under the foot and pulling the cardboard in different directions. They measured EMG, vertical ground reaction forces, and hip motion in the frontal and sagittal planes and found that the most potent sensory input to initiate the swing phase depended on the direction of stepping. Low load was always necessary to initiate swing of all directions of walking, but the preferred hip position was always one directly opposite the direction of walking (Pang and Yang 2002).

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These studies also showed compelling evidence for the existence of spinal rhythm generating networks in humans. The fact that human infants at a time prior to the completion of all cortico-spinal connections could produce rhythmic stepping movements, shortly after birth, when held erect over a horizontal surface suggests the some of the basic neuronal circuits of locomotion are developed genetically and these circuits may possibly be entirely within the spinal cord.

1.4.6 Spasticity in the Human Adult

Many individuals who have experienced a stroke often present with a condition referred to as spasticity. Spasticity is defined as a motor disorder characterized by a velocity dependent increase in tonic stretch reflexes with exaggerated tendon jerks, resulting from hyper-excitability of the stretch reflex (Lance 1980). Extensor spasms which consist of prolonged extension of the legs are a significant component of spasticity in patients with spinal cord injury. In 2002, Schmit and Benz performed a study to attempt to identify the role that hip proprioceptors play in triggering an extensor reflex response. They tested 10 subjects by doing a controlled hip extension movement on one leg while holding the knee and ankle in place. They collected joint torques of the hip, knee and ankle as well as EMG signals. EMGs indicated coactivation at the knee and ankle joints, with extensor activity generally outlasting flexor activity (Schmit and Benz 2002). The torque response to imposed hip extension consisted of hip flexion, knee extension and ankle plantar flexion while the torque response to hip flexion was comprised of hip extension, knee flexion and ankle extension. The

authors felt that the difference between the responses to hip flexion and hip extension suggested organized pathways for coordinating leg movements.

Wu et al have also done studies examining extensor spasms. In 2005, they did a study examining the contributions of knee proprioceptors to the origination of extensor spasms (Wu, Hornby et al. 2005). They conducted tests in both the flexed and extended positions and found that larger responses were observed in the EMGs with the hip in the extended position which emphasized the role of the knee and hip proprioceptors in the initiation of extensor spasms. In 2006, Wu and Schmit performed ramp and hold movements on 12 patients with SCI (Wu and Schmit 2006). They imposed an ankle plantarflexion load and released it after the hip reached a targeted position. Reflexes characterized by hip flexion torque, knee extension, and coactivation of ankle flexors and extensors were trigged by ankle load released when the hip was in an extended position. The authors found that the ankle load release enhanced the reflexes triggered by hip extension itself which suggested that ankle load afferents play an important role in spastic reflexes and that reflex pathways associated with ankle load afferents have important implications in the spinal reflex regulation of human movement (Wu and Schmit 2006).

While studies have examined the role of hip flexor afferents, relatively less has been done to study the effects of limb load on muscle function following a stroke. We feel that it is important to analyze the influence of limb load on the muscular response, both during an isolated setting and while walking on a treadmill. Using a body weight support system will allow us to determine the

effect that limb load has on gait. This may also help us discover why some subjects benefit from body weight support training. During body weight support training, our subjects will experience decreased limb load. It is expected that this decreased limb load will cause a decrease in knee extensor excitability. Because it is believed that prolonged knee extensor excitability is a main cause of stiff-legged gait, this decrease in knee extensor excitability should facilitate knee flexion during swing. Increased knee flexion during swing should lead to a more energy efficient gait cycle. If our study proves supports this contention, then the use of BWSTT may be a rehabilitation strategy that should be continued in the future to train subjects to walk more "normally."

1.5 Specific Aims and Hypotheses

Aim 1: To determine the effects of isolated limb loading on reflex excitability of the knee extensor muscles post-stroke.

Hypothesis 1: We hypothesize that increased load at the ankle in individuals with stiff-knee gait will cause increased extensor muscle activity in isolated reflex testing (Figure 5).

Figure 5: Extensor Excitability vs. Ankle Load (Hypothesized)
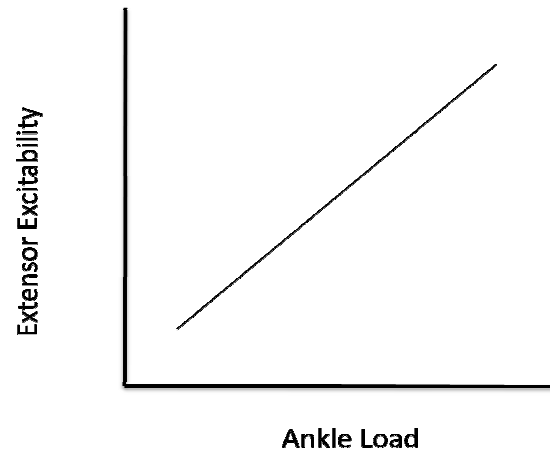
Aim 2: To determine the contribution of limb load on muscle activity during walking in individuals post-stroke.

Hypothesis 2.1: We hypothesize that there will be a direct relationship between the limb load and knee extensor excitability during late stance/early swing of gait (Figure 6).

Figure 6: Extensor Excitability vs. Body Weight Support (Hypothesized)
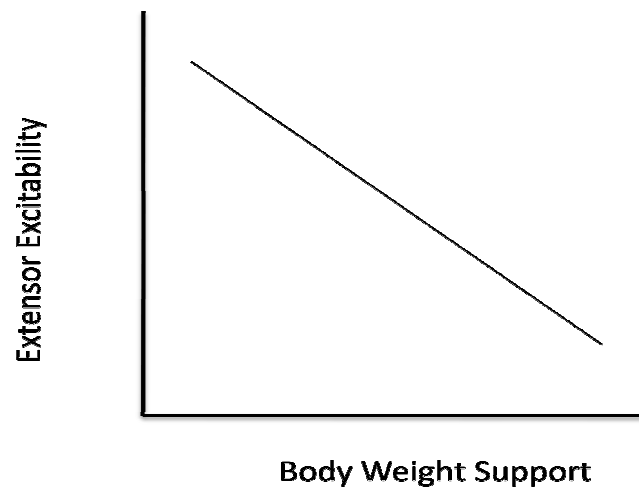
Hypothesis 2.2: We hypothesize that individuals who show greater extensor

excitability with limb load in an isolated situation will exhibit greater knee extensor

activity during gait (Figure 7).

**Figure 7: Relationship Between Gait Extensor Excitability & Isolated Reflex
Extensor Excitability (Hypothesized)**
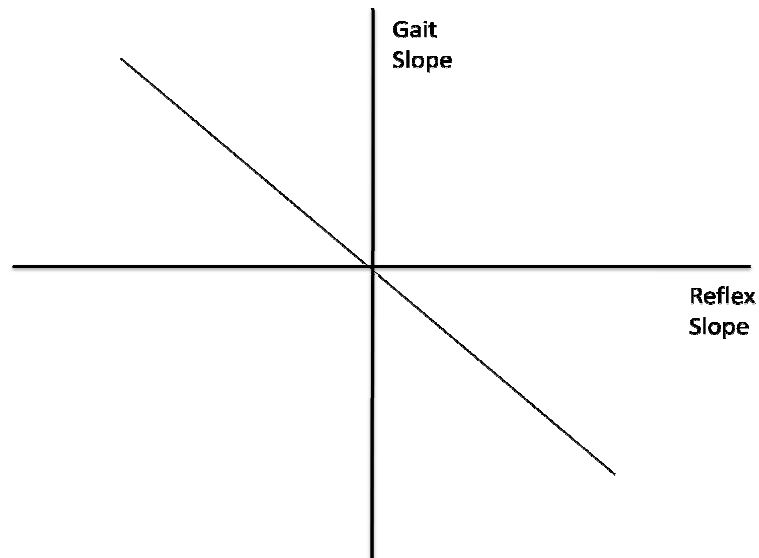
2.0 Methods

2.1 Subjects

Nine subjects who exhibited clinically identifiable stiff-legged gait participated in one testing session that consisted of two parts. The first part consisted of isolated reflex testing while the second was a gait analysis. These nine subjects had an average age of 53±5.5 years (Table 1). Six subjects were male while 3 subjects were female. Six subjects were right side paretic while 3 subjects were left side paretic. The subjects had a mean weight of 81.6±16.9 kg. The mean height was 68.8±4.9 inches (174.7±12.5 cm).Individuals were included if they showed clinical symptoms consistent with ischemic or hemorrhagic unilateral brain lesion resulting in sensory motor dysfunction more than 6 months prior to testing. Individuals had to be able to step independently without assistance while walking on the treadmill.

Table 1: Subject Data

Subject	Gender	Side Tested	Age (yrs)	Weight (kgs)	Height (cms)	Gait Speed (m/s)	Time Since Stroke (yrs)
1	M	R	55	98	180	.85	5
2	F	L	57	93	165	.6	7
3	M	R	55	86	180	1.0	3
4	M	R	60	68	170	1.0	2
5	F	R	51	43	150	.5	8
6	M	L	49	90	188	.55	6
7	F	L	54	83	170	.9	28
8	M	R	57	79	178	.65	4
9	M	R	41	93	191	.75	4
AVG			53.3	81.6	174.7	.72	7.4
STDEV			5.5	16.9	12.5	.24	7.9

2.2 Testing Equipment

2.2.1 Test Apparatus – Isolated reflex testing

For the isolated reflex testing, the experimental apparatus was an instrumented ankle/knee actuator.   Each subject was seated on a Cybex isokinetic dynamometer (Humac NORM; CSMI, Massachusetts) (Figure 8). A rigid frame was constructed for imposing and measuring ankle joint torque (Figures 9 and 10). The frame was affixed to the motor axis of the Cybex. The sagittal plane knee axis of rotation was aligned with the axis of the Cybex motor and the ankle axis of rotation was aligned with the center of rotation of the ankle loader by adjusting the subject's position in the chair and the length of the ankle actuator device. The foot was secured onto a footplate with two dorsal straps (Figures 11, 12, 13). This isolated reflex testing component is able to determine the relative contribution of ankle load to the knee extensor reflex excitability.

Figure 8: Isolated Reflex Testing Station

Figure 9: Ankle Loader Dimensions – Side View
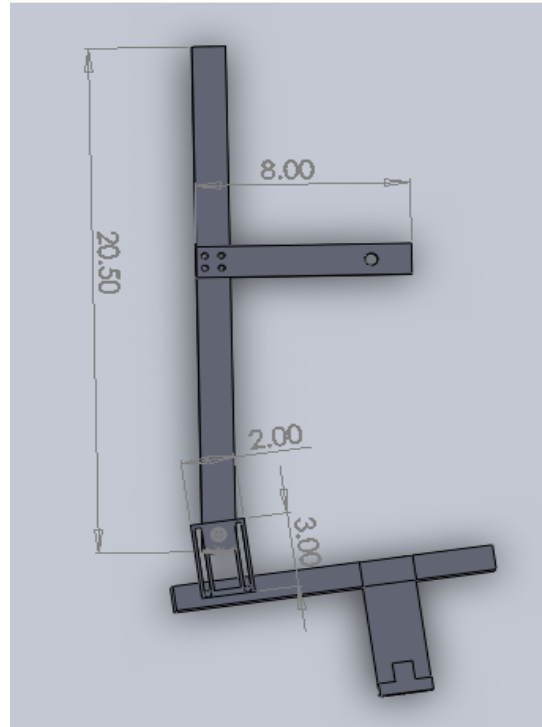
Figure 10: Ankle Loader Dimensions – Footplate View
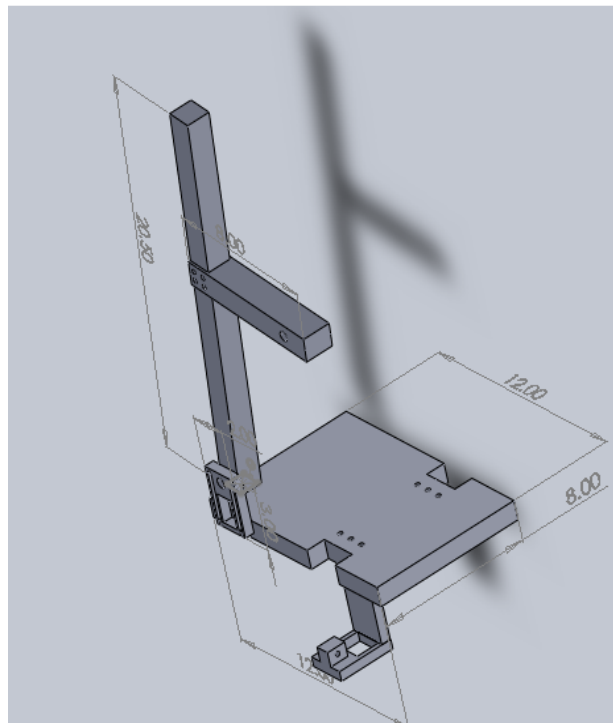
Figure 11: Ankle Loader Device

Figure 12: Ankle Loader with Knee in 30° of flexion

Figure 13: Ankle Loader with Knee in 90° of flexion

An air compressor was used to adjust the load applied to the ankle. A load cell was placed between the linear actuator and the ankle loader to provide feedback of the magnitude of the load being applied. A Labview (National Instruments, Austin, TX) interface program controlled a switch which gave the load command to the actuator (Figure 14).

Figure 14: Isolated Reflex Testing Full Setup Diagram
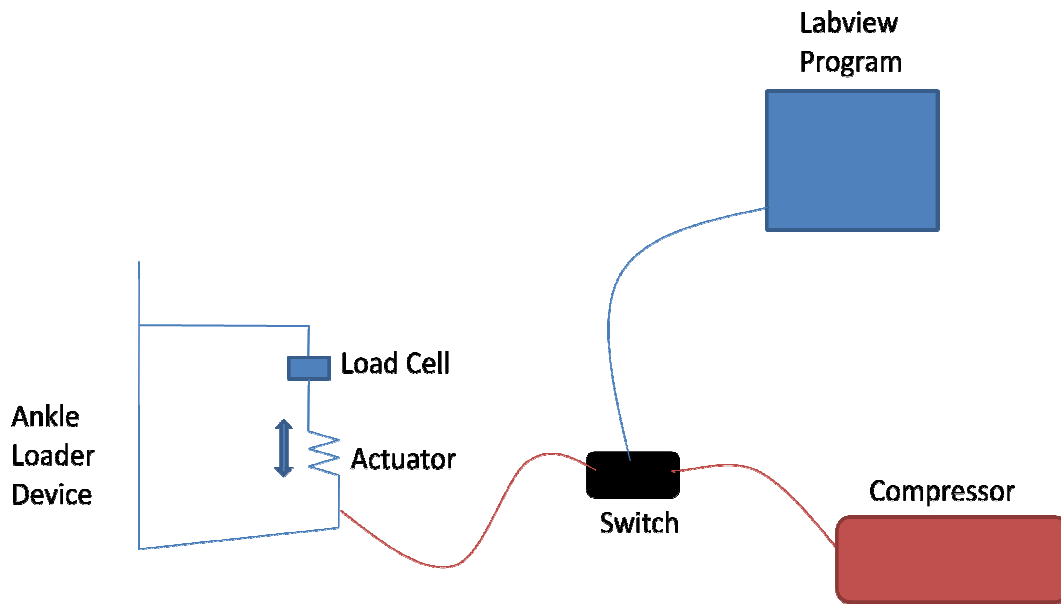
2.2.2 Test Apparatus – Gait Analysis Unloading

2.2.2.1 Unweighting Device for Gait

A custom unweighting structure was constructed using 8020 aluminum. The structure was rigidly mounted to the floor to improve stability and safety. The structure is 5 feet tall, 3 feet wide, and 2 feet deep (Figure 15).   Pulleys were connected to the top and bottom bars that allow for the dynamic counter weight theraband system. The 42" bar across the front was mobile and capable of sliding up and down along the structure to allow for dynamic adjustment as the subject completed their walking trial. The theraband load was attached to this moving bar at the bottom, while the top of the bar attached to the harness through the use of a winch on the 36" front bar (Figures 16 and 17).

Figure 15: Unweighting Structure Dimensions
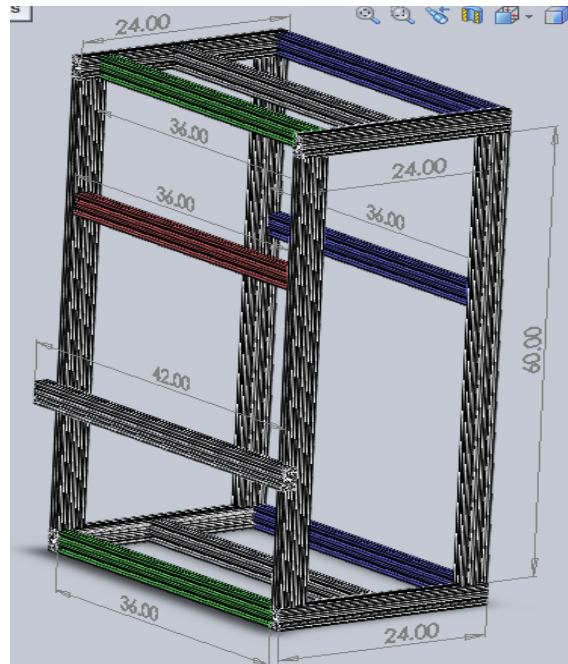
Figure 16: Unweighting System Diagram
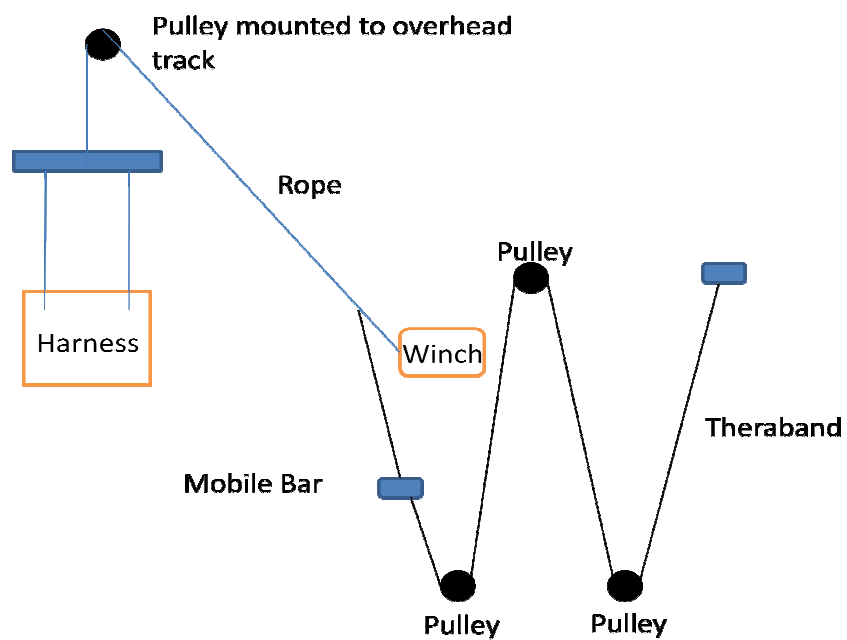
Figure 17: Constructed Unweighting Structure

The harness was attached to a rope that crosses over a pulley system

mounted to a track above the test subject on the ceiling so that it could move

freely with the subject. This allowed the subject to drift forward or back as they

were walking on the treadmill. The rope then continued from the other side of the

pulley to the moving bar on the unweighting structure. The load from the

therabands was then attached to the bar and a winch was used to adjust the

slack in the rope so that the subject was held at the appropriate unloading

amount. This allowed the unloading to be dynamic and to account for the

change in movement of the center of mass when taking a step.

2.2.2.2 Theraband Properties

Theraband latex resistance bands were used to unload the subject instead of

static weights (Figure 18). Due to the elastic properties of theraband, a long

piece of Theraband will change its load only minimally with the relatively smaller

superior/inferior changes in COM throughout the gait cycle.

Figure 18: Thera-Band Tubing
http://www.thera-band.com/store/products.php?ProductID=27

Different strengths of theraband were implemented and were stretched to the

appropriate length for the force we needed to produce (Table 2).

Table 2: Theraband Properties
(Patterson, Stegink Jansen et al. 2001)

Force Elongation (lbs) after stretching by percentage of its original length (%)										
	25	**50**	**75**	**100**	**125**	**150**	**175**	**200**	**225**	**250**
Yellow	1.1	1.8	2.4	2.9	3.4	3.9	4.3	4.8	5.3	5.8
Red	1.5	2.6	3.3	3.9	4.4	4.9	5.4	5.9	6.4	7.0
Green	2.0	3.2	4.2	5	5.7	6.5	7.2	7.9	8.8	9.6
Blue	2.8	4.6	5.9	7.1	8.1	9.1	10.1	11.1	12.1	13.3
Black	3.6	6.3	8.1	9.7	11	12.3	13.5	14.8	16.2	17.6
Silver	5.0	8.5	11.1	13.2	15.2	17.1	18.9	21.0	23.0	25.3
Gold	7.9	13.9	18.1	21.6	24.6	27.5	30.3	33.4	36.6	40.1

There were many possible combinations of weights that could be used in the unweighting device. It was set up with 5 different strengths available, 5 lbs, 20 lbs, 20 lbs, 40 lbs, and 10 lbs. They could be attached in any different combination to meet the necessary weight. For example, when trying to unload a 185 pound man by 40% of his body weight we would attach the 5 lb, 20 lb, 40 lb, and 10 lb therabands to reach a total of 75 lbs.

2.3 Methods – isolated reflex testing

Knee extensor reflex responses were elicited by rapidly rotating the knee into flexion using the Cybex isokinetic dynamometer. This provided a rapid passive stretch to the knee extensor muscles. The knee was rotated from 30 degrees to 90 degrees of knee flexion at both speeds of 90%sec an d 180%sec (Figures 12 and 13). This was done with muscles relaxed and with the hip flexors pre-activated to approximately 10% of their maximum voluntary isometric contraction.

Before knee flexion stretches were performed, the hemiparetic foot was fit into the previously described (see Section 2.2.1) custom brace which had the capability of providing a sustained load to the ankle extensors. This was done to simulate the load through the ankle (limb) during the stance phase of gait. The magnitude of the quadriceps muscle's response to the imposed knee movement was compared under different limb load conditions (no load, sustained load of 20Nm, and sustained load of 30 Nm). Five trials of each combination were performed of Cybex velocity (90°/sec and 180°/sec), appl ied load (no load, sustained load at the ankle plantarflexor of 20nm, and sustained load at the ankle plantarflexor of 30 nm.), and quadriceps activation level (relaxed and 10% pre-activated). The order of the conditions was randomized among the different subjects to account for an order effect. A 16-channel electromyographic (EMG) system (Motion Lab Systems, Baton Rouge, LA) was used to collect muscle responses from the vastus lateralis, vastus medialis, rectus femoris, medial hamstrings, medial gastrocnemius, tibialis anterior, and soleus using active surface electrodes.

During all trials, the knee joint position, torque, and velocity were recorded from the Cybex. Simultaneous collection of EMG data occurred at 1,000 Hz. Analysis of muscle activity was performed using custom designed Labview software. The EMG signals were full wave rectified and then filtered using a phase-corrected eighth-order, low-pass Butterworth filter with a cutoff frequency of10hz. The EMG data was normalized to the peak recorded during the MVIC collected before performing the reflex testing. The normalized EMG for each

muscle was combined into an ensemble average of the 5 trials for all conditions.

The individual muscle responses were compared for each condition.

2.4 Methods – Gait Analysis

All subjects also underwent gait analysis to determine lower extremity

kinematics and EMG immediately following the reflex testing. The EMG

electrodes stayed on the subject from the isolated reflex testing. Subjects

walked on a Bertec, dual belt, instrumented treadmill (Bertec Corp, Columbus,

OH) with different body weight support conditions which were controlled by the

dynamic unweighting device (Figure 19). The motions of the lower extremity

segments were tracked with an eight-camera motion capture system

(Vicon/Peak, Los Angeles, CA) collecting at 120 Hz. Joint centers were located

by the use of 12 mm retro reflective markers affixed to the iliac crests, the greater

trochanters (hip), the medial and lateral femoral condyles (knee), and the medial

and lateral malleoli (ankle). The foot was tracked by markers placed on the

posterior heel and over the 2nd and 5th metatarsal heads. Rigid thermoplastic

shells with four markers firmly affixed to each, were attached to the thigh and

shank and covered with elastic wrap. Another thermoplastic shell with three

markers was placed over the posterior sacrum (Figure 20).   The subjects wore a

harness that attached overhead which in turn attached to the unweighting device

at the side of the room. The weights were set at 40%, 20%, and 0% of the

subject's body weight. Treadmill speed was set to subject's comfortable walking

speed while the motion capture system calculated joint angles. The subject's

EMG data was also being recorded so that we could capture the magnitude of

thigh muscle activity during late stance/early swing under the different

unweighting conditions.

Figure 19: Treadmill and Harness Setup

Figure 20: Camera Marker Placement
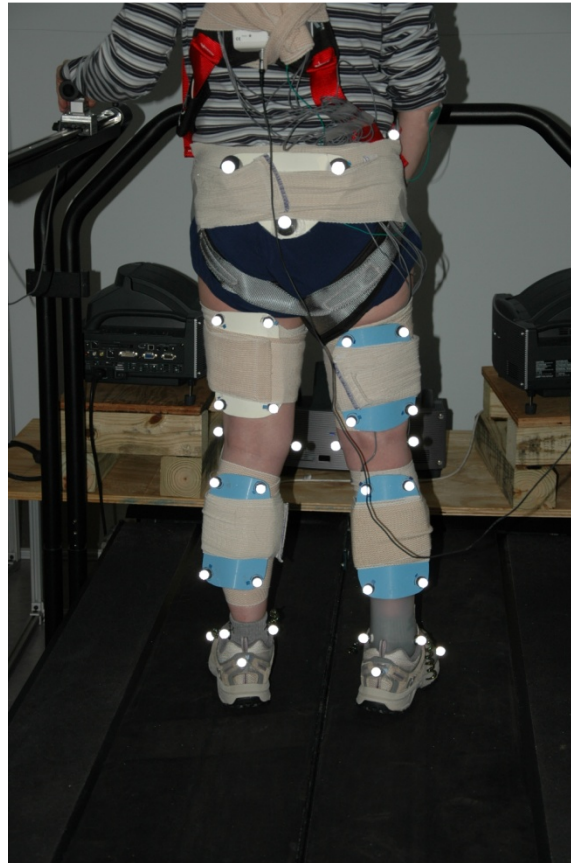
2.5   Statistical Analysis

Statistical analysis was performed using SPSS.   Outcome variables from the imposed knee and ankle movements in the isolated reflex testing were calculated from the ensemble average from the normalized EMG linear envelopes. These variables accounted for the baseline which was identified as the average muscle activity during the 250 ms period immediately before knee movement began. We were interested in the two variables: the peak reflexive response of the RF, VL, and VM muscles measured during the 150 ms period immediately following the beginning of the movement, and the prolonged response calculated as the

difference in EMG magnitude between the baseline average (250 ms before the movement started) and the final position average in muscle activity (250 ms after termination of the knee movement). Group means and standard deviations were compared using a two-way repeated measures ANOVA (repeated for both speed and load). If differences were found to be significant, post hoc testing was performed.

SPSS software was also used to analyze the gait data. An ensemble average of VL, RF, and VM activity was taken for each condition (0% BWS, 20% BWS, and 40%BWS). This was normalized to one gait cycle (heel strike to heel strike). An ANOVA was used to compare peak knee flexion angles during swing between conditions. An ANOVA was also used to analyze the VM, VL, and RF muscle activity between the three different load conditions (0%, 20%, and 40%).

Regression analyses were performed on both the isolated reflex data and the gait data. For the reflex data, slopes were calculated across the 3 different loads (no load, 20 Nm, and 30 Nm) for each muscle. This was done for both speeds (90°/sec and 180°/sec) and for both the peak response and the prolonged response. For the gait data, the slopes were calculated across the BWS conditions (0%, 20%, 40%) for each muscle. Regression analysis was performed on the slopes from each of these two settings to see if there was a relationship. During all tests, a significance level of $P < 0.05$ was used.

3.0   Results

3.1   Isolated Reflex Testing

As the knee began to flex rapidly, there was a brief burst of muscle activity from the quadriceps. In most cases, this muscle activity was sustained and tended to be greater than the Baseline period (Figure 21). We had unusable data from 2 subjects leaving N=7 during reflex testing.

Figure 21: Variables Obtained During Isolated Reflex Testing
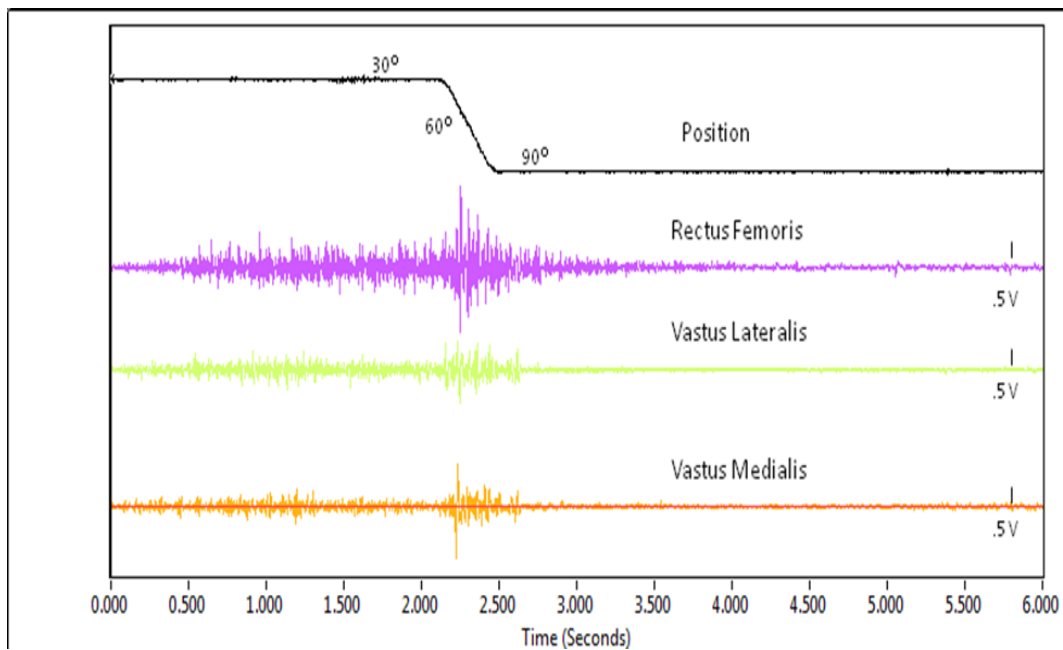
Baseline muscle activity was 4±6 %MVIC for the VL, 6±7 %MVIC for the RF, and 7±5 %MVIC for the VM prior to the knee flexion movement during the isolated reflex testing. For each muscle, there was no significant main effect for

load (VL: p=0.798; VM: p=0.217; RF: p=0.522) or speed (VL: p=0.565; VM: p=0.794; RF: p=0.902), and no interaction effect was present (VL: p=0.394; VM: p=0.588; RF: p=0.691) for Baseline muscle activity. Confirming this lack of difference between conditions provides confidence that the level of pre-activation did not influence the reflex results between conditions.

The initial peak muscle activity of the RF (p=0.027) and VM (p=0.035) exhibited a main effect for speed, although the VL did not (p=0.087) (Figure 22). No main effect for load was observed for any of the muscles(VL: p=0.755; VM: p=0.541; RF: p=0.873). The VM exhibited a significant interaction between load and speed (p=0.024), although the RF (p=0.542) and VL (p=0.163) did not. Post-hoc testing for the VM revealed that the difference between speeds was only significant with the application of load (0Nm: p = 0.559; 20Nm: p=0.030; 30Nm: p=0.012).   However, there was not a significant effect for load at the two different speeds (90°/sec: p=.991; 180°/sec: p=.306) (Figure 22).

Figure 22: Peak Muscle Response During Isolated Reflex Testing
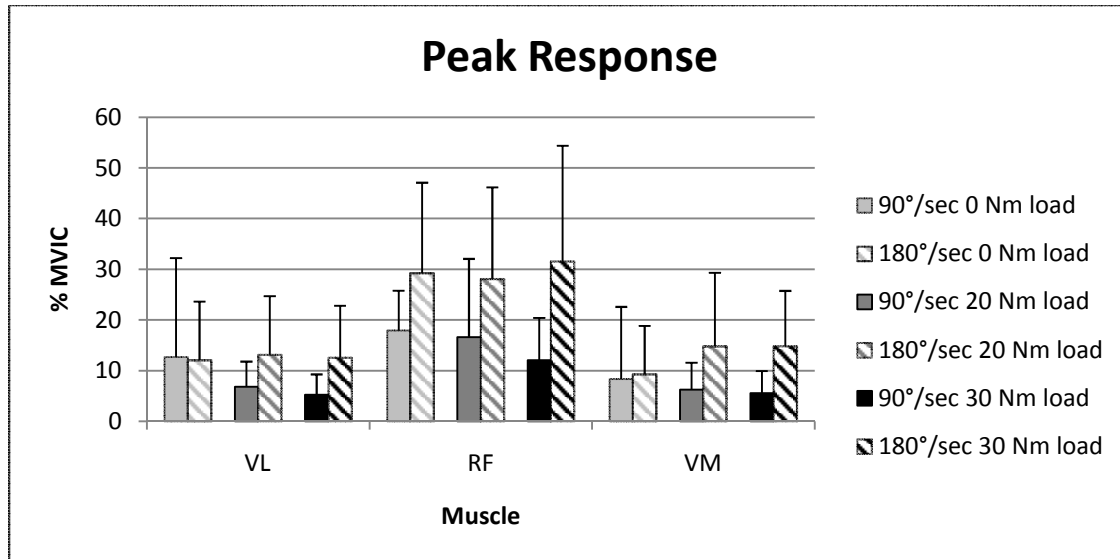
Although muscle activity appeared slightly greater immediately following the knee flexion movement compared to Baseline, this increased muscle activity was not influenced by load (VL: p=0.354; VM: p=0.208; RF: p=0.630) or speed (VL: p=0.646; VM: p=0.367; RF: p=0.123) with no significant interaction effect (VL: p=0.285; VM: p=0.217; RF: p=0.095) (Figure 23).

Figure 23: Prolonged Muscle Response During Isolated Reflex Testing
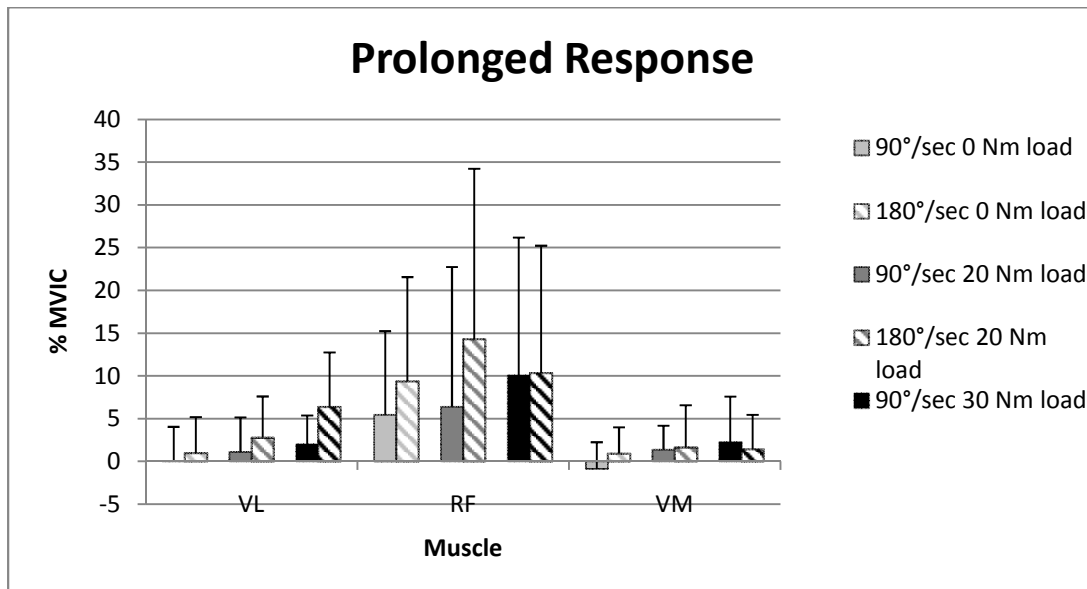
3.2 Gait Data

Subjects walked on the treadmill at their comfortable pace (mean: 0.72 m/s.) One subject had quadriceps data that was too noisy to use in the analysis. This may have been due to bad EMG electrode placement or movement during walking. This left N=8 during gait. With no body weight support, the subjects had an average knee flexion angle of 34±14° on their pa retic side and an average knee flexion angle of 65±4° on their non -paretic side. There was no significant difference of knee flexion at different BWS (p=0.569). However, there was a range of paretic knee flexion values, ranging from very stiff knee (i.e., 8.5°) to nearly normal (i.e., 55.4°) (Table 3).

Table 3: Knee Flexion Angle Results

	Paretic Knee Flexion Angle			Non-Paretic Knee Flexion Angle		
	0% BWS	20% BWS	40% BWS	0% BWS	20% BWS	40% BWS
All Subjects	34.1° (±14.4°)	33.3° (±14.7°)	33.7° (±14.8°)	65.1° (±4.6°)	64.4° (±4.4°)	62.9° (±5.7°)
Stiff-Knee Gait (n=4)	20.9° (±8.5°)	19.8° (±7.9°)	21.1° (±8.2°)	65.1° (±4.2°)	63.5° (±3.9°)	62.1° (±6.7°)
"Normal' knee flexion (n=5)	44.7° (±6.6°)	44.0° (±7.6°)	43.7° (±10.2°)	65.0° (±5.4°)	65.1° (±5.1°)	63.6° (±5.4°)

The muscle activity of the VL, RF, and VM were compared between the three different body weight support conditions (0% BWS, 20% BWS, and 40% BWS) for all subjects. For all subjects, the integral of muscle activity during late stance/early swing was not significantly different for VL ($p = 0.687$), VM ($p=0.469$), and RF ($p=0.499$) with different amounts of BWS (Figure 24).

Figure 24: Muscle Activity During Gait
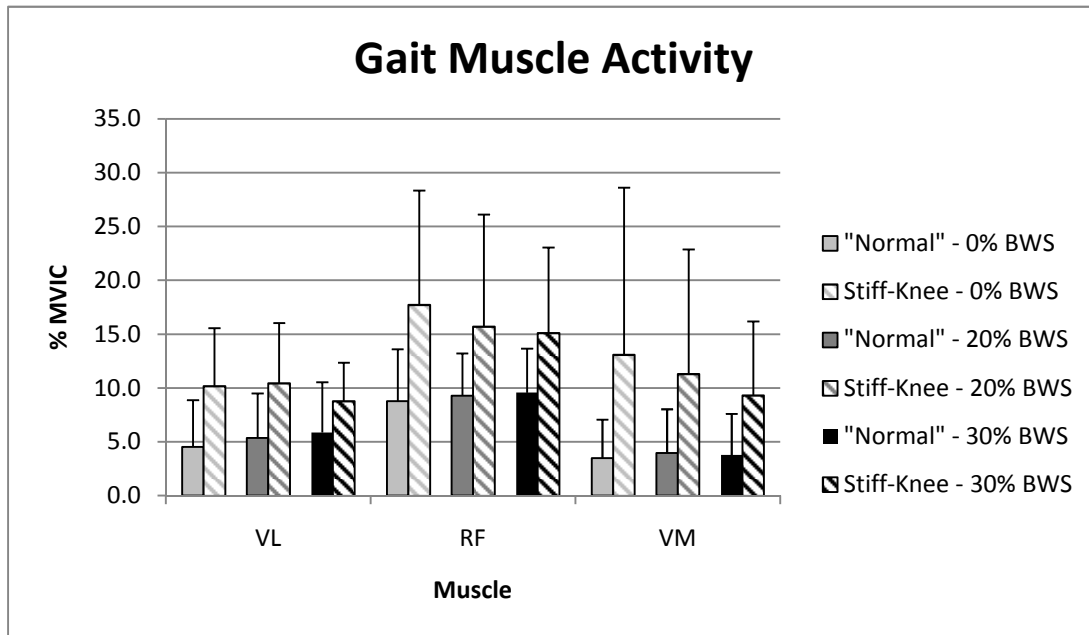
3.3 Relationship Between Gait and Reflex Testing

The regression analysis between the slopes of gait testing and the slopes for reflex testing showed no significant relationships (N=7) (Table 4).

Table 4: All Subjects – Regression Analysis Comparison

Muscle	Isolated Reflex condition	Gait condition	P value	R^2 value
VL	Final 90%sec	VL Gait	.814	.012
	Final 180%sec		.969	.000
	Peak 90%sec		.584	.064
	Peak 180%sec		.903	.003
RF	Final 90%sec	RF Gait	.616	.054
	Final 180%sec		.466	.110
	Peak 90%sec		.192	.312
	Peak 180%sec		.536	.081
VM	Final 90%sec	VM Gait	.476	.106
	Final 180%sec		.971	.000
	Peak 90%sec		.419	.134
	Peak 180%sec		.351	.175
Total Muscle Activity	Final 90%sec	Total Muscle Activity Gait	.711	.030
	Final 180%sec		.663	.041
	Peak 90%sec		.084	.481
	Peak 180%sec		.863	.007

When doing regression analysis on those subjects who exhibited very strong stiff-legged gait (knee flexion less than 30° during ga it, N = 4) significant relationships between isolated reflex testing and gait testing were apparent in the RF muscle. Specifically, the manner in which the prolonged response of the RF changed with the addition of load at 90°/sec was relate d to the way in which the RF changed when unloaded during gait. However, after the completion of post hoc tests, we observed that the relationship was due to the presence of one particular subject. We were unable to get a 30 Nm stretch on this subject, so this subject only had 2 data points which altered the reflex data slope. Once this subject was removed, no relationship was present. No significant relationship was present for the uniarticular VL or VM (Table 5).

Table 5: Stiff-Legged Gait Subjects – Regression Analysis Comparison

Muscle	Isolated Reflex condition	Gait condition	P value	R^2 value
VL	Final 90°/sec	VL Gait	.792	.043
	Final 180°/sec		.867	.018
	Peak 90°/sec		.896	.011
	Peak 180°/sec		.601	.159
RF	Final 90°/sec	RF Gait	**.011**	**.977**
	Final 180°/sec		**.040**	**.922**
	Peak 90°/sec		.597	.162
	Peak 180°/sec		.563	.191
VM	Final 90°/sec	VM Gait	.576	.180
	Final 180°/sec		.868	.017
	Peak 90°/sec		.625	.140
	Peak 180°/sec		.596	.163
Total Muscle Activity	Final 90°/sec	Total Muscle Activity Gait	.823	.031
	Final 180°/sec		.874	.016
	Peak 90°/sec		.283	.514
	Peak 180°/sec		.641	.129

4.0   Discussion

   My hypotheses were not supported by these data. Overall, these data

indicated that in both an isolated setting and a functional gait setting, the

modulation of limb load did not have a significant effect on quadriceps muscle

activity. There did, however, seem to be an effect of speed in the isolated reflex

setting for the rectus femoris and vastus medialis. Despite the lack of significant

findings in regards to my hypotheses, I believe that there are important

implications for gait retraining post-stroke.

   Most studies have focused heavily on the role of limb load on the ankle

muscles (gastrocnemius, soleus, tibialis anterior). My study, however, focused

on the knee extensors (vastus lateralis, rectus femoris, vastus medialis). Klarner

et al, performed a gait analysis study with a range of body weight support from

0% to 100% monitoring the knee extensors and the ankle plantarflexors. They

found the largest modulation of muscle activity in the medial gastrocnemius,

lateral gastrocnemius, and soleus (Klarner, Chan et al.). Another study that

confirmed our findings was performed by van Hedel, et al. This study also

examined muscle activity with varying amounts of body weight support. They

found that the VM, RF, and VL were much less influenced by unloading

variations than the TA and MG (van Hedel, Tomatis et al. 2006). Bachmann et

al. also found that the soleus was the muscle most affected by changes in body

weight support (Bachmann, Muller et al. 2008). These findings along with our

own, may suggest that the limb load sensors, which are believed to reside

primarily within the ankle plantarflexor muscles act more strongly at the local

level and have less of an excitatory input to the other musculature throughout the

limb.

4.1 Interpretation of Results

4.1.1 Relationship Between Isolated Reflex Testing and Gait Analysis

We hypothesized that the individuals who showed greater extensor

excitability with limb load in an isolated situation would exhibit greater knee

extensor activity during gait. However, when looking at the subjects as a whole

group, no significant relationship was found between the slopes of the regression

lines of gait for each muscle and that of isolated reflex testing. When examining

the regression data of only those subjects who exhibited strong stiff-knee gait

(knee flexion<40°), there was a significant relationshi p for the RF prolonged

response at both speeds of 90°/sec and 180°/sec. Howeve r, upon further

analysis, we observed that one subject was an extreme outlier because only 2 of

the 3 conditions (0 Nm and 20Nm) were accounted for due to not being able to

get a 30 Nm stretch. This altered the slope of the isolated reflex data. Once this

subject's data was removed, no significant relationship existed among the

remaining data points.

4.1.2 The Relationship Between Speed and Load

The vastus medialis showed a significant interaction between load and

speed during the isolated reflex testing. Further testing revealed that the

difference between speeds was only significant with the application of load (there

was no difference at 0Nm load). These results would seem to indicate that load

modulates the effect of speed. This could imply that as locomotor training is performed at increasing speeds, there is a potential to change the way that the load is sensed. However, we are unable to make a full conclusion because we did not do the gait analysis at different speeds. Therefore, we do not know if the same effect can be observed during walking.

4.1.3 Dynamic Unweighting System

As described in the methods, our unweighting system allows for dynamic unweighting. This differs from many other unweighting systems because it keeps the load constant. This could possibly explain why we are getting different results than some other published research (Mulroy, Klassen et al.; Visintin and Barbeau 1989; Finch, Barbeau et al. 1991; Hesse, Konrad et al. 1999; Barbeau and Visintin 2003). Others may find differences because their load is variable during walking rather than nearly constant (within 7%), as our subjects are experiencing. This implies that in other studies, the subject is generating additional muscle activity by the constant weighting and unweighting that they experience in their counterweight unloading devices. Our constant load system may be eliminating the change in muscle activity that is caused by the rapid loading and unloading in the counterweight systems.

4.2 Positive Aspects of the Study

One of the strongest aspects of the study was the fact that both sessions (isolated reflex testing and gait analysis) were done on the same day. This allowed the EMG electrodes to remain on the subject for both studies which means much less variability or error due to EMG placement or movement.

Another positive aspect of this study was the dynamic unweighting system. Our body weight support system allowed for there to be a constant load on the subject at all times unlike many of the static counterweight systems often used.

Each subject completed their test session in a short period of time, about an hour for the isolated reflex testing and only a few minutes on the treadmill for gait analysis. This reduced the potential effects of muscle fatigue.

Lastly, both test setups in the study could accommodate a large size range of subjects. The ankle loader could be adjusted to accommodate from a woman's size 5 shoe to a men's size 15 shoe. The ankle loader and Cybex both had adjustments for different leg lengths as well. The unweighting device also allowed room for a height of 6'8''. This was positive as we were not required to limit our subject pool.

4.3   Limitations of the Study

One limitation of this study was the small sample size (n=9). While there were several values that demonstrated significance, it is unknown whether a larger sample size would have lead to more observations of statistical or clinical significance.

The isolated reflex testing was further limited by the fact that we tested the subjects at 3 different fixed load conditions (0Nm, 20 Nm, and 30 Nm). It might have been more beneficial to normalize the load to the subject's body weight, a method used during body weight support testing, rather than choose 3 constant loads for all subjects.

We also were unable to get all subjects to "pre-activate" to the exact same level. This could have played a role in the outcome of the "peak" and "prolonged" values as they were calculated from the baseline data.

Appendix A: Subject Information Sheet

Subject: _____Today's Date: _____Birthdate: _____

Height: _____Weight:_____Stroke Onset: _____Paretic Side: L   R

Foot Width: Left _____Right_____   Pelvic Depth: _____

Fugl-Meyer: _____/(34)   Berg: _____/(56)

Minute	Gait Speed	Pounds Unweighted	FP0 Left	FP0 Right
0%				
20%				
40%				

Channels

0	Cybex Position
1	Cybex Velocity
2	Cybex Torque
3	Loader Torque (Force really)
4	Vastus Lateralis
5	Rectus Femoris
6	Vastus Medialis
7	Medial Hamstring
8	Medial Gastroc
9	Soleus
10	Tib Anterior
11	Load release (5V for loaded, 0V when release)

Trial Order:

___   off_90_pre
___   off_180_pre
___   off_90_nop
___   off_180_nop
___   20_90_pre
___   20_180_pre
___   20_90_nop
___   20_180_nop
___   30_90_pre
___   30_180_pre
___   30_90_nop
___   30_180_nop

Appendix B: Subject Consent Form

University of North Carolina-Chapel Hill
Consent to Participate in a Research Study
Adult Subjects
Biomedical Form

_____

IRB Study #____07-0845_____
Consent Form Version Date: __6/18/07_____

Title of Study: Limb Load Affects Reflexes During Walking Post-Stroke

Principal Investigator: Michael Lewek, PT, PhD
UNC-Chapel Hill Department: Department of Allied Health Sciences, Div of PT
UNC-Chapel Hill Phone number: 919-966-9732
Email Address: mlewek@med.unc.edu
Co-Investigator: Amy West; Richard Goldberg, PhD; Vicki Mercer, PT, PhD
Funding Source: University Research Council

Study Contact telephone number: 919-966-9732
Study Contact email: mlewek@med.unc.edu

_____
_

What are some general things you should know about research studies?
You are being asked to take part in a research study. To join the study is
voluntary.
You may refuse to join, or you may withdraw your consent to be in the study, for
any reason.

Research studies are designed to obtain new knowledge that may help other
people in the future. You may not receive any direct benefit from being in the
research study. There also may be risks to being in research studies.

Deciding not to be in the study or leaving the study before it is done will not affect
your relationship with the researcher, your health care provider, or the University
of North Carolina-Chapel Hill. If you are a patient with an illness, you do not
have to be in the research study in order to receive health care.

Details about this study are discussed below. It is important that you understand
this information so that you can make an informed choice about being in this
research study. You will be given a copy of this consent form. You should ask
the researchers named above, or staff members who may assist them, any
questions you have about this study at any time.

What is the purpose of this study?
Unloading patients with stroke using a harness support for walking training has become routine in the clinical setting. The purpose of this research study is to learn if weight bearing through your legs increases the amount of muscle activity while walking in people who have had a stroke. Walking difficulties are common in individuals who have had a stroke, and may be due to abnormal muscle control. Muscles are controlled, in part, by sensing the weight through the legs while walking. The aim of this study is to see if the amount of weight bearing during walking alters muscle activity in individuals who have had a stroke.

You are being asked to be in the study because you have either had a stroke or because you have not experienced a stroke and wish to volunteer as a control subject.

Are there any reasons you should not be in this study?
You should not be in this study if you have heart, muscle, or bone problems that would limit your ability to move your legs or walk.

How many people will take part in this study?
If you decide to be in this study, you will be one of approximately 40 people in this research study (20 individuals following stroke, and 20 individuals with no history of neurological injury). All individuals will perform identical testing.

How long will your part in this study last?
If you decide to participate, you will be asked to attend 2 sessions of testing. Each session will last approximately 3 hours, for a total of 6 hours. There will be no follow-up required to participate in this study.

What will happen if you take part in the study?
You will need to come to UNC-CH for 2 sessions of testing to participate in this study. For one session you will be asked to walk on a treadmill with a portion of your body weight supported by a harness, and for the other session, you will be asked to sit in a chair to have your reflexes tested. You will need to wear loose fitting shorts and comfortable shoes for both testing sessions. If you do not have a pair of shorts, we can provide them for you. The study will begin with a series of tests in which you will have your strength, balance, and walking ability assessed by a physical therapist. Once these tests are completed, you will have sensors taped to your thighs, calves, and shins to monitor the activity of your leg muscles. Before walking on the treadmill, we will also tape small reflective balls (about the size of a dime) to your legs to monitor how your hips, knees, and ankles bend. Once this equipment is taped to your legs, we will ask you to do various leg movements to ensure that the sensors are working properly. You will then perform one of the following two protocols, with the other protocol being performed on a separate day:
Reflex testing:

- You will be seated in a chair to begin the data collection.
- You will be asked to kick out a small amount against a machine that will bend your knee.
- The machine will bend your knee, while another machine either provides or does not provide a stretch to your calf muscles. When the stretch is applied, there will be some trials in which the stretch is released at the end of the knee movement, and some trials in which the stretch will persist for several seconds after the knee movement.
- The stretch and the knee movement should not hurt, and are just intended to produce a reflexive response in your thigh muscles (like a "tendon tap" or "knee-jerk response").

Walking Analysis:
- A harness will be placed around your waist that will be connected overhead to an unweighting system. The unweighting system will support a portion of your body weight while you walk. Support is provided at 75%, 50%, 25%, and 0% of your body weight.
- You will walk on a treadmill at a comfortable speed for 5 minutes at each body weight support level.
- While you are walking, we will be collecting data about how your legs are moving and how active your muscles are. Approximately 20 minutes of walking on the treadmill will be required for this session with rest breaks provided as needed.

Both of these procedures are required for participation in the study.

What are the possible benefits from being in this study?
Research is designed to benefit society by gaining new knowledge that will improve rehabilitation programs. You will not benefit personally from being in this research study.

What are the possible risks or discomforts involved with being in this study?
Participation in this study may involve the following risks:
There is a risk of muscle fatigue from walking on the treadmill, although this risk will be minimized by providing you with long rest breaks as needed. In addition, the skin may need to be shaved for the placement of sensors, and although there is a risk that the skin may receive a small cut from the electric shaver, our shavers are designed to minimize this risk. Aloe vera shaving cream can be used to minimize the risk of skin irritation that may be caused by shaving. There is also the possibility that the knee movement could elicit short lasting spasms in individuals with stroke. These spasms would be similar to what you may experiences on a day-to-day basis during rapid leg movements.

What if we learn about new findings or information during the study?
You will be given any new information gained during the course of the study that might affect your willingness to continue your participation.

How will your privacy be protected?
No subjects will be identified in any report or publication about this study. Although every effort will be made to keep research records private, there may be times when federal or state law requires the disclosure of such records, including personal information. This is very unlikely, but if disclosure is ever required, UNC-Chapel Hill will take steps allowable by law to protect the privacy of personal information. In some cases, your information in this research study could be reviewed by representatives of the University, research sponsors, or government agencies for purposes such as quality control or safety.
Any personal information collected will be stored in a locked office. This information will be linked to the collected data by a code that does not identify you. Only the investigator and study team will have access to this information.

What will happen if you are injured by this research?
All research involves a chance that something bad might happen to you. This may include the risk of personal injury. In spite of all safety measures, you might develop a reaction or injury from being in this study. If such problems occur, the researchers will help you get medical care, but any costs for the medical care will be billed to you and/or your insurance company. The University of North Carolina at Chapel Hill has not set aside funds to pay you for any such reactions or injuries, or for the related medical care. However, by signing this form, you do not give up any of your legal rights.

What if you want to stop before your part in the study is complete?
You can withdraw from this study at any time, without penalty. The investigators also have the right to stop your participation at any time. This could be because you have had an unexpected reaction, or have failed to follow instructions, or because the entire study has been stopped.

Will you receive anything for being in this study?
You will receive $40 for each testing session that you participate in. This means that you can receive up to $80 for completing both sessions involved in this study.

Will it cost you anything to be in this study?
If you enroll in this study, you will have some tests and procedures that are only part of the research study. You will not be billed for these tests and procedures. You will be responsible for the costs of transportation to and from the laboratory and for parking fees during the test session.

What if you are a UNC employee?
Taking part in this research is not a part of your University duties, and refusing will not affect your job. You will not be offered or receive any special job-related consideration if you take part in this research.

Who is sponsoring this study?

This research is funded by the University Research Council of the University of North Carolina at Chapel Hill.

What if you have questions about this study?
You have the right to ask, and have answered, any questions you may have about this research. If you have questions, or if a research-related injury occurs, you should contact the researchers listed on the first page of this form.

What if you have questions about your rights as a research subject?
All research on human volunteers is reviewed by a committee that works to protect your rights and welfare. If you have questions or concerns about your rights as a research subject you may contact, anonymously if you wish, the Institutional Review Board at 919-966-3113 or by email to IRB_subjects@unc.edu.

- -
- - -
Title of Study: Limb Load Affects Reflexes During Walking Post-Stroke
Principal Investigator: Michael Lewek, PT, PhD

Subject's Agreement:

I have read the information provided above. I have asked all the questions I have at this time. I voluntarily agree to participate in this research study.

_____
_____
Signature of Research Subject   Date
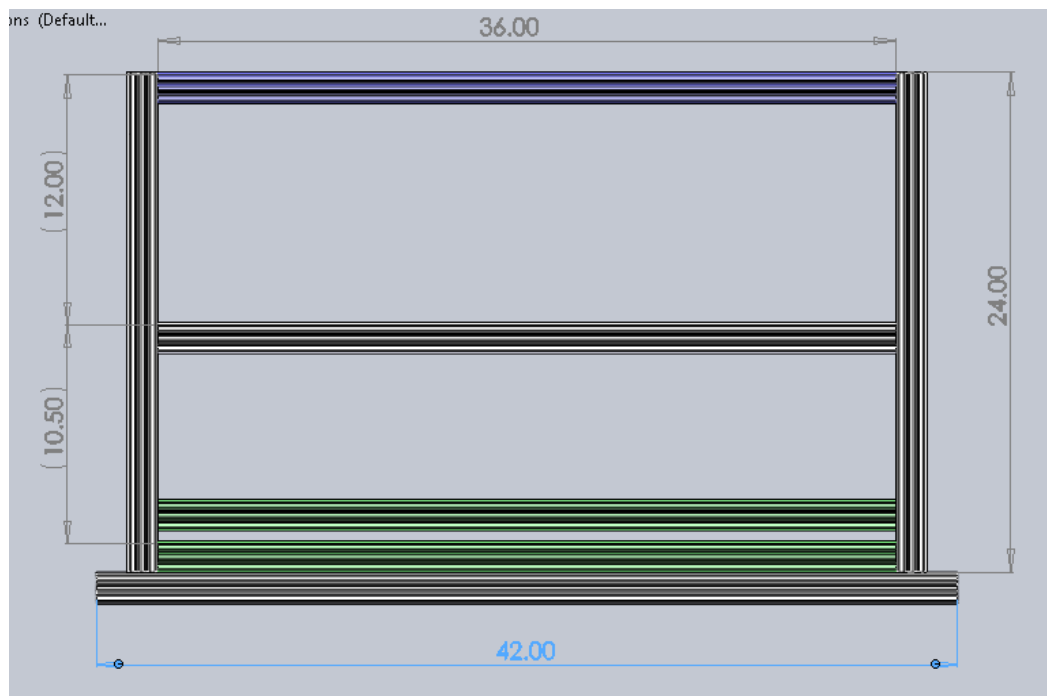
_____
Printed Name of Research Subject

_____
_____
Signature of Person Obtaining Consent   Date

_____
Printed Name of Person Obtaining Consent

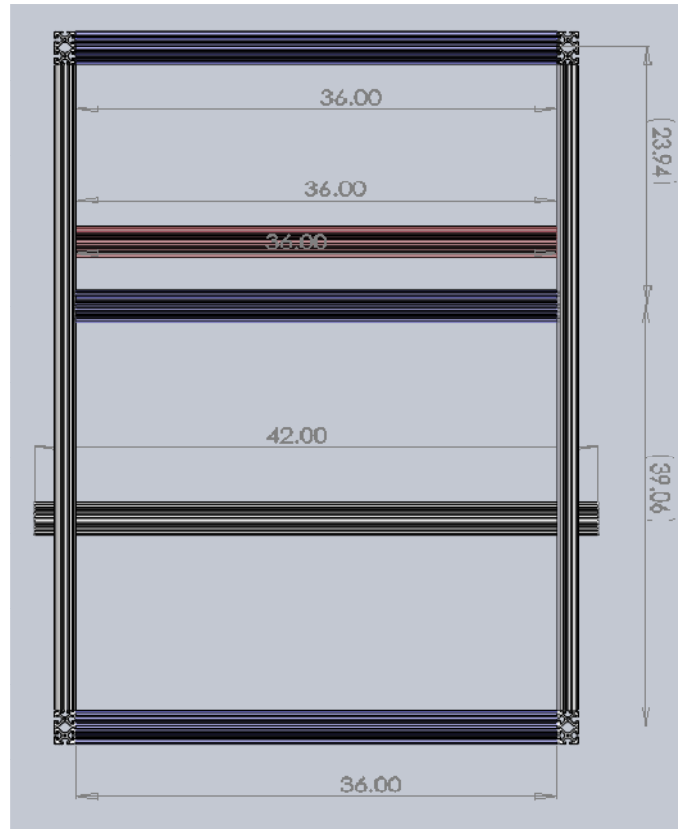**Appendix C: Device Figures**
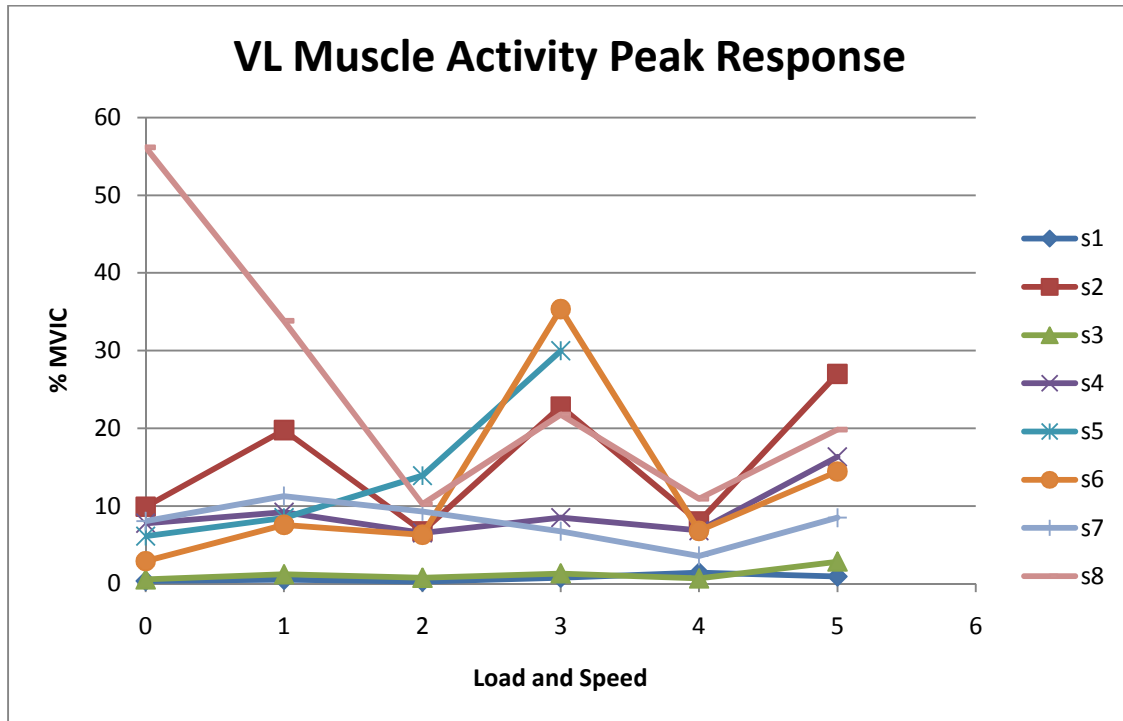
Figure 25: Unweighting Device: Top View

Figure 26: Unweighting Device: Front View
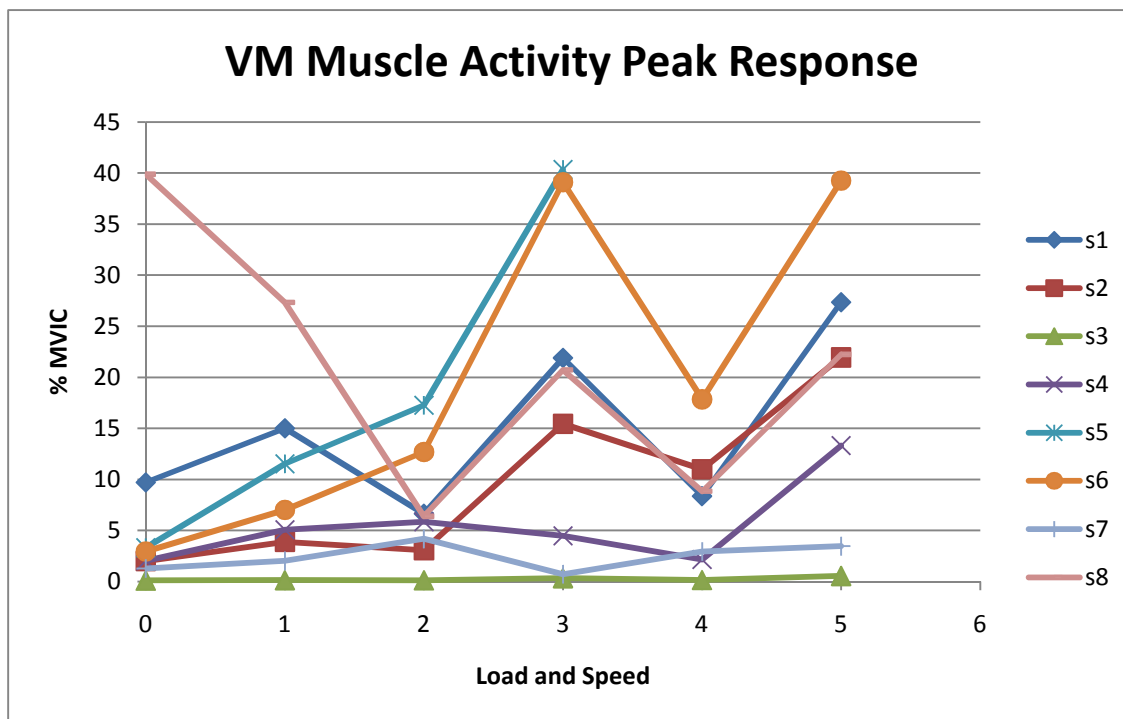
Appendix D: Isolated Reflex Testing Raw Data

Figure 27: VL Peak Response

Conditions: 0- 0 load, 90 speed; 1- 0 load, 180 speed; 2 - 20 load, 90 speed; 3 - 20 load, 180 speed; 4 - 30 load, 90 speed; 5 - 30 load, 180 speed
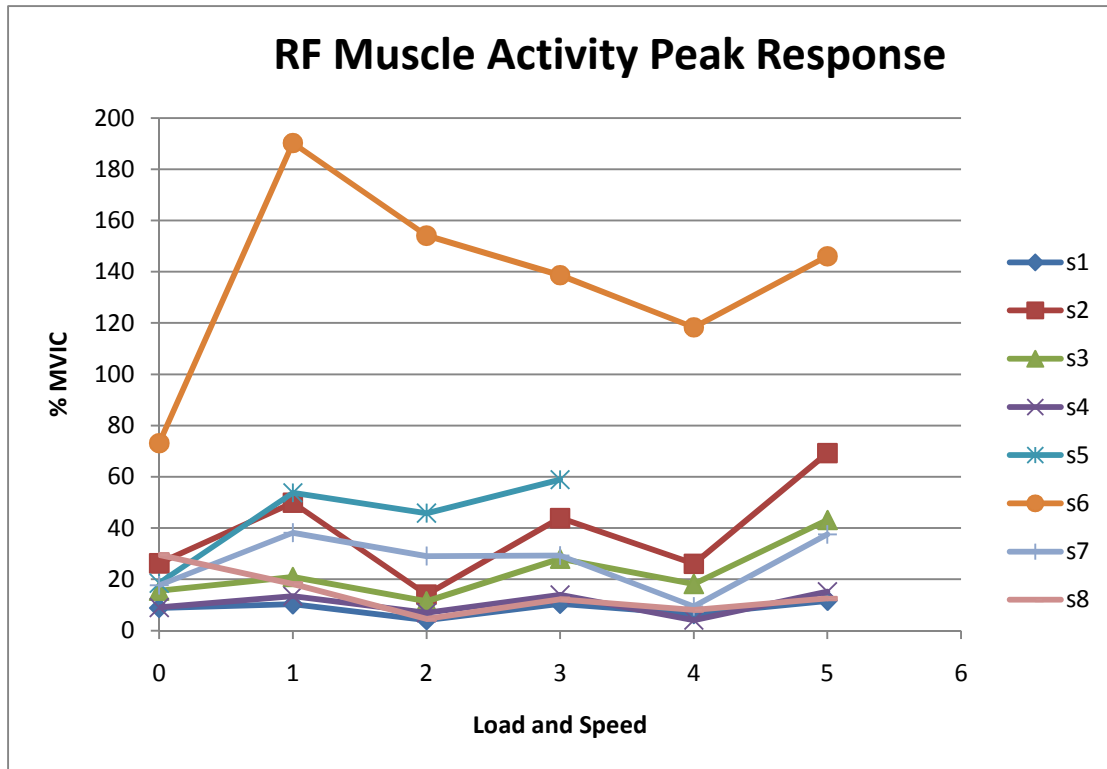
Figure 28: VM Peak Response

Conditions: 0- 0 load, 90 speed; 1- 0 load, 180 speed; 2 - 20 load, 90 speed; 3 - 20 load, 180 speed; 4 - 30 load, 90 speed; 5 - 30 load, 180 speed

Conditions: 0- 0 load, 90 speed; 1- 0 load, 180 speed; 2 - 20 load, 90 speed; 3 - 20 load, 180 speed; 4 - 30 load, 90 speed; 5 - 30 load, 180 speed

Figure 30: VL Prolonged Response

Conditions: 0- 0 load, 90 speed; 1- 0 load, 180 speed; 2 - 20 load, 90 speed; 3 - 20 load, 180 speed; 4 - 30 load, 90 speed; 5 - 30 load, 180 speed
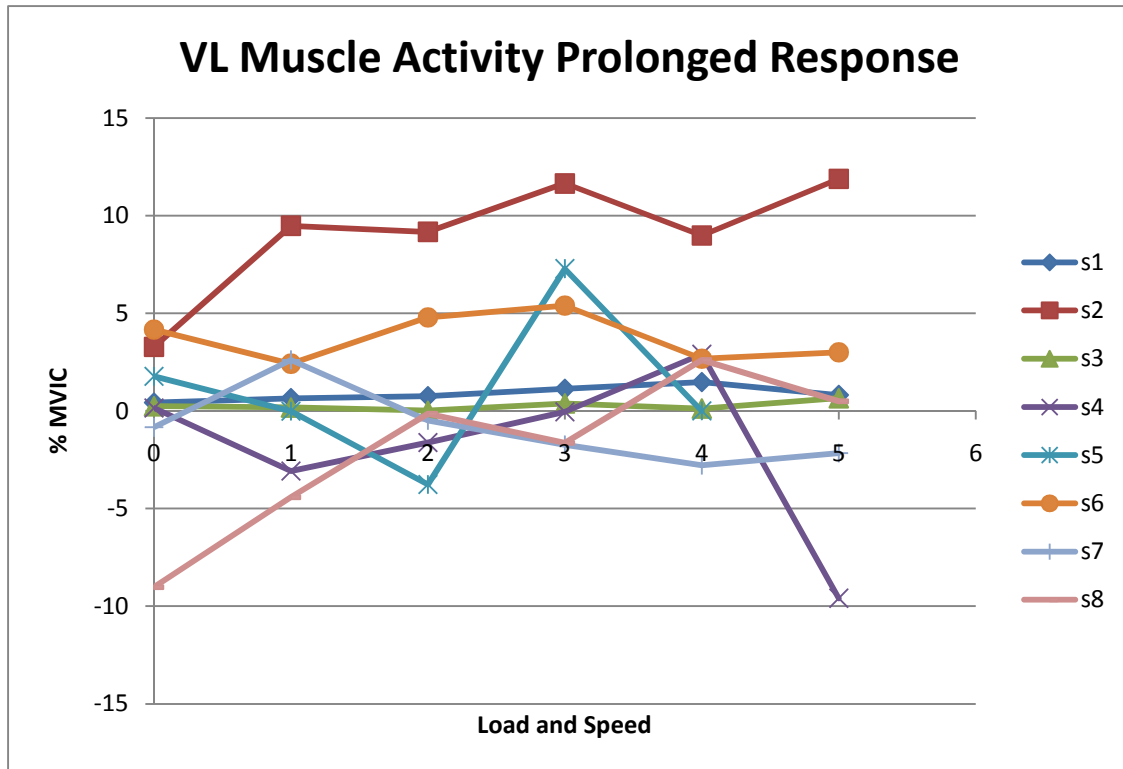
Figure 31: VM Prolonged Response

Conditions: 0- 0 load, 90 speed; 1- 0 load, 180 speed; 2 - 20 load, 90 speed; 3 - 20 load, 180 speed; 4 - 30 load, 90 speed; 5 - 30 load, 180 speed
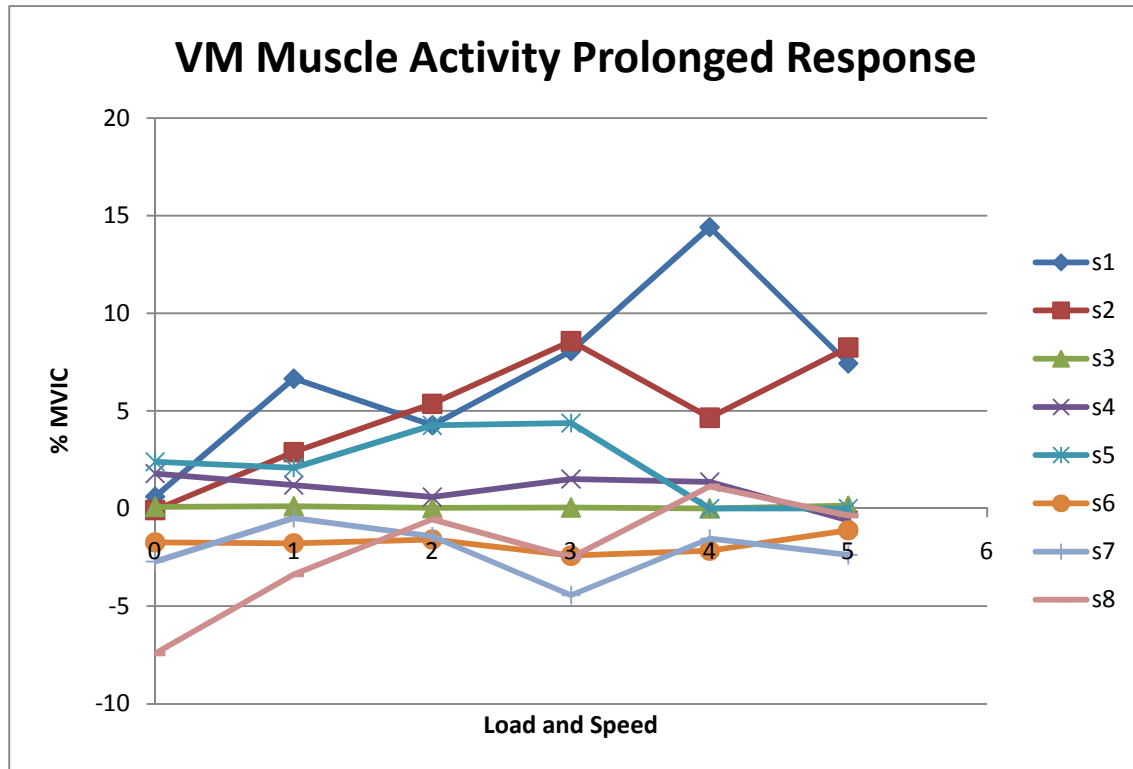
Figure 32: RF Prolonged Response

Conditions: 0- 0 load, 90 speed; 1- 0 load, 180 speed; 2 - 20 load, 90 speed; 3 - 20 load, 180 speed; 4 - 30 load, 90 speed; 5 - 30 load, 180 speed
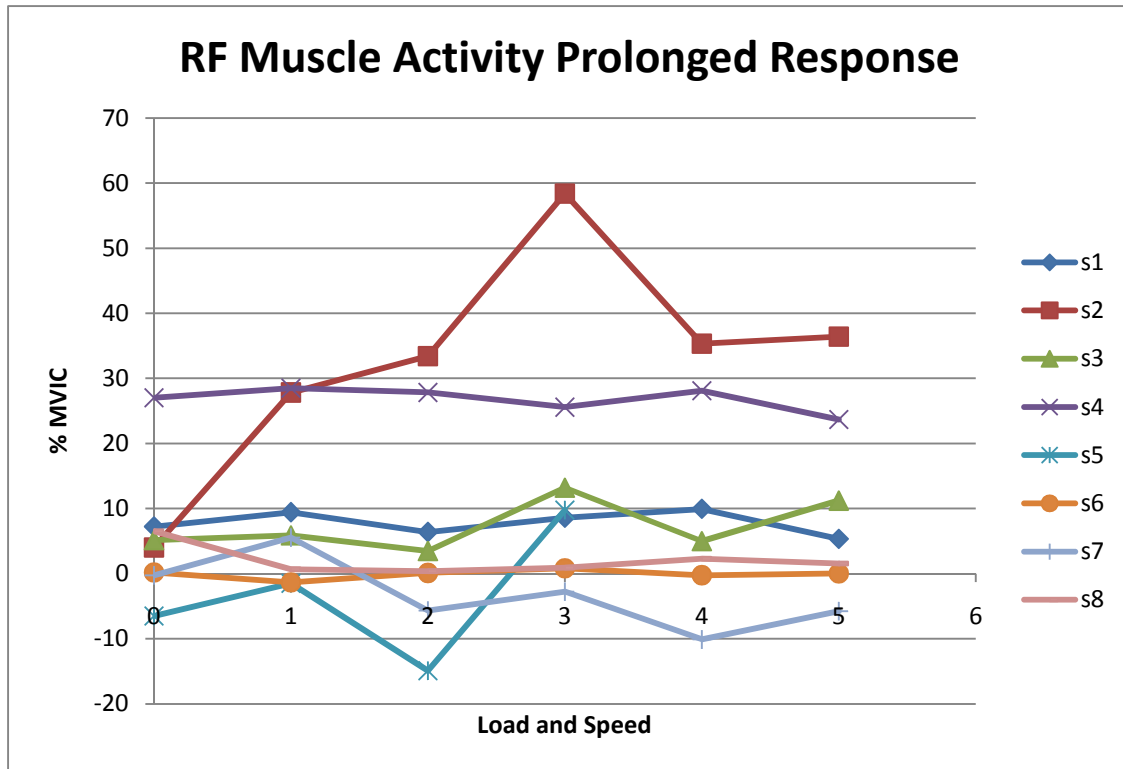
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