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Effects of Various Mobility Aids on Lower-Extremity Muscle Activity

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ABSTRACT

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Millions of people each year spend some portion of their time using mobility aids to facilitate periods of non-weight bearing ambulation. The use of these devices changes the loading conditions of the lower extremities, which may result in skeletal muscle adaptations. The purpose of this work was to evaluate the effects of 3 types of mobility aids on lower-extremity muscle activity. Evaluation was based on 1) measured muscle activation signals using electromyography (EMG), and 2) measured joint kinematics and ground reaction forces, which were used to predict muscle forces.

16 healthy subjects (7 female, 9 male), ages 18-27 participated in the study. Subjects were instructed to ambulate using each of three mobility aids (crutches, a knee scooter and a temporary-injury prosthesis) as well as normal walking. EMG and motion capture were used to obtain bilateral data from the lower half of the body during ambulation on each of these mobility aids and walking (10 trials on each per subject). Muscles studied were right and left vastus lateralis (VLR, VLL), rectus femoris (RFR, RFL), Biceps femoris long head (BFR, BFL), and gastrocnemius medialis (GMR, GML). Joint kinematics and ground reaction force data (joint kinetics) were acquired using a standard camera-based motion capture system. The measured joint kinetics were used as inputs to the open source musculoskeletal biomechanics software OpenSim (SimTK, Stanford, CA), which allowed prediction of muscle force data for a representative subject during each mode of ambulation.

As compared to walking, the following differences in EMG activation were significant. For the knee scooter, increases in VLR, RFR, BFL and decreases in GMR. For the TI prosthesis, increases in VLR, RFR, BFR, VLL, RFL, GML and decreases in GMR. For crutches, increases in BFR, VLL, RFL, BFL, GML and decreases in VLR, GMR. Muscle force results were similar, but demonstrated inadequacy of current musculoskeletal simulation software to resolve muscle forces during non-weight bearing portions of gait based solely on kinetic data. Results for walking data were similar to what is reported in the literature for normal gait.

This study provides useful bilateral data that describe measured lower-extremity EMG activation amplitudes and muscle force predictions based on kinetic data during ambulation using three different ambulatory aids, compared to normal walking. Based on a criteria of maintaining muscle activation, the TI prosthesis proved most effective among the devices tested. The data presented will be valuable to clinicians in providing insight into which mobility aid may be best suited for a particular patient. It is anticipated that these data will provide designers of mobility aids with a protocol for evaluation of designs based on their potential to cause or prevent muscle adaptations.

Keywords: leg, crutches, knee scooter, mobility aid, electromyography

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1 INTRODUCTION

1.1 Problem statement and motivation

There are over 4.5 million visits to the emergency room for below-the-knee injuries each year in the U.S. [1]. These injuries are commonly followed by a period of non-weight bearing, which may be 6-8 weeks or longer depending on the type of injury. In the past, patients have been limited primarily to the use of crutches or wheel chairs to enable compliance with this period of non-weight bearing. In recent years new devices, such as knee scooters and others, have been introduced to the market allowing a broader selection of devices to choose from and offering greater mobility or range for the user. In some cases a patient will use a combination of devices to meet their needs.

Each type of device causes different loading conditions on the muscles, which may lead to variations in disuse muscle atrophy. Atrophy is partly characterized by a reduction in cross sectional area (CSA) of the fibers [2]. Research has shown that atrophy can begin within 12-48 hours and is followed by rapid and dramatic loss of muscle mass and strength [3-5]. Previously researched methods for counteracting or preventing atrophy have largely shown partial or limited success [4-20]. While the research to date has focused primarily on the nature of atrophy as well as various methods to treat or reduce it, very little focus has been placed on the effects mobility aids have on atrophy. Given the number of individuals who require the use of these devices each year and the potential they have to affect atrophy, there is clear motivation to pursue further investigation of muscle activity during use of these devices. Increasing understanding of this

may aid healthcare professionals in their recommendations of devices for a particular patient based on their rehabilitation needs. It may also provide designers of such devices a protocol for evaluation of a design against it's potential to cause or prevent atrophy. Ultimately, this may lead to reducing the effects of atrophy, which will improve quality of life during and after recovery.

1.2 Research objective

The purpose of this work was to evaluate the effects of 3 types of mobility aids on lower-extremity muscle activity. Evaluation was based on 1) measured muscle activation signals using electromyography (EMG), and 2) measured joint kinematics and ground reaction forces, which were used to predict muscle forces for a representative subject. We hypothesized that among the mobility aids tested, crutches will be least effective at maintaining muscle EMG activation and predicted muscle forces relative to normal walking.

1.3 Chapter summaries

Chapter 2 consists of a review of anatomy of the lower extremities, electromyography and normal walking, the nature of skeletal muscle atrophy, artificially inducing and measuring atrophy, methods of preventing or treating atrophy, electromyography and atrophy, motion capture, and current mobility aids. Understanding these topics will help provide a background for the present study.

Chapter 3 is comprised of an original article to be submitted to the Journal of Sport and Health Science. It constitutes the introduction, methods, results, discussion and conclusion related to primary research conducted as part of this thesis. The focus of the paper is to show the effects of various mobility aids on lower-extremity muscle activity.

Chapter 4 provides a more in-depth look at the comparison of predicted muscle force and EMG for a representative subject for various devices used in this research. It also further validates the results of EMG data presented in chapter 3.

Chapter 5 provides a summary, which includes contributions of this work, conclusions and recommendations on future work.

2 BACKGROUND

2.1 Anatomy of the lower extremities

Key components of the lower extremities are bone, articular cartilage, ligaments, tendons and skeletal muscle. Bones provide structural support and contribute to locomotion by transferring loads. Bones also contribute to locomotion by providing points of attachment for muscle to exert force on the skeletal system via tendons, which connect muscle to bone. Ligaments connect bone to bone and articular cartilage provides ideal surface conditions for relative motion between bones at the joint interface [21]. Many muscles of the lower extremities contribute to locomotion by performing functions related to antigravity (i.e. opposing the forces of gravity during ground contact), stabilization (including deceleration of a limb), propulsion (including acceleration of the limb), and suspension (e.g. during swing phase of gait). All components of the lower extremity are subject to injury, which may lead to some period of non-weight bearing to facilitate healing. A few examples include broken or fractured bones, tendon ruptures, ligament tears, lacerations or disease.

Skeletal muscles are typically divided into two major groups, namely extensors and flexors. Extensors act to straighten a joint, while flexors act to bend it. An example of this would be the quadriceps (which include the vastus lateralis and rectus femoris), which are knee extensors. Of particular interest to this study are the vastus lateralis (knee extensor), rectus femoris (hip flexor and knee extensor), long head of the biceps femoris (knee flexor), and gastrocnemius medialis (plantar flexor) (Figure 2-1 [22, 23]). These muscles play some of the

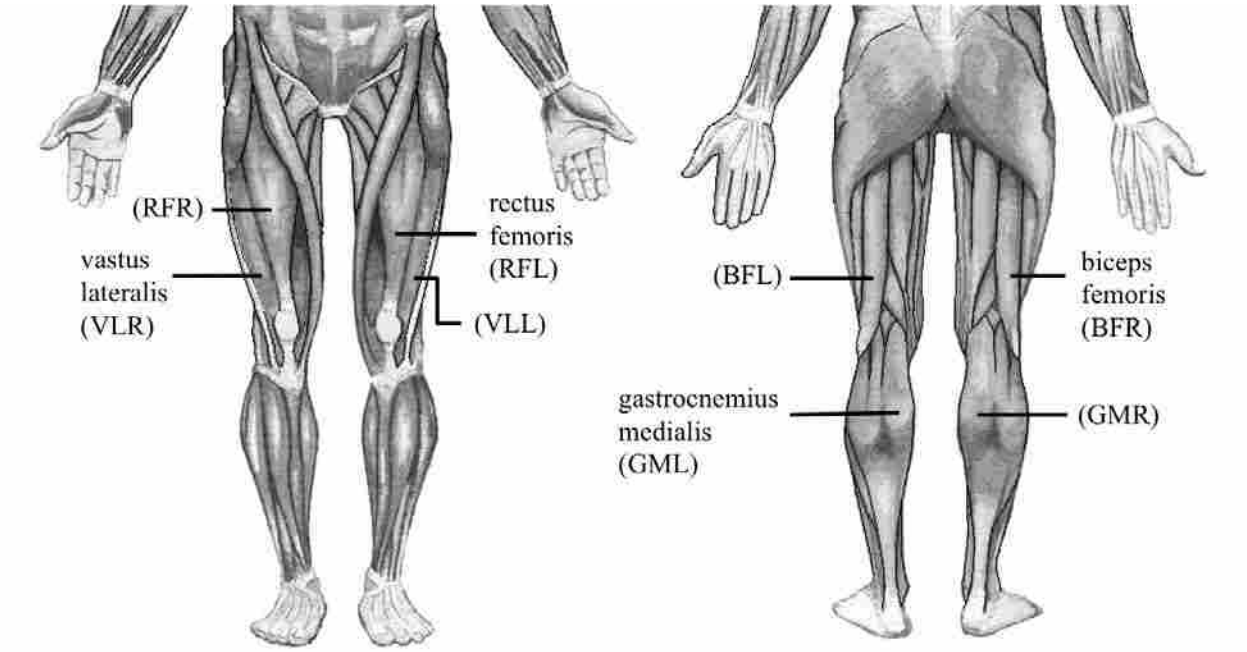

Figure 2-1: Anatomy of the Lower Extremities

most significant roles in the lower extremity for locomotion. Muscles may produce isometric (no motion), eccentric (contraction while lengthening), and concentric (contraction while shortening) contractions.

2.2 Electromyography and normal walking

A muscle is made up of many muscle fibers. These fibers may have an orientation that is along the major axis of the muscle or they may be oriented at some angle off that axis (called pennation angle). Contractions of these fibers are controlled by motor neurons, which are activated by the central nervous system. An electrical impulse is transmitted down the motor neuron to the neuromuscular junction where it affects the flow of ions (influx of Na+ then efflux of K+) across the sarcolemma. The action potential propagates along the sarcolemma and electromyography (EMG) measures this electrical signal or action potential and is the sum of all

the active motor units. Greater amplitudes in EMG signals indicate a large number of muscle fibers being activated and an overall stronger contractile force generated in the muscle. However, the relationship between increased EMG amplitude and muscle force is typically not a linear one, except for in cases of isometric contraction under steady state conditions [21, 24].

EMG data may be gathered either through surface or indwelling EMG, however the present research utilizes surface EMG sensors, which are non-invasive. Several factors may influence the quality of an EMG signal obtained with surface EMG. Among these are the interface between the electrode and skin (dead skin, oils or other contaminants can effect the resistivity of the skin), adipose (increased adipose decreases signal quality), muscle length (as length increases the frequency is shifted down), and muscle depth (when using surface EMG deep muscles cannot be effectively studied). Crosstalk occurs when the electrode is inadvertently sampling multiple muscle groups at once and can be another source of error. Adipose can increase the risk of crosstalk, with women and children being more susceptible to this. Ambient noise can also affect signal quality. One method of mitigating this is to reject signals that are common to both inputs in a biopolar EMG electrode. This is called common mode rejection and it operates under the assumption that true signals will arrive at slightly different times as the impulse travels down the fiber, where as noise will arrive at the same time and can therefore be rejected. CMRR is the common mode rejection rate and gives an indication of how effective an EMG electrode will be at this. Typical frequencies for EMG are in the range from 10-400 Hz, which illustrates the need for such a method given that a great deal of ambient noise is approximately 60 Hz [24].

Often EMG signals are normalized to maximum isometric contraction or to the average from some portion of the gate cycle to remove variability, especially if the study will span

several data collection sessions. However when the study consists of a single visit and reporting absolute value is possible, studies have shown this to be valid and reliable and can be more meaningful than reporting based on normalization [24].

EMG has been used as a tool in numerous studies investigating muscle function and a summary of the basic timing and function of the vastus lateralis (VL), rectus femoris (RF), long head of the biceps femoris (BF), and gastrocnemius medialis (GM) may prove helpful. Walking gait is comprised of stance (~0-60%) and swing phases (~60-100%). Some key features are initial contact, midstance, preswing and toe off. Terminal swing occurs just before initial contact. VL is active from just before initial contact, during terminal swing, in preparation for loading. It then handles loading of the leg and finishes a bit after initial contact in midstance to straighten the leg. RF is active with VL during initial contact, but to a lesser degree. It can also be active just before and during toe off in preswing where it limits knee flexion and helps hip flexion. RF continues the preswing functions into initial swing. BF is active just before and at initial contact, during terminal swing, working to decelerate the shank. It also contributes to the loading response in controlling hip flexion. Activity may also be observed just before toe off during preswing to encourage knee flexion if the subject is walking slowly. GM is active after initial contact, starting during midstance and typically peaking at toe off. During midstance it acts to control tibial rotation over the ankle joint. As progress is made towards terminal stance, the muscle transitions to isometric contraction, which causes the heel to rise from the ground and the knee to remain in extension. During preswing, GM contracts concentrically to push off from the ground [25-30].

2.3 The nature of skeletal muscle atrophy

A better understanding of skeletal muscle atrophy has been a major focus of previous research and review. Investigators have looked at whether there was inhomogeneous atrophy within a particular muscle [31], whether structural changes occur due to atrophy and what they are [32-34], and how those structural changes affect its ability to function [32, 34, 35]. Focus has also been placed on the level and effects of atrophy for various periods of disuse and age of subjects [3, 36-39] as well as the neurological effects of atrophy on strength and control of the muscle [37, 40] and every day function [41]. Attention has been given to considering the causes, effects and ways to prevent atrophy [19, 42, 43], what regulates muscle mass during periods of atrophy, exercise and rehabilitation [44, 45], and the rate of muscular atrophy compared to atrophy of other material such as tendons [46].

It is well know that bone and muscle in the body adapt to the stresses and strains they experience [21]. Generally, in the absence of normal loading, muscles will reduce in size as well as undergo architectural changes such as changes in pennation angle or muscle shortening [5, 36, 46]. There are also neurological changes and changes in a fiber's capacity to produce force [10, 37, 42, 46]. These other neurological and architectural changes also contribute to loss of strength [10, 37, 42, 46]. Depending on the extent of atrophy and whether or not there is intervention, full recovery of strength may take longer than the actual NWB period [10, 21, 31].

Atrophy is partly characterized by a reduction in cross sectional area (CSA) of the fibers [2]. This atrophy is relatively rapid at first and then tapers off with major changes occurring within the first 2 weeks [3, 10]. Significant atrophy can occur within 5 days [3], while the extent of atrophy has also been shown to vary along the length of the muscle [31]. Average loss of

muscle mass is approximately 0.4-0.6%/day with onset of atrophy beginning within 12-48 hours [3-5]. Average per day losses in strength are around 1.3%/day [3, 4].

Declines in strength have been shown to be more rapid than declines in size during atrophy. Loss of muscle quality and neurological degradation has been shown to play a key role in this [5, 9, 18, 37]. Change in muscle structure also occurs rapidly (within 14 days) with disuse. It is believed that the changes help compensate for loss of mass to maintain some strength [33, 35, 36]. The level of atrophy is thought to be driven by the daily function of the particular muscle prior to unloading and is muscle group specific. If the muscle was involved in antigravity (maintaining posture and standing) the effects were more drastic such as is seen in the knee extensor muscles and plantar flexor muscles [21, 36, 40].

2.4 Artificially inducing and measuring atrophy

In order to facilitate the study of atrophy, researchers have had to develop and validate ways to induce atrophy and measure it. Magnetic resonance imaging (MRI) has been validated as one way to effectively track atrophy [2] while unilateral lower limb suspension has been shown to effectively induce atrophy [47, 48]. Other methods of data gathering include various strength tests, biopsy, ultra sound, computed tomography (CT) and electromyography (EMG). The most common are MRI, various strength tests and biopsy. These tools are used most commonly to gather information on cross-sectional area (CSA), muscle volume, muscle fiber length, pennation angle, force, and neurological activity. Lengths of disuse ranged from 5 days to 90 days while number of subjects within a group ranged from 5 to 17. The most commonly studied muscles were the quadriceps (especially vastus lateralis), the gastrocnemius and the soleus. Several methods of imposing disuse where employed, which included casting and the use of crutches, bed rest and unilateral lower limb suspension (ULLS). ULLS is accomplished by placing a shoe

with a thicker sole on the load-bearing leg such that the unloaded leg can hang free and go through full range of motion without contacting the ground or engaging the muscles of the unloaded leg [47, 48]. Crutches are also employed for ULLS. Appendix A may be referenced for further detail on methods of inducing and measuring atrophy.

2.5 Methods of preventing or treating atrophy

Research has also been done on potential methods of recovering from atrophy as well as preventing it. For example, several have look at the effects of exercise and retraining both in old and young subjects on recovery from disuse atrophy [5-7]. In an attempt to reduce atrophy, others have looked at using whole body vibration [8], periodic blood flow restriction [9-11], neuromuscular electrical stimulation (NMES) [4, 20], resistance exercise [12-17], nutrition [16, 19], and motor imagery [18].

Older individuals experience more drastic effects neurologically while younger people are more heavily impacted muscularly. Older individuals also do not seem to benefit as much from retraining after disuse [7]. These deficiencies among the older population are more clearly manifested at higher contractile velocities [38]. Periodic restricted blood flow can have a positive effect on strength and size [9-11]. For short-term disuse, neuromuscular electrical stimulation may help prevent loss of muscle mass, but does not help preserve strength [4, 20]. While relatively small bouts of resistive exercise during periods of unloading have limited benefit on preventing atrophy (application in space flight) [13, 17], more intensive resistance exercise can have a marked positive effect [15]. On the other hand, whole body vibration has not been show to significantly reduce atrophy [8] nor has motor imagery been shown to have an effect on preservation of strength [18]. Understanding that the balance between protein synthesis and

protein breakdown drives atrophy or hypertrophy, proper nutrition is an important part of maintaining a positive balance and thus preventing atrophy [16, 19].

2.6 Electromyography and atrophy

As has been mentioned, skeletal muscle adapts to meet changes in the demands placed upon it. If decreased loads are sustained muscle will atrophy. If loads are increased above a threshold stimulus hypertrophy will occur. This threshold is based on the max force generating capacity of a give muscle (overload principle). Hypertrophy is different than atrophy in that the time required for measurable change is approximately six weeks. Strength gains do occur within the first few weeks, however these are due to improvements in muscle function such as increased motor unit recruitment, coordinated contraction within the muscle and the individual's ability to contract the muscles in a coordinated manner. After six weeks, increases in muscle size are noticeable and contribute to strength increases. Also important to recall is that the degree to which a muscle will atrophy or hypertrophy is dependent on the function of the muscle (e.g. knee extensors tend to atrophy more that knee flexors) [21].

Changes in the loads placed on muscle elicit changes in the electrochemical stimuli sent from the CNS to control muscle contraction [21]. It follows that changes in the stimuli correspond to ensuing adaptations in the skeletal muscle such as atrophy or hypertrophy. EMG measures this stimulus and can thus detect changes in the stimuli. Care should be taken to recall that adaptations in muscle such as atrophy and especially hypertrophy occur over days or weeks and therefore will precede changes in the stimulus as measured by EMG. It should also be remembered that the occurrence and degree of muscle adaptations will depend on several factors such as the magnitude of the applied loads, whether or not the change in load is long enough to induce atrophy or hypertrophy, and the particular muscles studied to name a few.

2.7 Motion capture

Motion capture systems typically consist of equipment for capturing ground reaction forces (GRF) and motion data via markers and cameras. Such systems are used in many places for a variety of purposes, such as in company research labs like the Nike ® Innovation Kitchen or in the film industry. These reflective markers are placed on anatomically significant sites, such as joint rotation centers and on limb segments so that motion in 3-dimensional space can be characterized and digitized. This information combined with GRF can be analyzed by software and estimates on individual muscle forces can be made.

Some motion capture systems consist of passive reflective markers and cameras which emit an infrared (IR) signal, that signal is then reflected off the markers and detected by the camera. Because the purpose of the markers is to represent the motion of the skeletal system, they are typically placed on boney landmarks that are near the surface of the body.

Cameras capture motion occurring in 3-demensional (3D) space as a 2-dimensional (2D) image and therefore cannot fully describe the motion. A common method for obtaining a 3D representation of motion from cameras is direct linear transformation also know as DLT. This method provides two linear equations relating marker position data from a camera to the desired coordinates in 3D space for any given marker. Given that the coordinates for 3D space represent three unknowns and there are only two equations, more equations are needed. Two more equations for any given marker can be obtained by having data for that marker from a second camera. Hence a marker must remain visible to a minimum of two cameras at all times.

In practice, more than two cameras are needed because at any given time one or more cameras may not have line of sight on a given marker, thus reducing the effective number of cameras available to accurately determine marker position. Additionally, in order to reduce error,

it is recommended that enough cameras be used such that three or more cameras can view a marker at all times. Another important consideration for minimizing error is camera placement. Cameras are often placed in a circle surrounding the motion capture volume and care should be taken to ensure they are not placed too close to each other or all in the same plane.

Ultimately the desire is to represent the motion of a body segment in 3D space. To do so, the body segment is assumed to be rigid and methods for rigid body kinematics are applied. A global coordinate frame is established during setup. Using the data obtained from a minimum of three markers, three unit vectors can be determined which define the axes of a segment-fixed coordinate frame. The position and orientation of this segment-fixed frame relative to the global frame can now be determined. If less than three markers are used orientation and position will not be fully defined. For example, if one marker was used then position would be known but orientation would not.

Theoretically, three markers are all that is needed, however this is dependent on two assumptions, namely that the body segment was actually rigid and that no errors exist in the marker positions. Skin and underlying soft tissue lead to relative motion between markers (relative marker error) making the rigid body assumption false. This motion of the markers also leads to absolute marker error meaning the marker no longer represents the boney landmark that it is supposed to. Using rigid body marker clusters is one method of attenuating relative marker error. These clusters affix three or more markers to a rigid material, which is then attached to the body segment.

Other errors can be the result of poor marker placement or a poorly defined marker set (e.g. markers too close together, markers do not define a plane or are along a single line). Calibration errors have been a source of error, however they have been reduced through the use

of software and calibration wands. During calibration the wand "paints" the motion capture volume. Digitization is part of post processing wherein the raw data is correlated to virtual anatomical markers in the model throughout the motion of interest. Errors in digitization can be mitigated by following the recommendations previously mentioned and through the use of spherical markers (apparent shape is independent of viewing angle) and markers that are as large as is practical for a given study [21].

2.8 Current mobility aids

The purpose of any mobility aid is to provide the conditions necessary for the affected portion of the body to heal while still allowing some degree of mobility. When bone is broken or a ligament or muscle is damaged or not functioning properly, loading of that portion of the body needs to be removed temporarily. A very common device used to facilitate this is the crutch (Figure 2-2 A). In the case of the crutch, ground reaction forces generated during ambulation are transferred up through the crutch to the arm and through the shoulder complex. This removes loading from the entire suspended leg whether or not unloading of the entire leg is necessary. For example, in the case of a foot or ankle injury, unloading of the upper leg is not needed. Another device in common use is the wheelchair. This device may be used in cases where the injury has affected both lower extremities (bilateral injuries) or where weight bearing through the wrist or shoulder complex is not feasible. Such may be the case where the injuries sustained included upper extremities or disease or other malformities have rendered those extremities unable to bear weight.

In both of these devices some portions of the body are unnecessarily unloaded and atrophy is allowed to occur, which might have otherwise been prevented. Understandably, isolating only that portion of the body which needs to be unloaded and maintaining typical

Figure 2-2: Crutches (A), knee scooter (B), and a temporary-injury prosthesis or TI prosthesis (C) used in the study.

function of all other portions may be difficult, however it should be clear that such should be regarded as the gold standard if minimizing atrophy is the goal. More recently, devices that increase mobility and allow unloading of the lower leg only have been developed and released to the market (Figure 2-2 B and C). One of these devices, the knee scooter (sometimes called knee walker) allows the patient to kneel on a pad while moving around through rolling on the cart to which the pad is attached. Unlike crutches, loading still occurs in the upper portion of the leg. Another device, termed a hands-free crutch alternative or temporary injury prosthesis (TI prosthesis), attaches to the leg like a prosthesis and supports the leg in a fashion similar to the knee scooter. However, unlike the scooter, this device allows the patient to achieve mobility through "walking" around on the device.

3 EFFECTS OF VARIOUS MOBILITY AIDS ON LOWER-EXTREMITY MUSCLE ACTIVITY

3.1 Introduction

There are over 4.5 million visits to the emergency room for below-the-knee injuries each year in the U.S. [1]. These injuries are commonly followed by a 6-8 week period of non-weight bearing facilitated by the use of one or more different mobility aids (Figure 2-2), which offload the body weight from the injured extremity to other parts of the body. Each type of mobility aid likely results in different loading conditions on the muscles, which lead to variations in disuse muscle atrophy. Depending on the extent of atrophy and whether or not there is intervention, full recovery of strength may take longer than the actual non-weight bearing period [5, 31]. Other consequences of non-weight bearing include decreased muscle function, changes in metabolism, muscle insulin resistance and increased body fat mass [3, 49]. These consequences, along with rapid and dramatic loss of muscle mass and strength, provide a rationale for investigating muscle activity during use of these devices [3-5, 10].

It is well known that muscle adapts to functional load [21]. In the absence of normal loading, muscles usually become less massive and undergo architectural changes such as changes in pennation angle or muscle shortening [5, 36, 46]. There are also neurological changes and changes in a fiber's capacity to produce force [5, 37, 42, 46]. These neurological and architectural changes contribute to loss of strength [5, 37, 42, 46]. Atrophy affects many muscles that may or may not be near the immobilization site [50]. Average loss of muscle mass

during non-weight bearing is approximately 0.4-0.6%/day with onset of atrophy beginning within 12-48 hours [3-5]. Average per day losses in strength are around 1.3% per day [3, 4]. Justifiably, attention has been given to considering causes, effects and ways to mitigate or prevent atrophy [19, 42, 43].

Despite what is currently understood about muscle adaptations due to unloading, most of the methods used to counteract or prevent atrophy have shown partial or limited success [4-20]. Based on well established understanding of muscle activation and adaptation mechanisms, it should be clear that the more closely muscle function during non-weight bearing matches normal muscle function, the more likely it is that muscle adaptations, negative or positive, will be avoided [21]. One way to accomplish this is to increase understanding of how different mobility aids used during non-weight bearing ambulation affect muscle activation; this increased understanding will assist (1) healthcare professionals in recommending a mobility aid for a particular patient based on rehabilitation needs and (2) assist designers of mobility aids by providing a protocol for evaluation of a design against it's potential to mitigate muscle atrophy.

The purpose of this work was to evaluate the effects of 3 types of mobility aids on lower-extremity muscle activity. Evaluation was based on 1) measured muscle activation signals using electromyography (EMG), and 2) measured joint kinematics and ground reaction forces, which were used to predict muscle forces for a representative subject. We hypothesized that among the mobility aids tested, crutches will be least effective at maintaining muscle EMG activation and predicted muscle forces relative to normal walking.

3.2 Methods and materials

3.2.1 Subjects

Nine healthy male and seven healthy female subjects participated in the study. Height, weight, and age were self-reported at the time of the visit. Average age was 21.3 years (range 18-27 years), average height was 177.3 cm (range 154.9-203.2 cm), and average weight was 66.9 kg (range 45.4-89.8 kg). Average BMI was calculated to be 21.2. Exclusion criteria included (1) Participation in organized strength-training within 6 months prior to data collection, (2) Current or previous musculoskeletal injuries affecting the upper- or lower-extremities, (3) Current or previous musculoskeletal disease, (4) chronic disease with regular clinical treatment, (5) regular drug, alcohol or tobacco intake, and (6) any metabolic or hormonal disorder. All subjects were right leg dominant as determined by their choice of leg to forcefully kick a soccer ball. The study was conducted in a biomechanics lab on our university campus. Prior to data collection, approval was obtained from the appropriate institutional review board, and all subjects gave informed consent prior to data collection.

3.2.2 Setup and equipment

All data were collected in a single visit, which lasted between 1.5-2 hours. A member of the research team demonstrated each device and instructed use of the device in accordance with manufacture recommendations. A device was then properly sized and fit to the subject. In every case the right leg was treated as the injured or involved leg and was suspended or supported by the device. The three devices that were presently tested included: traditional axillary crutches (Medline part number MDS80534HW-$P; Figure 2-2 A), a knee scooter (Roscoe Medical ROS-

KSBG; Figure 2-2 B) and a temporary-injury prosthesis (TI prosthesis; FlexLeg model# FL009-A; Figure 2-2 C).

Prior to data collection, the subjects practiced using each mobility aid. During this time, subjects were asked to rate their level of confidence on a scale from 1-10, with 1 representing insufficient confidence to use the device and 10 representing the same confidence they would have while walking on their own two feet. Subjects practiced using the device until they achieved a confidence level of 7 or higher. Subjects were not allowed to proceed to data collection until they and the primary researcher deemed the subject to be proficient.

Once subjects had been trained on all devices, they changed into spandex exercise clothing and athletic shoes. Subjects were then prepped for surface electromyography (EMG) sensors that were placed in accordance with standard practices [51]. Wireless surface EMG sensors with a common mode rejection ratio >80 dB and sampling at 2500 Hz were used (Trigno Wireless, Delsys Inc, Boston, MA, USA). The analog EMG signals were first band-pass filtered (20-450 Hz) and amplified at a gain of 1000. VICON Nexus (Santa Rosa, CA, USA) and Delsys software were used for capturing all data, which included synchronized surface EMG, reflective marker position and ground reaction force (GRF) data. Two 60 × 90 cm force plates embedded in a lab floor were used (1000 Hz; AMTI, Watertown, MA, USA). EMG Sensors were placed bilaterally on the vastus lateralis (VL), rectus femoris (RF) biceps femoris long head (BF), and gastrocnemius medialis (GM). These muscles play important roles in ambulation and are commonly studied in relation to mobility and atrophy [36, 42, 48]. Manual muscle tests were performed to ensure proper placement and function of the wireless sensors. Then, the sensors were further secured with Powerflex tape.

High-speed video cameras (100 Hz; VICON) were used to record spatial position of 22 individual reflective markers and four rigid body marker clusters (four markers per cluster). After surface EMG sensors were placed, motion markers were placed on subjects. Eleven individual markers and two rigid body marker sets were placed bilaterally over the following locations: 4 markers on foot (dorsum, head of fifth metatarsal, third distal phalanx, and posterior calcaneus), 2 markers on ankle (lateral and medial malleolus), 2 markers on knee (lateral and medial femoral condyle), 2 markers on hip (anterior superior iliac spine and posterior superior iliac spine), 1 marker on greater trochanter, and 2 rigid body marker clusters (one on the anterior thigh and one on the distal lateral shank). After the reflective markers were placed, subjects stood still while a static trial was recorded.

3.2.3 Data collection

There were 8 trial types: involved and uninvolved leg for each of the three devices as well as walking (Figure 3-1). Data were being captured for both legs whether it was an uninvolved or involved leg trial. The designation "uninvolved leg" or "involved leg" simply refers to which leg struck the force plate for any particular trial. For the trials involving the various devices, "involved leg" means that the device contacted the force plate. At least five successful trials of each trial type (a set) were captured. A member of the research team demonstrated and explained how each trial type should be conducted before the subject began that trial type. Orange cones were set up on both sides of the force plate to mark where the subject was to start and stop. Subjects were instructed to not target the plate, but to walk or ambulate on a device as felt natural and to not slow down until they reached the opposite cone. Members of the research team then adjusted cone location until the correct foot or portion of the device, depending on the trial type, fully contacted the force plate. The subject was not

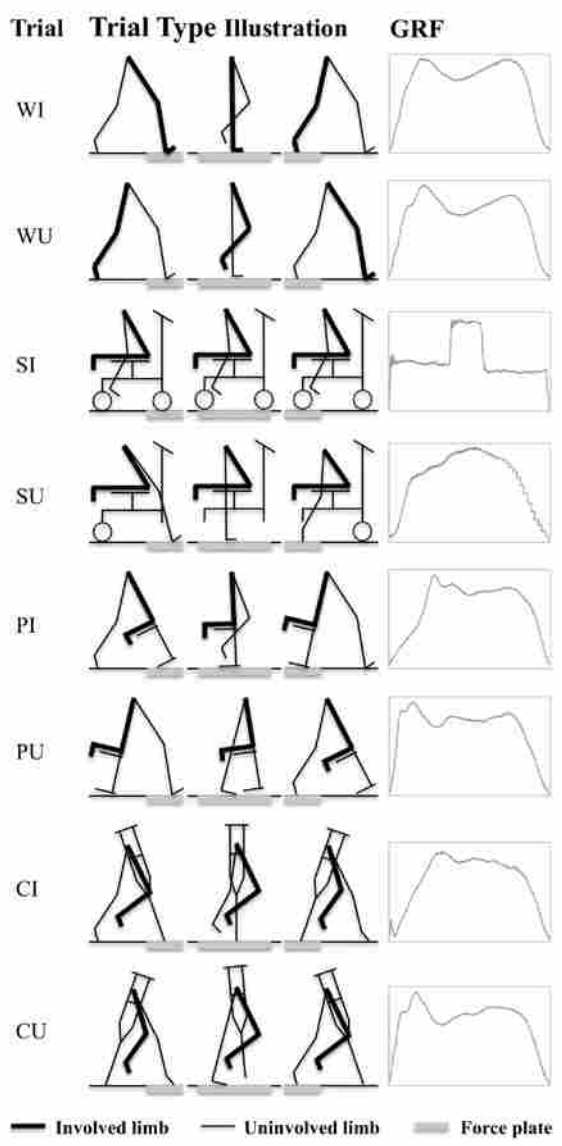

Figure 3-1: The study was comprised of 8 trial types defined by which device was used and whether the involved limb (or device) or uninvolved limb contacted the force plate. Namely there was walking involved and uninvolved (WI, WU), knee scooter involved and uninvolved (SI, SU), TI prosthesis involved and uninvolved (PI, PU), and crutches involved and uninvolved (CI, CU). Note that for SU the knee scooter wheels do not touch the force plate at anytime during the trial, only the left leg engages the plate.

necessarily informed of whether a trial was considered good or bad. Rather they were instructed to continue going back and forth between cones unless requested to stop or they became uncomfortable for any reason. The order in which devices (including walking) were tested was randomized.

Walking

With both feet lined up at a cone, the subject started from a dead stop and stepped forward with whichever foot felt most natural and walked until they reached the opposite cone, the involved foot having contacted the force plate. The instructions for an uninvolved-leg trial were the same as for the involved leg, however, cone positions were adjusted so that the uninvolved leg fully contacted the force plate.

Devices

An involved-leg crutch trial included a suspended right leg (as is typically done). The subject line up the crutches and their left foot at a cone and started from a dead stop, then performed traditional crutch ambulation until the opposite cone was reached. Both crutch tips contacted a single force plate. The uninvolved-leg crutch trials were similar, except that uninvolved foot contacted the force plate rather than the crutches.

Trials for the TI prosthesis were the same as for walking, except that for involved-leg trials it was the device that contacted the force plate, rather than the actual foot.

Involved-leg knee scooter trials began with the scooter lined up at a cone. Starting from a dead stop and pushing off with the left foot, the subject pushed until shortly before the scooter made initial contact with the force plate, at which point they were required to coast with the left leg suspended (as is typically done with these devices) until they reached the opposite cone. An uninvolved-leg trial differed in that the subject pushed with the left foot, which was required to complete one single plant and push-off on the force plate; during this trial, the knee scooter did not contact any force plate.

3.2.4 Data analyses

Surface EMG data

Raw EMG data were read into Matlab for processing, which included the DC offset removal and smoothing via a root mean square algorithm (window width = 52 ms; [25, 52-54]). For all trials, smoothed EMG was plotted for each of the eight muscles and stacked above the corresponding GRF, in a single window, using Matlab. These windows were used to subjectively check and validate the data as it was processed.

Ground contact, for each trial, was determined via the vertical GRF. Mean EMG amplitude, across all of ground contact, was calculated. This resulted in eight means, one for each muscle, for each trial. The average of all these means for a given muscle (e.g. VLR) across all subjects for a given trial type was calculated. This resulted in an 8x8 matrix (8 trial types by 8 muscles). The average of that was then taken for involved and uninvolved leg trials for a particular muscle (e.g. VLR) for a particular device (e.g. crutches) so as to represent both states the leg would be in through a full cycle on a particular device. This resulted in a 4x8 matrix (4 devices by 8 muscles) of the aggregate (across all subjects) average EMG amplitude and was used to produce the column graph shown in Figure 3-2. Another column graph was produced by taking the values from Figure 3-2 for each device relative to the values for walking. This was done by subtracting the value for a given device and muscle from the corresponding muscle for walking and is represented by Figure 3-3. For more detail on formulas and code used in excel and matlab to accomplish this, please refer to appendix C and appendix B respectively.

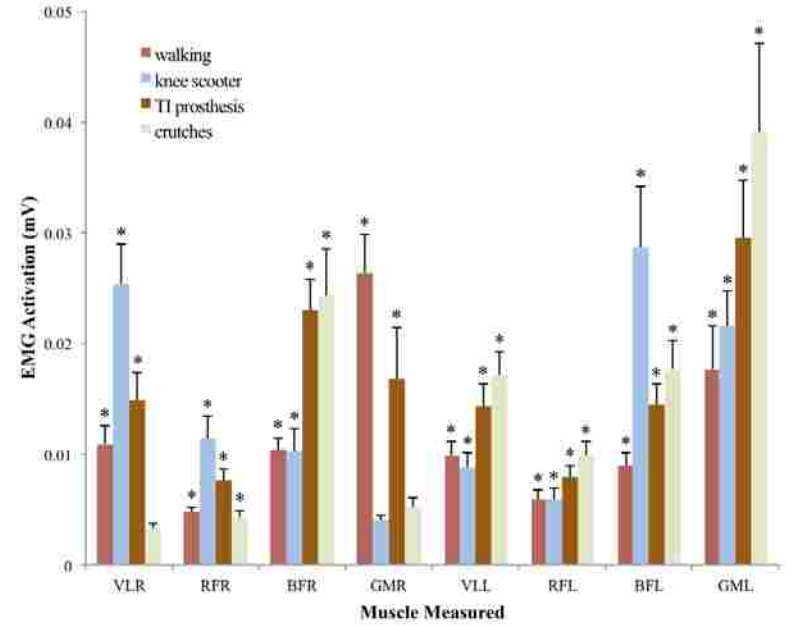

Figure 3-2: The mean EMG amplitude during ground contact of both involved and uninvolved trial types were calculated and averaged together. Asterisk indicates $p < 0.01$.

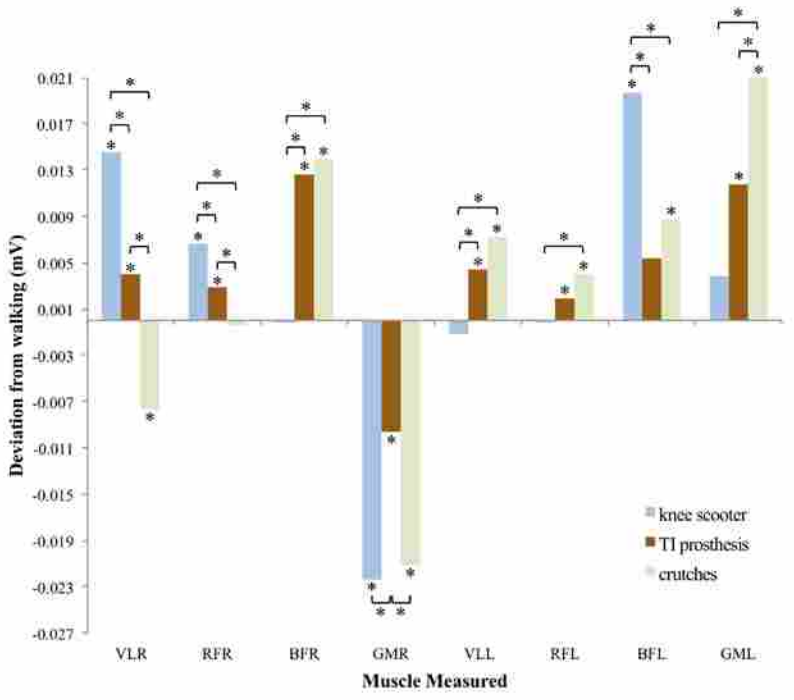

Figure 3-3: Values for EMG activation for a given device and muscle were subtracted from the corresponding muscle for walking. Positive values indicate an increase in EMG activation compared to walking while negative values indicate a decrease.

Processing motion data

All motion data processing was done using OpenSim from Simtk (OpenSim v3.2). Motion data captured with Vicon were converted to a file format compatible with OpenSim using Lee-Son ToolBox v1.51 software. A standard model for OpenSim (FullBodyModel_Hamner 2010_v2_0.osim) was downloaded from simtk.org and modified to create 3 models for use with each of our treatments, namely walking, crutches, TI prosthesis, and knee scooter. The standard model was modified to match our marker set described in section 2.2. This modified model was used for both walking and knee scooter trials. Because the knee scooter does not leave the ground nor is it attached to the subject, its mass was not added to any body segment and the same model used for walking trials could be used for knee scooter trials. Additional modifications for the crutch model included adding half of the mass of the crutch to the left humerus and the other half to the right humerus, while the TI prosthesis model was obtained by adding the mass of the device to the right tibia. The tibia and femur of the models were scaled to match the subject data based on marker positions and the mass was set to the subject's measured mass. GRF data measured by the force plate during each trial were applied to the simulation as a point force at the measured center of pressure location. For uninvolved limb trial types the location was always applied to the calcaneus. For involved limb trial types the location varied by mobility aid. For walking trials it was applied to the calcaneus, for crutch trials to the pelvis and for TI prosthesis and knee scooter trials to the tibia. Inverse kinematics were calculated by OpenSim, followed by calculation of computed muscle control. 6 Hz was used for filtering kinematics within OpenSim during computed muscle control operations.

Statistical analysis

Statistical analysis was performed on the aggregate EMG data for each device as compared to walking. A mixed model analysis of covariance blocking on subjects was used, because we have repeated measures for each subject across devices, which causes correlation in the data. Because we did pairwise comparisons we also employed a Tukey-Kramer adjustment for multiple comparisons.

3.3 Results

Figure 3-4 shows a subset of the calculated muscle force from OpenSim temporally aligned and superimposed on the corresponding measured EMG. Specifically, this figure represents the muscles of the involved leg for each involved-leg trial type for a representative subject. From subjective comparison to all subjects, this figure represents well what was observed across all subjects for EMG. No statistical analysis was done on muscle force data. Aggregate average EMG amplitude is shown in Figure 3-2 along with the standard error, providing an indication of the inter-subject/inter-trial variability. Statistical significance versus zero signal is indicated with an asterisk. This figure displays the data for all muscles and devices including walking, on a per muscle per device basis. Figure 3-3 shows the differences between muscle activation for each device as compared to walking. Values that are positive indicate greater EMG activation than walking, while negative values indicate EMG activation that is less than walking.

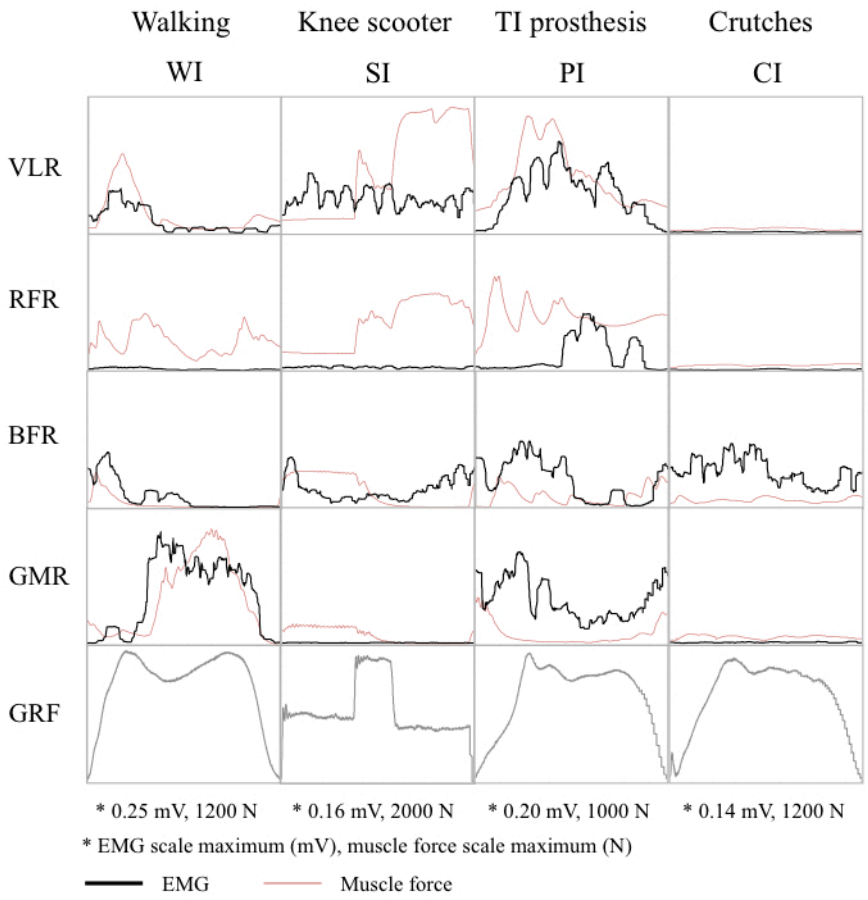

Figure 3-4: Calculated muscle force temporally aligned and superimposed on the corresponding measured EMG. Specifically it is a subset including the muscles of the involved leg for the involved-leg trial types for walking (WI), scooter (SI), TI prosthesis (PI), and crutches (CI) for a representative subject. It begins at initial ground contact and ends at final ground contact with the vertical scale set on a per device basis.

3.3.1 EMG activation for devices compared to walking

The knee scooter experienced increased EMG activation compared to walking for VLR ($p < 0.01$), RFR ($p < 0.01$), and BFL ($p < 0.01$). There was decreased activation for knee scooter for GMR ($p < 0.01$). Crutches experienced increased activation for BFR ($p < 0.01$), VLL ($p < 0.01$), RFL ($p < 0.01$), BFL ($p < 0.01$), and GML ($p < 0.01$). Decreases for crutches were observed for VLR ($p < 0.01$) and GMR ($p < 0.01$). For the TI prosthesis, decreases were found

only in GMR ($p < 0.01$) and to a lesser degree than either knee scooter or crutches. Increases were experienced in VLR ($p = 0.0429$), RFR ($p < 0.01$), BFR ($p < 0.01$), VLL ($p < 0.01$), RFL ($p = 0.0131$), and GML ($p < 0.01$). Statistically significant differences are denoted in Figure 3-3 by an asterisk.

3.3.2 EMG activation comparisons between devices

Differences in EMG activation between knee scooter and TI prosthesis were significant for VLR ($p < 0.01$), RFR ($p < 0.01$), BFR ($p < 0.01$), GMR ($p < 0.01$), VLL ($p < 0.01$), and BFL ($p < 0.01$). When comparing TI prosthesis to crutches, significant differences were observed for VRL ($p < 0.01$), RFR ($p = 0.0217$) GMR ($p < 0.01$), and GML ($p = 0.0260$). Comparison between knee scooter and crutches showed significant differences for VLR ($p < 0.01$), RFR ($p < 0.01$), BFR ($p < 0.01$), VLL ($p < 0.01$), RFL ($p < 0.01$), BFL ($p = 0.0283$), and GML ($p < 0.01$). Statistically significant differences for between device comparisons are denoted in Figure 3-3 by brackets for the comparison and an asterisk.

3.4 Discussion

We hypothesized that among the mobility aids tested, crutches would be least effective at maintaining muscle EMG activation and muscle forces relative to normal walking. Results of this study confirmed that crutches proved least effective at maintaining average EMG activation magnitudes. Predicted muscle forces from OpenSim suggested that muscle forces for the uninvolved limb were similar in patterns and timing to those of walking. OpenSim predictions for the involved limb during crutch ambulation were deemed unreliable, largely due to the lack of direct ground reaction force inputs on the involved limb.

Although there is a relationship between EMG and muscle force, there is not a linear relationship in most cases (isometric contraction with muscles in steady state is one exception) [21]. Therefore we felt it was valuable to look at muscle force as well and provide our findings. Predictions made by OpenSim on muscle force tended to be most consistent with measured EMG data for muscles on the leg in contact with the ground. In these cases, good correlation in pattern and timing between predicted muscle force and EMG is seen in these muscles for walking and device trials (i.e. right leg muscles for WI and left leg muscles for WU, SU, PU, and CU). Additionally, walking trial data matched well with EMG patterns, timing and muscle force magnitudes found in literature, which provides increased confidence that results for other devices are valid [25, 26, 55, 56]. Given that the sample size for muscle force predictions were small, these findings should be taken as preliminary.

Conclusions on which device may be best depends on what outcomes are deemed most important to the clinician or designer. The concept of keeping muscle EMG and muscle forces close to walking during healing of lower-limb injuries itself, is somewhat controversial, however for most patients, recovery from atrophy post-healing is assumed to be a significant hurdle. Thus, we identify three quantitative outcomes that could be considered. 1) Minimizing deviation from walking in total muscle EMG activation. Considering all muscles with statistically significant differences compared to walking in Figure 3-3, deviation may be taken as the sum of absolute differences for all muscles for a particular mobility aid. In this case the TI prosthesis deviates the least (0.047 mV) followed by the knee scooter (0.063 mV) with crutches as most deviant (0.084 mV). 2) Minimizing deviation from walking, with deviation taken as the total number of muscles with a significant difference from walking for muscle EMG activation. In this case the scooter deviates the least (four of eight muscles) followed by the TI prosthesis and

crutches both of which showed seven of eight muscles deviating from walking. 3) Minimizing muscle losses in the unloaded shank, while encouraging the maintenance or increase of EMG activation in other areas. Based on this criteria, the TI prosthesis proved significantly better than crutches or the knee scooter and maintaining activation in the unloaded shank while still maintaining or increasing activation in all other muscles that were measured.

In addition to the quantitative criteria identified above, there are several qualitative considerations that a clinician may consider in recommending a particular mobility aid. It has been shown that muscles which are statically or isometrically loaded experience strength gains centered about the joint angle at which it was loaded, whereas if the muscle is dynamically loaded (i.e. loaded throughout the range of motion of intended use) strength gains will be realized through the entire range [57]. In light of this, the desired outcome may be to ensure that the muscles are exposed to dynamic loading as much as possible. Observation of how the tested mobility aids are used during ambulation may suggest that the TI prosthesis ensures the greatest degree of dynamic loading. Both the TI prosthesis and crutches display similar function as walking for the uninvolved limb. However, based on subjective observation, the TI prosthesis also requires the RF to perform similar range of motion as walking. In the case of scooters while coasting, both the involved and uninvolved limbs are being statically loaded. Other considerations which may drive desired outcome might include existing injuries to important weight-bearing structures (e.g., shoulder or upper leg injuries), as well as potential for secondary injuries. For example, a patient may be predisposed to injures to the arms, wrist or shoulder complex.

Patterns and timing for the results of the devices correspond well with observed use of the device and expressed muscle soreness upon initial use. Those muscles which are shown to

have significant increase in activation are the same muscles which tend to get sore or which seemed to contribute more significantly to antigravity, propulsion, stabilization or suspension as a result of ambulation on a particular device. Muscles with decreased activation correspond to those muscles with an anticipated decreased functional role.

3.4.1 Knee scooter

It was observed that when using a knee scooter, subjects tended to lean back, which would require additional activation of the VLR and RFR to counteract this while VLR, RFR and BFR would all contribute to stabilization. The right shank was supported by the knee scooter, which would render GMR effectively inactive. VLL and RFL play an antigravity role upon planting of the left foot prior to push off so maintenance of activation should be expected. BFL experiences increased activation due to its role in propulsion as the leg is used in a pulling motion similar to a skateboard. It also prevents the knee from hyperextension during this movement. Additionally it plays a role in keeping the leg suspended via knee flexion during coasting. GML participates in propulsion during push off and thus maintains similar activation levels. These activities are consistent with measured EMG activation data for the knee scooter. Thus, we anticipate that the VLR, RFR and BFL would experience muscle hypertrophy or maintenance, and GMR would experience muscle atrophy while the remaining evaluated muscles would not experience significant change.

3.4.2 TI prosthesis

At initial ground contact and during initial loading of the pseudo foot, the device creates a moment, which the VLR must control and counteract to prevent a decrease in knee flexion angle. This is seen as increased activation of the VLR. Control of the hip angle as the upper body

accelerates due to this moment along with adaptation for the extra weight of the device may be a reason for increased RFR activation. BFR is especially active in controlling the device through controlling right leg knee flexion angle. Specifically when the device is off the ground or during initial ground contact and final ground contact, BFR must act to maintain an approximate right angle in the knee.

Because all three devices provide for non-weight bearing of the shank, it was assumed that the EMG activation for GMR would be similarly reduced among them. However, for the TI prosthesis GMR activation remained significantly higher than crutches or knee scooter although still reduced compared to walking. This may be explained by isometric contraction as the entire right leg is stiffened to control the device. Such rigidness was visually observed among many subjects as they ambulated on the device. This isometric contraction while maintaining non-weight bearing may be considered good or bad depending on the type of injury. For example, isometric contraction could increase blood flow and neurological maintenance, while potentially increasing force placed on tendons or cartilage.

Given that the uninvolved limb performs a similar function during ambulation on the TI prosthesis as in walking, the muscles of the uninvolved limb should appear similar in pattern and timing as those in walking trials. This occurrence is seen in the results along with increased activation for VLL and RFL. The increased activation may be explained by the extra stabilization and increased time on the left leg due to an overall slower gait, the extra weight of the device and the lack of a knee or ankle joint. Some have reported soreness in the GML upon initial use, which is consistent with the data. This is likely because propulsion is primarily provided for by the GML and left leg. Such patterns are seen in measured EMG activation data for TI prosthesis. Consequently, for the TI prosthesis we expect hypertrophy or maintenance in

the VLR, RFR, BFR, VLL, RFL and GML muscles, with atrophy in the GMR muscle and no significant change in BFL.

3.4.3 Crutches

In the case of crutches, the entire right leg is non-weight bearing. The hip and knee maintain flexion so as to keep the foot suspended from the ground. As a result there is a marked decrease in VLR, while RFR, and to a much greater extend BFR, are active in maintaining hip and knee joint flexion angles respectively to suspend the foot. GMR is effectively inactive as a result of non-weight bearing. As in the case of the TI prosthesis, the uninvolved limb during crutch ambulation performs a similar function as in walking, and therefore the muscles of the uninvolved limb should also appear similar in pattern and timing as those in walking trials. This was observed and increases were noted in the data for VLL, RFL, BFL and GML. These increases are consistent with the left leg being involved in increased stabilization when the crutches are off the ground as well as playing a major role in propulsion and controlling landing upon initial ground contact of the left foot. Accordingly, we would expect hypertrophy or maintenance to be most likely experienced among the BFR, VLL, RFL, BFL, and GML muscles. GMR and VLR muscles are most likely to experience atrophy while the RFR muscle would see no significant change.

Previous studies have most commonly used ULLS or bed rest to induce atrophy. Therefore, location and degree of muscle adaptation will differ from what would occur using crutches in an injury setting, making direct correlation to this study difficult, however some limited comparisons may be made. ULLS supports the entire lower limb in suspension without requiring activation of the muscles of the lower limb to maintain suspension [47, 48]. Accordingly, atrophy is experienced in VL, RF, BF and GM on the suspended limb [5, 15, 46,

48]. This is in partial contrast to the present study in which decreased activation is only seen in VL and GM. However, this may be explained by differences in methodology, where in the present study the subject was required to use the RF and BF at all times to maintain suspension of the leg as would be done in the case of an actual injury. This is supported by the results of another study involving traditional suspension of the lower extremity due to an injury, in which maintenance of the BF and RF were observed in the patient [50]. Studies also showed maintenance of the uninvolved limb [15, 48]. This is in agreement with the present study in which the results would suggest maintenance or hypertrophy of the uninvolved limb. The occurrence of measurable hypertrophy depends on whether the increase in load is significant enough (as compared to maximum contractile capacity) and sustained for long enough (>6 weeks) [21]. None of the previous study's methods were conducted over an adequately long period as well as sufficiently similar to the present study to make further comparisons regarding hypertrophy.

3.4.4 Limitations and future work

Inputs used for OpenSim were limited to marker data and GRF. Therefore, OpenSim had no way of accounting for isometric contractions. This may be one explanation for why the correlation between EMG and muscle force, as calculated by OpenSim, was typically better for muscles on the leg that was in contact with the force plate for any particular trial type versus the leg that was not or that was engaged in a device.

Because this study measured variations in muscle activation across devices at a single time point, rather than a longitudinal study of non-weight bearing using these devices, definitive statements regarding atrophy or hypertrophy cannot be made. Instead we focus on EMG data as potential stimulus for muscle atrophy/hypertrophy. Additionally, this study was limited to

measurements of 4 muscles on each leg. While these muscles are well known to play significant roles in normal gait and experience adaptations to unloading [36, 42, 48], they are not the only muscles involved in gait. Indeed, some of the devices evaluated in this study may induce unexpectedly high muscle activation of alternative muscle groups. Therefore, the full scope of the effect of these devices on muscle adaptations cannot be captured by this study. Pain or other factors may affect muscle atrophy, however to reduce variability, the current study was conducted using healthy subjects.

Different mobility aids may require a different number of steps or motions to ambulate over a given distance (step-cycle frequency). A difference in step-cycle frequency may lead to more or less muscle activity over a period of non-weight bearing on a particular mobility aid. Our study focused on the muscle activity for one step-cycle and the relative increases or decreases as compared to walking.

Future work would include the use of these devices to induce atrophy through a period of non-weight bearing (a minimum of 2 weeks suggested for atrophy [3, 10], greater than 6 weeks for hypertrophy [21]). MRI and strength tests could be used before and after the non-weight bearing period to confirm a correlation between EMG activation and muscle atrophy. A similar study done on injured subjects could uncover the effects of that variable on atrophy as it relates to these devices. A more expansive study on what muscles in the body are affected by the use of these devices is also recommended. Future work would also include an investigation into the variations in step-cycle frequency between the difference mobility aids.

3.5 Conclusions

This study provides useful bilateral data that describe measured lower-extremity EMG activation amplitudes and muscle force predictions based on kinetic data during ambulation

using three different mobility aids, compared to normal walking on a per step-cycle basis. Each of the tested mobility aids resulted in statistically different EMG activation amplitudes, for different lower-extremity muscles, relative to one another and relative to normal walking. Good correlation was also observed between predicted muscle force and EMG for muscles on the leg in contact with the ground for the representative subject. These differences in lower-extremity muscle activity may lead to different levels of atrophy or hypertrophy. Based on a criteria of maintaining muscle activation, the TI prosthesis proved most effective among the devices tested, while the crutches proved least effective. The data presented will be valuable to clinicians in providing insight into which mobility aid may be best suited for a particular patient. It is anticipated that these data will provide designers of mobility aids with a protocol for evaluation of designs based on their potential to cause or prevent muscle adaptations.

4 OBSERVATIONS ON MUSCLE FORCE AND EMG

To further validate the results of EMG data, motion capture data together with GRF data for a representative subject was used to calculate estimates on individual muscle force contributions. These estimates were made using OpenSim; a software-based analysis tool that has been used in many studies focused on gait. These estimates could then be compared with corresponding EMG data for the representative subject as well as what is known from literature on patterns, timing and magnitude. A subset of this was presented in chapter 3, which focused on the right leg muscles of the involved limb trial types. Presented here are all muscles for all trial types with observations on EMG patterns, a comparison of muscle force to EMG, and estimates on individual muscle force contributions.

4.1 About the graph

Figure 4-1 shows the calculated muscle force temporally aligned and superimposed on the corresponding measured EMG for a representative subject. It represents each of the 8 trial types and provides insight into the patterns and timing for EMG that was observed among the subjects. Graphs for all the EMG data of a representative subject were generated and comparisons within trial type were made for this subject as well as comparisons with what is know from literature for normal walking gait. From these graphs, eight were selected and then used to make subjective comparisons with corresponding generated graphs of all trails for all

subjects. EMG for both walking trials and devices trials match well with what is known from literature and observed device use, lending confidence to the results [25, 26, 55, 56].

By looking at WI and WU trial types, both states the leg will be in over a full cycle for walking may be seen. The same observations can be made for the devices by looking at the involved (e.g. CI, etc.) and uninvolved (e.g. CU, etc.) trial types for those devices. As a clarifying example, WI represents stance phase for muscles on the involved limb (VLR, RFR, BFR, GMR) and WU represents stance phase for muscles on the uninvolved limb (VLL, RFL, BFL, GML). The concept of stance phase and swing phase loses meaning as upon considering the device trials, but as much as can make sense, these devices follow the same phases as explained for walking. For example it might be said that CI represents stance phase for the muscles of the involved limb (VLR, RFR, BFR, GMR).

Because the vertical scale for each device and walking was independently set, observations on relative magnitudes among muscles within a device or walking can be made. By noting the scale values listed at the bottom Figure 4-1, magnitude comparisons can also be made between devices and walking. Noting this also provides insight as activation patterns for one muscle are compared to another within walking or device type. GRF is aligned along the bottom and provides context for the timing of activation.

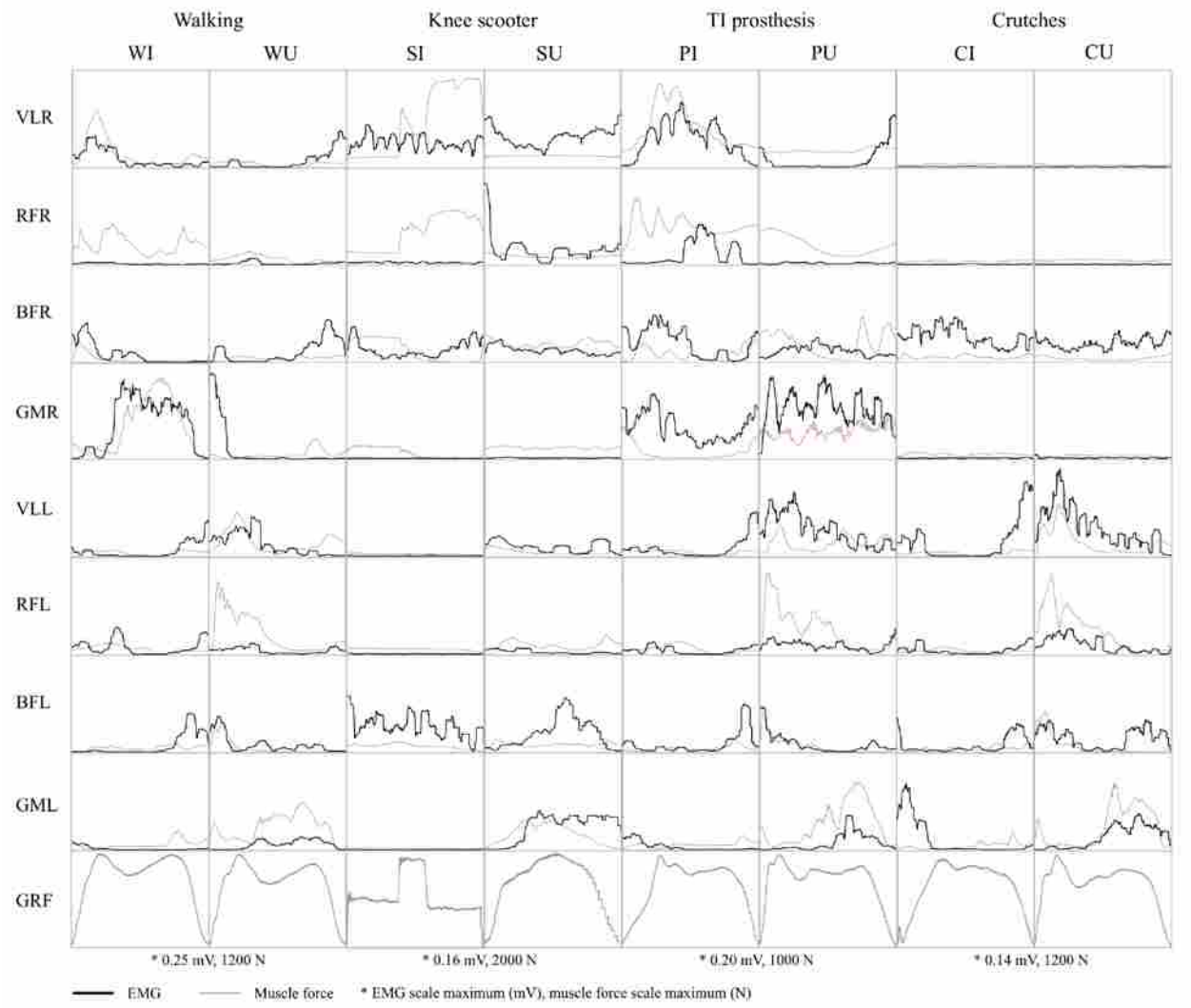

Figure 4-1: Calculated muscle force temporally aligned and superimposed on the corresponding measured EMG for each of the eight trial types for a representative subject. It begins at initial ground contact and ends at final ground contact with the vertical scale set on a per device basis.

4.2 Observations on EMG patterns

Supposing the subject exhibited perfect symmetry in gait between left and right legs, the muscles of the right leg (e.g. VLR, etc.) during WI trials would match the muscles of the left leg (e.g. VLL, etc.) during WU trials. Further, the right leg muscles during WU trials would match the left leg muscles during WI trials. As we look Figure 4-1, a reasonably good match is noted.

Given that the left leg functions similarly for all but the knee scooter trials, muscles of the left leg should look similar for walking, TI prosthesis and crutches. This occurrence is also manifest Figure 4-1.

Insight into which muscles may atrophy or hypertrophy come as comparisons to walking are made across devices for a muscle. BFR is especially active for TI prosthesis and crutches, which are about twice the peak amplitude of the knee scooter when noting the relative scales and consistently more active than walking. This agrees well with how these devices are used requiring more effort from that muscle to maintain required knee flexion. Increased and sustained activity in GMR is noted for TI prosthesis as compared to the other devices. Increases in VLR and RFR are observed during SU trial type over SI, which fits well with the subject lowering their center of gravity and raising it to enable plant and push on the left foot. In the case of crutch trials, right leg muscles are very inactive with BFR as the only exception, supporting the observation that BFR is employed in maintaining knee flexion to keep the right foot suspended.

4.3 Comparison of muscle force to EMG

Results of any model are dependent on the inputs; therefore how closely it models the physical event relies on this. Inputs used in OpenSim for this study included GRF and motion capture data. Additional inputs were subject weight and marker data from a static pose, used to scale the model so as to represent the subject. OpenSim uses the hill-type muscle model, which accounts for the passive and active response of muscle tissue. Also included in the model is pennation angle.

For muscles on the leg in contact with the ground, it can be seen that the pattern matches well with EMG. For example, as magnitude in EMG increases and decreases there is typically a

corresponding rise and fall in calculated muscle force. When evaluating the correlation, it is beneficial to note that the scales for muscle force and EMG are independent and therefore the trace for muscle force will not necessarily align vertically with the EMG trace. For muscles on the leg that are not in contact with the force plate, the correlation breaks down. As mentioned in chapter three, this may be explained in part by considering the inputs given to OpenSim, which would limit its ability to account for isometric contractions.

4.4 Estimates on individual muscle force contributions

Determining individual muscle force contributions during gait continues to prove a challenge, however some estimates have been made [21]. The following values are given as multipliers of subject or model body weight and are peak values. One study estimated vastus lateralis at 0.68 and 0.86 with gastrocnemius medialis at 2 and 1.2 through two different methods [55]. Another study estimated all the vasti together at 1.75, rectus femoris at 0.29, all the muscles of the hamstring except the short head of the biceps femoris at 0.41, and the gastrocnemius at 1.44 [56]. Results of the current study estimate VLR at 1.02, RFR at 0.68, BFR at 0.45 and GMR at 1.47. In comparison, estimates from the current study were moderately higher. However, considering the difficult nature of making estimates on individual muscle force contributions and the variability that exists from one individual to another these results were deemed sufficiently good to have confidence in the methods employed and the results produce.

5 SUMMARY AND FUTURE WORK

5.1 Contributions and conclusions

Muscle activations on the lower extremities were measured during ambulation on 3 types of mobility aids using EMG. Each of the tested ambulatory aids resulted in statistically different EMG activation amplitudes, for different lower-extremity muscles, relative to one another and relative to normal walking. Based on a criteria of maintaining muscle activation on a per step-cycle basis, the TI prosthesis proved most effective among the devices tested, while crutches proved least effective.

Joint kinematics and GRF were measured during ambulation on 3 types of mobility aids and were used to predict muscle forces of the lower extremities for a representative subject. Muscle force predictions tended to be best for muscles on the leg while it was in contact with the ground. In these cases, muscle force predictions for walking data matched well with literature and with EMG results from this study. Good correlation was also observed in these cases between predicted muscle force and EMG for the devices. Where there was increase or decrease in EMG there was a corresponding increase or decrease in predicted muscle force.

These differences in lower-extremity muscle activity may lead to different levels of atrophy or hypertrophy. The data presented will be valuable to clinicians in providing insight into which mobility aid may be best suited for a particular patient. It is anticipated that these data will provide designers of mobility aids with a protocol for evaluation of designs based on their potential to cause or prevent muscle adaptations.

5.2 Future work

A future study would include the use of these devices by healthy subjects to induce atrophy through a period of non-weight bearing (a minimum of 2 weeks suggested [3, 10], greater than 6 weeks for hypertrophy [21]). EMG data would be gathered following the methods described in the present study, while MRI and strength tests could be used before and after the non-weight bearing period. The results could then be used to quantify the correlation between changes in EMG activation and changes in muscle size and strength.

The current work focused on healthy subjects. Given that non-weight bearing is often required as a result of injury, a similar study conducted on injured subjects is advised. Such a study would uncover the effects of injury on muscle activity as it relates to these mobility aids.

Another area of future work would employ the methods described in this research to investigate the effects of these mobility aids on other muscles involved in ambulation. The results of that research may also prove to be helpful in considering device recommendations and design.

Investigation into the variations in step-cycle frequency between the difference mobility aids is also recommended. Important elements of that study would include developing methods to standardize the distance and speed (e.g. subjects allowed to ambulate at self-determined pace or required to cover the distance in a set amount of time). Considerations would need to be given to the type of terrain to be covered (e.g. level walking, some incline or a mix) based on what might be most representative of daily living. Data collected for devices would be compared to walking to determine a multiplier, which could then be used with muscle activity data such as was obtained from the present study.

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APPENDIX A. METHODOLOGY OF STUDIES INVESTIGATING ATROPHY

Table A--1: Methodology of previous studies investigating atrophy of the lower extremities.

Reference	Length of study	Number of participants	Location on body studied	Data collection methods	Other notables
Campbell et al., 2013	3 weeks	16 young males (8 in control)	QF muscle, VL	biopsy; torque test	3 weeks ULLS followed by 3 week resistance training program
Kubota et al., 2008	2 weeks	15 healthy males	left ankle, thigh/leg	strength test; thigh/leg circumference; serum growth hormone levels	immobilized and NWB (use of casts and crutches)
(Miokovic et al., 2012	60 days	9 males	muscles from hip to ankle	MRI	60 days bedrest. absolute NWB
Seynnes et al., 2008	14 and 23 days	8 young healthy males	SOL, GM, GL	MVC, EMG, MRI, ultra sound	ULLS
Zange et al., 2009	14 days	8 healthy males	upper and lower leg muscles	MRI	6 degrees head down tilt bedrest
Wall et al., 2013a	5 and 14 days	24 young healthy males (5 days, n=12; 14 days, n=12)	quadriceps, VL	CT, DEXA, 1 rep max, biopsies on vastus lateralis	one-legged knee immobilization, use of full leg cast
Dirks et al., 2013	5 days	24 young healthy males (12 in control)	quadriceps	CT scan (CSA), strength test, biopsies	one-legged knee immobilization with or without supervised NMES sessions
Kubota et al., 2011	2 weeks	11 healthy males (6 in control)	knee extensor-flexor and ankle PF	torque test	use of casts and crutches NWB; one group having a blood flow restriction
Psatha et al., 2012	5, 8,15,29, and 43 days	18 (8 males, 10 females)	TA, GM, GL, SOL	3.0 Tesla MRI	one lower leg immobilized
Hotta et al., 2011	20 days	11 healthy males (5 in control)	plantar flexor	MVC, MRI	ULLS, non control group performed intense 25 minute interval cycling training up to 80% of peak oxygen uptake on alternate days
Suetta et al., 2009	2 and 4 weeks	9 elderly and 11 young males	quadriceps		2 weeks immobilization followed by 4 weeks retraining
Akima et al., 2009	20 days	11 healthy males (5 in control)	QF and thigh muscles	MVC, mfMRI	TRN performed intense interval training on alternate days
Deschenes et al., 2008	7 days	10 young and 10 elderly	KE, VL, patellar tendon	isometric strength, isokinetic contractions	ULLS
de Boer et al., 2008	5 weeks	10 healthy males	antigravity (GM, VL), non-antigravity (TA, BB)	ultrasonography	horizontal bed rest
de Boer et al., 2007	23 days	17 young males (8 in control)	KE, VL, patellar tendon	torque test, MRI, ultrasonography	ULLS
Clark et al., 2007	4 weeks	17 subjects	soleus	H-reflex, MRI, EMG	performed steady isometric PF and KE tasks and KE shortening and lengthening contractions (25% maximum intensity)
Clark et al., 2006b)	4 weeks	18 young subjects, 6 in subset	PF	measurements of central activation, the H-reflex, and nerve conduction	ULLS, subset of subjects (n=6) performed PF MI training 4 days/wk.
Clark et al., 2006a	4 weeks	18 young subjects, 6 in subset	PF	Measurements of voluntary and evoked forces, CMAP	a subset of subjects (n=6) received applications of Isc 3 days/wk.

key to abbreviations: QF=quadricep femoris ; VL=vastus lateralis; ULLS=unilateral lower limb suspension; CSA=cross sectional area; SOL=soleus; GM=gastrocnemius medialis; GL=gastrocnemius lateralis; KE=knee extensor; TA=tibialis anterior; BB=biceps brachii; PF=plantar flexor; NWB=non-weight bearing

Table A--1 continued

Reference	Length of study	Number of participants	Location on body studied	Data collection methods	Other notables
Rittweger et al., 2005	90 days	25 young healthy males	calf and forearm	pQCT to assess BMC and muscle CSA	strict bed rest with -6 degrees head-down tilt. 3 groups: 1) practiced resistive exercises, 2) received 60mg pamidronate i.v. prior to bed rest, 3) control group. Ca(++) an protein intake were controlled during study. Tests at 28, 89 days during bed rest; 14 days after recovery.
MacIntyre et al., 2005	6 weeks	12 (7 NWB, 5 healthy)	quadriceps		12 weeks recovery period
Tesch et al., 2004	5 weeks	21 (11 in control)	KE, PF	muscle strength, muscle volume measured by MRI, surface EMG	used unilateral lower limb unloading. Noncontrol group used unilateral KE resistance exercise.
Deschenes et al., 2002	2 weeks		Knee extensors and flexors	blood samples, muscle biopsies, muscle function data, EMG recordings	data analyzed before and after 14 days of muscle unloading
Narici and Cerretelli, 1998		8 males	GM	MRI and real-time ultrasound	patients had unilateral lower limb atrophy
Hather et al., 1992	6 weeks	8 subjects	thigh and calf muscles, VL	MRI, biopsy	right shoe outfitted with platform sole that prevented the left foot from bearing weight while walking with crutches, yet allowed freedom of movement about the ankle, knee, and hip. MRI pre and post ULLS
Berg et al., 1991	4 weeks	6 healthy males	quadriceps, thigh	strength tests, CT scan	ULLS, subjects on crutches. Tests before and after 4 weeks of suspension and after 4 days and 7 weeks of uncontrolled recovery.

key to abbreviations: QF=quadricep femoris ; VL=vastus lateralis; ULLS=unilateral lower limb suspension; CSA=cross sectional area; SOL=soleus; GM=gastrocnemius medialis; GL=gastrocnemius lateralis; KE=knee extensor; TA=tibialis anterior; BB=biceps brachii; PF=plantar flexor; NWB=non-weight bearing

APPENDIX B. MATLAB CODE

```
%an m file that will process (remove DC offset, rectify, and smooth) the emg and
%output the average emg activation between initial ground contact and final
%ground contact. reads in a file in using xlsread so that we don't have to trim the
%headers off the end of our csv files. Puts all the smoothed muscle data into one
%subplot with positive GRF.

clear; %clears out all variables
clc; %clears the matlab command window
close all; %closes all open figures
%% read in data and find the range of elements with only numbers
prompt = 'What is the file name (include file extension .csv)?';
filename = input(prompt,'s');

data = xlsread(filename); %import the grf and emg data for each control trial

%create a vector (we're calling it tf) of 1s and 0s to distinguish where
%there are numbers and text (NaN means not a number). isnan assigns a 1
%to NaN elements and 0 to numeric elements.
tf = isnan(data(:,3));

%locate the element location where there are only numeric values. you can
%then eliminate the cells that have text.
cnt1 = 4;
for i = 5:length(data)
        if tf(i,1) == 0
                cnt1 = cnt1 + 1;
        elseif tf(i,1) == 1
                break
        end
end

%create a matrix that now only contains numeric values
data = data(5:cnt1,1:28);

%% assign variables to data
westgrf = data(1:length(data),14); %distinguish the grf data from the west plate
eastgrf = data(1:length(data),5); %distinguish the grf data from the east plate
vlrraw = data(1:length(data),21); %distinguish the emg data for each muscle
rfrraw = data(1:length(data),22);
```

```
bfrraw = data(1:length(data),23);
gmrraw = data(1:length(data),24);
vllraw = data(1:length(data),25);
rflraw = data(1:length(data),26);
bflraw = data(1:length(data),27);
gmlraw = data(1:length(data),28);
```
%% prompt to determine which device so we know which plate to use and convert GRF to positive numbers
```
prompt = 'is the device for this trial a scooter (type 1 for yes and 0 for no)?';
device = input(prompt);
if device == 1
        grf = eastgrf.*(-1);
else
        grf = westgrf.*(-1);
end
```
%% process the raw emg data
```
%calculate and remove the mean offset and rectify
vlroff = (vlrraw - mean(vlrraw)).^2;
rfroff = (rfrraw - mean(rfrraw)).^2;
bfroff = (bfrraw - mean(bfrraw)).^2;
gmroff = (gmrraw - mean(gmrraw)).^2;
vlloff = (vllraw - mean(vllraw)).^2;
rfloff = (rflraw - mean(rflraw)).^2;
bfloff = (bflraw - mean(bflraw)).^2;
gmloff = (gmlraw - mean(gmlraw)).^2;

%define the sample rate for the data collection session
sr = 2500;

%perform smoothing with a rms moving window for each muscle
%define width of the moving window in ms (mw*2)
%example: if mw below is 25 then the moving window width is 50 ms.
%ours will be 52 milliseconds (ms). this avoids a non-integer of
%62.5 which arises if you use 25.
%see formula below where it has (mw*sr/1000).
mww = 52;
mw = mww/2;

%vlr
for i = (mw*sr/1000)+1:1:length(vlroff)-((mw*sr/1000)+1)
        vlrs(i) = sqrt(mean(vlroff(i-(mw*sr/1000):i+(mw*sr/1000))));
end
vlrs = vlrs'; % then we take the transpose of the smoothed data.

%rfr
for i = (mw*sr/1000)+1:1:length(rfroff)-((mw*sr/1000)+1)
        rfrs(i) = sqrt(mean(rfroff(i-(mw*sr/1000):i+(mw*sr/1000))));
end
rfrs = rfrs';
```

```
%bfr
for i = (mw*sr/1000)+1:1:length(bfroff)-((mw*sr/1000)+1)
        bfrs(i) = sqrt(mean(bfroff(i-(mw*sr/1000):i+(mw*sr/1000))));
end
bfrs = bfrs';

%gmr
for i = (mw*sr/1000)+1:1:length(gmroff)-((mw*sr/1000)+1)
        gmrs(i) = sqrt(mean(gmroff(i-(mw*sr/1000):i+(mw*sr/1000))));
end
gmrs = gmrs';

%vll
for i = (mw*sr/1000)+1:1:length(vlloff)-((mw*sr/1000)+1)
        vlls(i) = sqrt(mean(vlloff(i-(mw*sr/1000):i+(mw*sr/1000))));
end
vlls = vlls';

%rfl
for i = (mw*sr/1000)+1:1:length(rfloff)-((mw*sr/1000)+1)
        rfls(i) = sqrt(mean(rfloff(i-(mw*sr/1000):i+(mw*sr/1000))));
end
rfls = rfls';

%bfl
for i = (mw*sr/1000)+1:1:length(bfloff)-((mw*sr/1000)+1)
        bfls(i) = sqrt(mean(bfloff(i-(mw*sr/1000):i+(mw*sr/1000))));
end
bfls = bfls';

%gml
for i = (mw*sr/1000)+1:1:length(gmloff)-((mw*sr/1000)+1)
        gmls(i) = sqrt(mean(gmloff(i-(mw*sr/1000):i+(mw*sr/1000))));
end
gmls = gmls';
```
%% plot vertical ground reaction force and smoothed data
```
%Plot all muscles and ground reaction force.
figure('name','All Muscles');
figheight = 1100;%in pixels
figwidth = figheight*(.35);%in pixels
fromleft = 10;%in pixels
frombottom = 10;%in pixels
hFig = figure('outerposition',[fromleft frombottom figwidth figheight],...
        'position',[(fromleft+5) (frombottom+5) (figwidth-5) (figheight-5)]);
set(hFig,'NumberTitle','off'); % to hide the title

subplot(9,1,1);
plot(vlrs);
title('VLR');
set(gca,'Xtick',[]);
```

```
subplot(9,1,2);
plot(rfrs);
title('RFR');
set(gca,'Xtick',[]);

subplot(9,1,3);
plot(bfrs);
title('BFR');
set(gca,'Xtick',[]);

subplot(9,1,4);
plot(gmrs);
title('GMR');
set(gca,'Xtick',[]);

subplot(9,1,5);
plot(vlls);
title('VLL');
set(gca,'Xtick',[]);

subplot(9,1,6);
plot(rfls);
title('RFL');
set(gca,'Xtick',[]);

subplot(9,1,7);
plot(bfls);
title('BFL');
set(gca,'Xtick',[]);

subplot(9,1,8);
plot(gmls);
title('GML');
set(gca,'Xtick',[]);

subplot(9,1,9);
plot(1:length(data),grf);
title('GRF');
```

%% determine heel strike and toe off

```
%identify the sample number for initial ground contact (igc) or heel strike
cnt = 0; %create a place holder variable for counting (cnt)
checkcnt = 0; %create a place holder variable for performing check to ensure there are 25 rows of
non-zero
for i=1:length(grf)
        if checkcnt >= 25%if ALL rows in our nested loop were non-zero, checkcnt will equal 25 and
we've found our heel stike.
                cnt = i-1;%we set count equal to the location of the first non-zero that preceded a set
        of at least 25 non-zeros
                break%we break out of our loop.
```

```
        else
        for j= i:i+24 %we look through the vector "grf" 25 rows at a time and check for non-zeros
(e.g. i=1 we are looking from 1 to 25, i=2 we are looking from 2 to 26)
                if grf(j)== 0 %if we find a zero we set "checkcnt" to zero and end the nested loop.
                        checkcnt = 0;
                        break
                else
                        checkcnt = checkcnt + 1; %if we find a non-zero we increment checkcnt.
                end
        end
        end
end

igc = cnt;

%identify the sample number for toe off (to)
to = 0;
checkcnt2 = 0;
for k = igc:length(grf) %count from where you left off above.
        if checkcnt2 >= 10 %if ALL rows in our nested loop were zero, checkcnt2 will equal 10 and
we've found our heel strike.
                to = k-1;% we set "to" equal to the location of the first non-zero that preceded a set
        of at least 10 zeros
                break% we break out of our loop.
        else
        for m= k:k+9 %we look through the vector "grf" 10 rows at a time and check for zeros
                if grf(m)== 0 %if we find a zero we increment checkcnt2
                        checkcnt2 = checkcnt2 + 1;
                else
                        checkcnt2 = 0; %if we find a non-zero we set "checkcnt" to zero and end the
nested loop.
                        break
                end
        end
        end
end
```

%% calculate the mean amplitude between igc and to

```
%This goes into your spreadsheet for each muscle for each trial
% this finds the mean for just the range you found above.
maVLR = mean(vlrs(igc:to));
maRFR = mean(rfrs(igc:to));
maBFR = mean(bfrs(igc:to));
maGMR = mean(gmrs(igc:to));
maVLL = mean(vlls(igc:to));
maRFL = mean(rfls(igc:to));
maBFL = mean(bfls(igc:to));
maGML = mean(gmls(igc:to));
MeanAmplitudes = [maVLR maRFR maBFR maGMR maVLL maRFL maBFL maGML]
```

%% create a variable that contains all the smoothed EMG data

```
times = ((mw*sr/1000)+1:length(data)-1)*(1/sr);
times = times';
grfs = grf((mw*sr/1000)+1:length(grf)-1);
AllEMG = [times vlrs rfrs bfrs gmrs vlls rfls bfls gmls grfs];
%% save variables to an excel compatable file
%prompt user for what they would like to name the file that they will be
%saving the emg data to.
prompt = 'type the file name you would like to save the emg data to:';
filenamesave = input(prompt,'s');
%save command. syntax is save(filename,variables,fmt)
save(filenamesave,'AllEMG','-ascii')
%type('AllEMGTest.txt')%this reads out the contents of the file
```

APPENDIX C. EMG DATA TABLES FOR GRAPH GENERATION

The following represent the formulas used in excel to process mean EMG amplitude data from Matlab and the resulting tables of data used to produce the figures on EMG activation (Figure 3-2) and EMG activation relative to walking (Figure 3-3). Green cells designate positive values while red cells designate negative values.

C.1 Excel formulas sheet for EMG analysis

Cells reference other tabs for each trial type where the data from Matlab for every trial are stored. Recall that for this study, right was the involved limb and left was the uninvolved. Column B rows 30-33 show how the standard error was achieved and the same formula is repeated across columns C-I, but not displayed.

Table C--1 Excel formulas used to process mean EMG amplitude data from Matlab.

	A	B	C	D	E	F	G	H	I
1	**Aggregate Subject Data**								
2									
3	**Averaged across all subjects for each of the 8 trial types**								
4	Trial type	VLR	RFR	BFR	GMR	VLL	RFL	BFL	GML
5	walking right	='Walking Right'!B4	='Walking Right'!C4	='Walking Right'!D4	='Walking Right'!E4	='Walking Right'!F4	='Walking Right'!G4	='Walking Right'!H4	='Walking Right'!I4
6	walking left	='Walking Left'!B4	='Walking Left'!C4	='Walking Left'!D4	='Walking Left'!E4	='Walking Left'!F4	='Walking Left'!G4	='Walking Left'!H4	='Walking Left'!I4
7	scooter right	='scooter right'!B4	='scooter right'!C4	='scooter right'!D4	='scooter right'!E4	='scooter right'!F4	='scooter right'!G4	='scooter right'!H4	='scooter right'!I4
8	scooter left	='scooter left'!B4	='scooter left'!C4	='scooter left'!D4	='scooter left'!E4	='scooter left'!F4	='scooter left'!G4	='scooter left'!H4	='scooter left'!I4
9	prosthesis right	='prosthesis right'!B4	='prosthesis right'!C4	='prosthesis right'!D4	='prosthesis right'!E4	='prosthesis right'!F4	='prosthesis right'!G4	='prosthesis right'!H4	='prosthesis right'!I4
10	prosthesis left	='prosthesis left'!B4	='prosthesis left'!C4	='prosthesis left'!D4	='prosthesis left'!E4	='prosthesis left'!F4	='prosthesis left'!G4	='prosthesis left'!H4	='prosthesis left'!I4
11	crutch right	='crutch right'!B4	='crutch right'!C4	='crutch right'!D4	='crutch right'!E4	='crutch right'!F4	='crutch right'!G4	='crutch right'!H4	='crutch right'!I4
12	crutch left	='crutch left'!B4	='crutch left'!C4	='crutch left'!D4	='crutch left'!E4	='crutch left'!F4	='crutch left'!G4	='crutch left'!H4	='crutch left'!I4
13									
14	**averaged left and right for full gait (absolute EMG)**								
15		VLR	RFR	BFR	GMR	VLL	RFL	BFL	GML
16	walking	=AVERAGE(B5:B6)	=AVERAGE(C5:C6)	=AVERAGE(D5:D6)	=AVERAGE(E5:E6)	=AVERAGE(F5:F6)	=AVERAGE(G5:G6)	=AVERAGE(H5:H6)	=AVERAGE(I5:I6)
17	knee scooter	=AVERAGE(B7:B8)	=AVERAGE(C7:C8)	=AVERAGE(D7:D8)	=AVERAGE(E7:E8)	=AVERAGE(F7:F8)	=AVERAGE(G7:G8)	=AVERAGE(H7:H8)	=AVERAGE(I7:I8)
18	TI prosthesis	=AVERAGE(B9:B10)	=AVERAGE(C9:C10)	=AVERAGE(D9:D10)	=AVERAGE(E9:E10)	=AVERAGE(F9:F10)	=AVERAGE(G9:G10)	=AVERAGE(H9:H10)	=AVERAGE(I9:I10)
19	crutches	=AVERAGE(B11:B12)	=AVERAGE(C11:C12)	=AVERAGE(D11:D12)	=AVERAGE(E11:E12)	=AVERAGE(F11:F12)	=AVERAGE(G11:G12)	=AVERAGE(H11:H12)	=AVERAGE(I11:I12)
20									
21	**Take the difference compared to walking (relative EMG)**								
22		VLR	RFR	BFR	GMR	VLL	RFL	BFL	GML
23	walking	=B16-B16	=C16-C16	=D16-D16	=E16-E16	=F16-F16	=G16-G16	=H16-H16	=I16-I16
24	knee scooter	=B17-B16	=C17-C16	=D17-D16	=E17-E16	=F17-F16	=G17-G16	=H17-H16	=I17-I16
25	TI prosthesis	=B18-B16	=C18-C16	=D18-D16	=E18-E16	=F18-F16	=G18-G16	=H18-H16	=I18-I16
26	crutches	=B19-B16	=C19-C16	=D19-D16	=E19-E16	=F19-F16	=G19-G16	=H19-H16	=I19-I16
27									
28	**Standard error (average of the standard deviation for right leg trials and left leg trials divided by sqrt(n) where n=16)**								
29		VLR	RFR	BFR	GMR	VLL	RFL	BFL	GML
30	walking	=(AVERAGE('Walking Right'!B5,'Walking Left'!B5))/4	=(AVERAGE('Walking Righ	=(AVERAGE('Walking Righ	=(AVERAGE('Walking Righ	=(AVERAGE('Walking Righ	=(AVERAGE('Walking Righ	=(AVERAGE('Walking Righ	=(AVERAGE('Walking Righ
31	scooter	=(AVERAGE('scooter right'!B5,'scooter left'!B5))/4	=(AVERAGE('scooter right'	=(AVERAGE('scooter right'	=(AVERAGE('scooter right'	=(AVERAGE('scooter right'	=(AVERAGE('scooter right'	=(AVERAGE('scooter right'	=(AVERAGE('scooter right'
32	prosthesis	=(AVERAGE('prosthesis right'!B5,'prosthesis left'!B5))/4	=(AVERAGE('prosthesis rig	=(AVERAGE('prosthesis rig	=(AVERAGE('prosthesis rig	=(AVERAGE('prosthesis rig	=(AVERAGE('prosthesis rig	=(AVERAGE('prosthesis rig	=(AVERAGE('prosthesis rig
33	crutch	=(AVERAGE('crutch right'!B5,'crutch left'!B5))/4	=(AVERAGE('crutch right'!	=(AVERAGE('crutch right'!	=(AVERAGE('crutch right'!	=(AVERAGE('crutch right'!	=(AVERAGE('crutch right'!	=(AVERAGE('crutch right'!	=(AVERAGE('crutch right'!

C.2 EMG data tables: all subjects

Table C--2: Aggregate subject data tables for generating mean EMG activation graphs.

Aggregate Subject Data

Averaged across all subjects for each of the 8 trial types

Trial type	VLR	RFR	BFR	GMR	VLL	RFL	BFL	GML
walking right	0.01242021	0.00460769	0.00972902	0.04268302	0.00871836	0.00620579	0.01001657	0.00838235
walking left	0.00929785	0.00489271	0.01092228	0.01004374	0.01101027	0.00562198	0.00790617	0.02695128
scooter right	0.02874068	0.01165826	0.00824737	0.00401965	0.00361606	0.00513806	0.02930035	0.00331228
scooter left	0.02195769	0.01106385	0.01209456	0.00397732	0.01392335	0.00650636	0.02788453	0.03972872
prosthesis right	0.02031739	0.0087882	0.01712724	0.01293689	0.01165781	0.00743732	0.01613929	0.02002941
prosthesis left	0.00932965	0.00635702	0.02886906	0.0205368	0.01697614	0.00831681	0.01263778	0.03891766
crutch right	0.0033196	0.00444087	0.02648417	0.00552075	0.01388309	0.0096126	0.01695795	0.03255897
crutch left	0.00316084	0.00410493	0.02203363	0.00483469	0.02030228	0.0101048	0.01842558	0.04564345

averaged left and right for full gait (absolute EMG)

	VLR	RFR	BFR	GMR	VLL	RFL	BFL	GML
walking	0.01085903	0.0047502	0.01032565	0.02636338	0.00986431	0.00591389	0.00896137	0.01766681
knee scooter	0.02534919	0.01136105	0.01017097	0.00399848	0.0087697	0.00582221	0.02859244	0.0215205
TI prosthesis	0.01482352	0.00757261	0.02299815	0.01673684	0.01431698	0.00787707	0.01438854	0.02947354
crutches	0.00324022	0.0042729	0.0242589	0.00517772	0.01709268	0.0098587	0.01769177	0.03910121

Take the difference compared to walking (relative EMG)

	VLR	RFR	BFR	GMR	VLL	RFL	BFL	GML
walking	0	0	0	0	0	0	0	0
knee scooter	0.01449016	0.00661085	-0.0001547	-0.0223649	-0.0010946	-9.167E-05	0.01963107	0.00385369
TI prosthesis	0.00396449	0.00282241	0.0126725	-0.0096265	0.00445266	0.00196318	0.00542717	0.01180672
crutches	-0.0076188	-0.0004773	0.01393325	-0.0211857	0.00722837	0.00394481	0.00873039	0.0214344

Standard error (average of the standard deviation for right leg trials and left leg trials divided by sqrt(n) where n=16)

	VLR	RFR	BFR	GMR	VLL	RFL	BFL	GML
walking	0.0016268	0.0004524	0.00112976	0.00340344	0.00131068	0.0008466	0.00106171	0.003939
scooter	0.00351381	0.00209755	0.0020538	0.00049459	0.00133311	0.00113957	0.00561413	0.00323723
prosthesis	0.00246585	0.00112058	0.00272685	0.00468228	0.0020735	0.00103204	0.00190836	0.00523206
crutch	0.00041019	0.00060324	0.00426011	0.00080268	0.00210835	0.00131284	0.00252002	0.00787743

C.3 EMG data tables: representative subject (subject 9)

Table C--3: Representative subject (subject 9) data tables for generating the mean EMG activation graph in appendix C (Figure C--1).

Representative Subject Data

Averaged across all subjects for each of the 8 trial types

Trial type	VLR	RFR	BFR	GMR	VLL	RFL	BFL	GML
walking right	0.02587691	0.00463888	0.01925927	0.09231512	0.01894877	0.0172574	0.01540626	0.00344554
walking left	0.01916082	0.00502097	0.02212658	0.01603875	0.02859229	0.01293671	0.01742754	0.0130997
scooter right	0.04711716	0.0054217	0.0118748	0.00539205	0.0014683	0.0023672	0.02456119	0.00129121
scooter left	0.04950624	0.0248209	0.01963815	0.00273062	0.01923841	0.00660222	0.04161213	0.04358397
prosthesis right	0.05939604	0.01530515	0.01979266	0.03665857	0.02845196	0.01583102	0.01545686	0.00740519
prosthesis left	0.02154368	0.00839033	0.03710348	0.06117434	0.05164202	0.01914571	0.0185152	0.01525697
crutch right	0.00256817	0.00233602	0.04198445	0.00275229	0.02228762	0.01014451	0.01124045	0.00989043
crutch left	0.00167093	0.00187763	0.03465048	0.00380385	0.04193083	0.01665617	0.02225077	0.02473264

averaged left and right for full gait (absolute EMG)

	VLR	RFR	BFR	GMR	VLL	RFL	BFL	GML
walking	0.02251887	0.00482993	0.02069292	0.05417693	0.02377053	0.01509705	0.0164169	0.00827262
knee scooter	0.0483117	0.0151213	0.01575647	0.00406133	0.01035335	0.00448471	0.03308666	0.02243759
TI prosthesis	0.04046986	0.01184774	0.02844807	0.04891645	0.04004699	0.01748837	0.01698603	0.01133108
crutches	0.00211955	0.00210683	0.03831746	0.00327807	0.03210923	0.01340034	0.01674561	0.01731154

Take the difference compared to walking (relative EMG)

	VLR	RFR	BFR	GMR	VLL	RFL	BFL	GML
walking	0	0	0	0	0	0	0	0
knee scooter	0.02579283	0.01029138	-0.0049365	-0.0501156	-0.0134172	-0.0106123	0.01666976	0.01416497
TI prosthesis	0.01795099	0.00701781	0.00775514	-0.0052605	0.01627646	0.00239131	0.00056913	0.00305846
crutches	-0.0203993	-0.0027231	0.01762454	-0.0508989	0.0083387	-0.0016967	0.00032871	0.00903892

Standard error (average of the standard deviation for right leg trials and left leg trials divided by sqrt(n) where n=16)

	VLR	RFR	BFR	GMR	VLL	RFL	BFL	GML
walking	0.00068523	8.1023E-05	0.00091799	0.00152002	0.0006161	0.00120503	0.00083279	0.00030208
scooter	0.00347632	0.00131481	0.00158822	0.0006818	0.00084413	0.00039518	0.00318215	0.00123959
prosthesis	0.00106282	0.00118708	0.00305822	0.0068492	0.00138578	0.0009077	0.00082275	0.00038606
crutch	0.00015887	8.6288E-05	0.00169497	0.00015155	0.00149447	0.00113073	0.00066775	0.00087014

C.4 Mean EMG activation graph for the representative subject (subject 9)

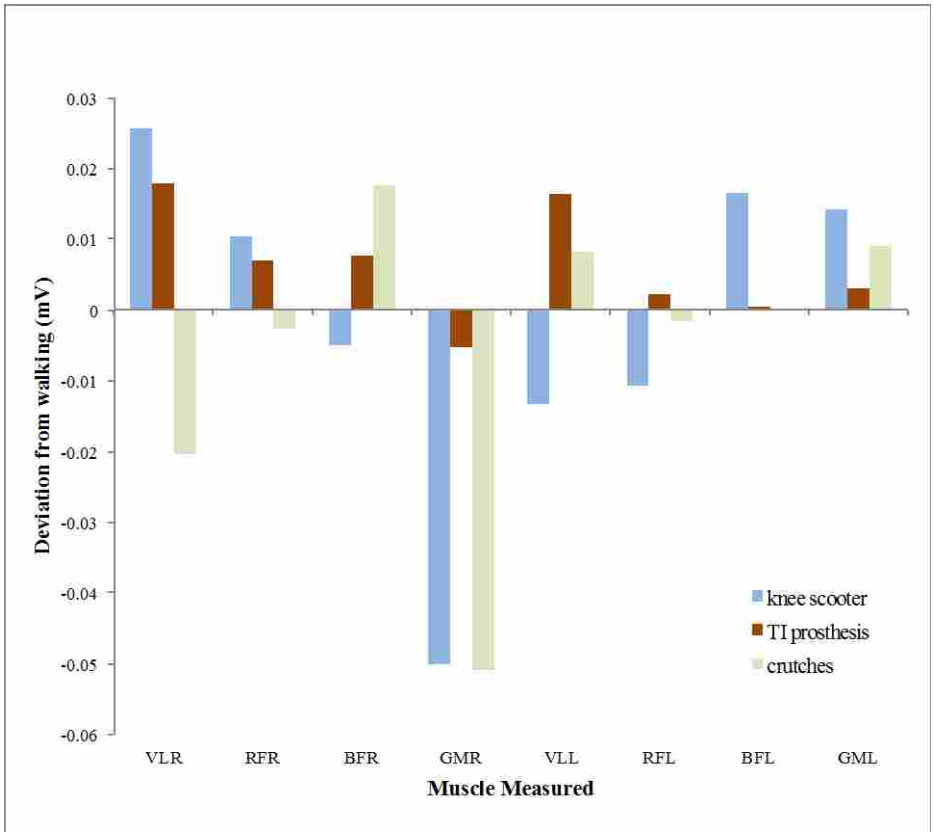

Figure C--1: Data shown is for the representative subject (subject 9). Values for EMG activation for a given device and muscle were subtracted from the corresponding muscle for walking. Positive values indicate an increase in EMG activation compared to walking while negative values indicate a decrease.

APPENDIX D. STATISTICS RESULTS

D.1 Statistics for graph representing mean EMG activation

The analysis for VLR with walking included
Least Squares Means

Effect	device	Estimate	Standard Error	DF	t Value	Pr > |t|
device	a	0.02169	0.004200	45	5.17	<.0001
device	c	0.006465	0.004200	45	1.54	0.1307
device	p	0.02965	0.004200	45	7.06	<.0001
device	s	0.05076	0.004200	45	12.09	<.0001

The analysis for RFR with walking included
Least Squares Means

Effect	device	Estimate	Standard Error	DF	t Value	Pr > |t|
device	a	0.009470	0.001880	45	5.04	<.0001
device	c	0.008521	0.001880	45	4.53	<.0001
device	p	0.01515	0.001880	45	8.06	<.0001
device	s	0.02285	0.001880	45	12.15	<.0001

The analysis for BFR with walking included
Least Squares Means

Effect	device	Estimate	Standard Error	DF	t Value	Pr > |t|
device	a	0.02059	0.004942	45	4.17	0.0001
device	c	0.04832	0.004942	45	9.78	<.0001
device	p	0.04600	0.004942	45	9.31	<.0001
device	s	0.02030	0.004942	45	4.11	0.0002

The analysis for GMR with walking included
Least Squares Means

Effect	device	Estimate	Standard Error	DF	t Value	Pr > |t|
device	a	0.05229	0.005320	45	9.83	<.0001
device	c	0.01032	0.005320	45	1.94	0.0586
device	p	0.03347	0.005320	45	6.29	<.0001
device	s	0.007971	0.005320	45	1.50	0.1411

The analysis for VLL with walking included
Least Squares Means

Effect	device	Estimate	Standard Error	DF	t Value	Pr > |t|
device	a	0.01969	0.003148	45	6.26	<.0001
device	c	0.03410	0.003148	45	10.83	<.0001
device	p	0.02863	0.003148	45	9.10	<.0001
device	s	0.01767	0.003148	45	5.61	<.0001

The analysis for RFL with walking included
Least Squares Means

Effect	device	Estimate	Standard Error	DF	t Value	Pr > |t|
device	a	0.01177	0.001738	45	6.77	<.0001
device	c	0.01978	0.001738	45	11.38	<.0001
device	p	0.01575	0.001738	45	9.06	<.0001
device	s	0.01168	0.001738	45	6.72	<.0001

The analysis for BFL with walking included
Least Squares Means

Effect	device	Estimate	Standard Error	DF	t Value	Pr > |t|
device	a	0.01781	0.005286	45	3.37	0.0016
device	c	0.03526	0.005286	45	6.67	<.0001
device	p	0.02878	0.005286	45	5.44	<.0001
device	s	0.05728	0.005286	45	10.84	<.0001

The analysis for GML with walking included
Least Squares Means

Effect	device	Estimate	Standard Error	DF	t Value	Pr > |t|
device	a	0.03503	0.009625	45	3.64	0.0007
device	c	0.07801	0.009625	45	8.11	<.0001
device	p	0.05895	0.009625	45	6.12	<.0001
device	s	0.04351	0.009625	45	4.52	<.0001

D.2 Statistics for graph representing mean EMG activation relative to walking

The analysis for VLR
Least Squares Means

Effect	device	Estimate	Standard Error	DF	t Value	Pr > |t|
device	c	-0.01523	0.003763	30	-4.05	0.0003
device	p	0.007954	0.003763	30	2.11	0.0429
device	s	0.02906	0.003763	30	7.72	<.0001

Differences of Least Squares Means

Effect	device	_device	Estimate	Standard Error	DF	t Value	Pr > |t|
device	c	p	-0.02318	0.004770	30	-4.86	<.0001
device	c	s	-0.04429	0.004770	30	-9.29	<.0001
device	p	s	-0.02111	0.004770	30	-4.43	0.0001

...Continued

Adjustment	Adj P
Tukey-Kramer	0.0001
Tukey-Kramer	<.0001
Tukey-Kramer	0.0003

The analysis for RFR
Least Squares Means

Effect	device	Estimate	Standard Error	DF	t Value	Pr > |t|
device	c	-0.00095	0.001669	30	-0.57	0.5742
device	p	0.005675	0.001669	30	3.40	0.0019
device	s	0.01338	0.001669	30	8.01	<.0001

Differences of Least Squares Means

Effect	device	_device	Estimate	Standard Error	DF	t Value	Pr > |t|
device	c	p	-0.00662	0.002340	30	-2.83	0.0082
device	c	s	-0.01432	0.002340	30	-6.12	<.0001
device	p	s	-0.00770	0.002340	30	-3.29	0.0026

...Continued

Adjustment	Adj P
Tukey-Kramer	0.0217
Tukey-Kramer	<.0001
Tukey-Kramer	0.0070

The analysis for BFR
Least Squares Means

Effect	device	Estimate	Standard Error	DF	t Value	Pr > |t|
device	c	0.02773	0.004959	30	5.59	<.0001
device	p	0.02541	0.004959	30	5.12	<.0001
device	s	-0.00028	0.004959	30	-0.06	0.9548

Differences of Least Squares Means

Effect	device	_device	Estimate	Standard Error	DF	t Value	Pr > |t|
device	c	p	0.002321	0.005305	30	0.44	0.6648
device	c	s	0.02802	0.005305	30	5.28	<.0001
device	p	s	0.02569	0.005305	30	4.84	<.0001

...Continued

Adjustment	Adj P
Tukey-Kramer	0.9002
Tukey-Kramer	<.0001
Tukey-Kramer	0.0001

The analysis for GMR
Least Squares Means

Effect	device	Estimate	Standard Error	DF	t Value	Pr > |t|
device	c	-0.04196	0.004780	30	-8.78	<.0001
device	p	-0.01881	0.004780	30	-3.94	0.0005
device	s	-0.04432	0.004780	30	-9.27	<.0001

Differences of Least Squares Means

Effect	device	_device	Estimate	Standard Error	DF	t Value	Pr > |t|	Adjustment	Adj P
device	c	p	-0.02315	0.006623	30	-3.50	0.0015	Tukey-Kramer	0.0041
device	c	s	0.002353	0.006623	30	0.36	0.7248	Tukey-Kramer	0.9329
device	p	s	0.02550	0.006623	30	3.85	0.0006	Tukey-Kramer	0.0016

The analysis for VLL
Least Squares Means

Effect	device	Estimate	Standard Error	DF	t Value	Pr > |t|
device	c	0.01441	0.002446	30	5.89	<.0001
device	p	0.008941	0.002446	30	3.66	0.0010
device	s	-0.00203	0.002446	30	-0.83	0.4141

Differences of Least Squares Means

Effect	device	_device	Estimate	Standard Error	DF	t Value	Pr > |t|	Adjustment	Adj P
device	c	p	0.005469	0.003459	30	1.58	0.1244	Tukey	0.2692
device	c	s	0.01644	0.003459	30	4.75	<.0001	Tukey	0.0001
device	p	s	0.01097	0.003459	30	3.17	0.0035	Tukey	0.0095

The analysis for RFL
Least Squares Means

Effect	device	Estimate	Standard Error	DF	t Value	Pr > |t|
device	c	0.008019	0.001512	30	5.30	<.0001
device	p	0.003988	0.001512	30	2.64	0.0131
device	s	-0.00008	0.001512	30	-0.05	0.9571

Differences of Least Squares Means

Effect	device	_device	Estimate	Standard Error	DF	t Value	Pr > |t|
device	c	p	0.004031	0.001893	30	2.13	0.0415
device	c	s	0.008101	0.001893	30	4.28	0.0002
device	p	s	0.004070	0.001893	30	2.15	0.0398

...Continued

Adjustment	Adj P
Tukey-Kramer	0.1008
Tukey-Kramer	0.0005
Tukey-Kramer	0.0968

The analysis for BFL
Least Squares Means

Effect	device	Estimate	Standard Error	DF	t Value	Pr > |t|
device	c	0.01744	0.005731	30	3.04	0.0048
device	p	0.01096	0.005731	30	1.91	0.0653
device	s	0.03947	0.005731	30	6.89	<.0001

Differences of Least Squares Means

Effect	device	_device	Estimate	Standard Error	DF	t Value	Pr > |t|
device	c	p	0.006480	0.008105	30	0.80	0.4303
device	c	s	-0.02202	0.008105	30	-2.72	0.0108
device	p	s	-0.02850	0.008105	30	-3.52	0.0014

...Continued

Adjustment	Adj P
Tukey	0.7062
Tukey	0.0283
Tukey	0.0039

The analysis for GML
Least Squares Means

Effect	device	Estimate	Standard Error	DF	t Value	Pr > |t|
device	c	0.04299	0.005689	30	7.56	<.0001
device	p	0.02392	0.005689	30	4.21	0.0002
device	s	0.008479	0.005689	30	1.49	0.1465

Differences of Least Squares Means

Effect	device	_device	Estimate	Standard Error	DF	t Value	Pr > |t|
device	c	p	0.01906	0.006920	30	2.76	0.0099
device	c	s	0.03451	0.006920	30	4.99	<.0001
device	p	s	0.01544	0.006920	30	2.23	0.0333

...Continued

Adjustment	Adj P
Tukey-Kramer	0.0260
Tukey-Kramer	<.0001
Tukey-Kramer	0.0821

APPENDIX E. GRAPHS FOR CROSS SUBJECT COMPARISON OF EMG DATA

The follow graphs were chosen by subjective comparison to be most representative of subject 9 (representative subject in this study) EMG data. These graphs were then used in subjective comparison with all corresponding graphs of all subjects as data was processed. Units are millivolts (mV) and newtons (N).

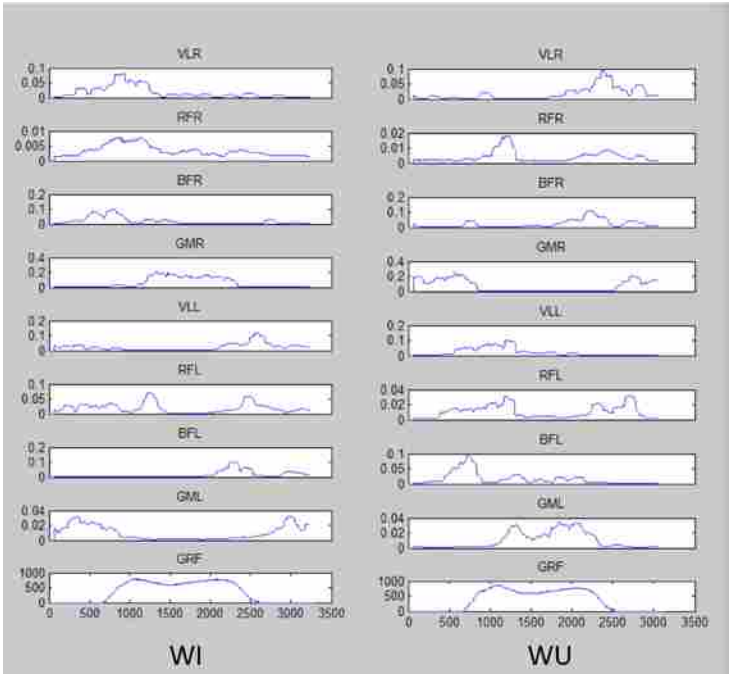

Figure E--1: EMG for an involved (WI) and uninvolved limb (WU) walking trial for the representative subject.

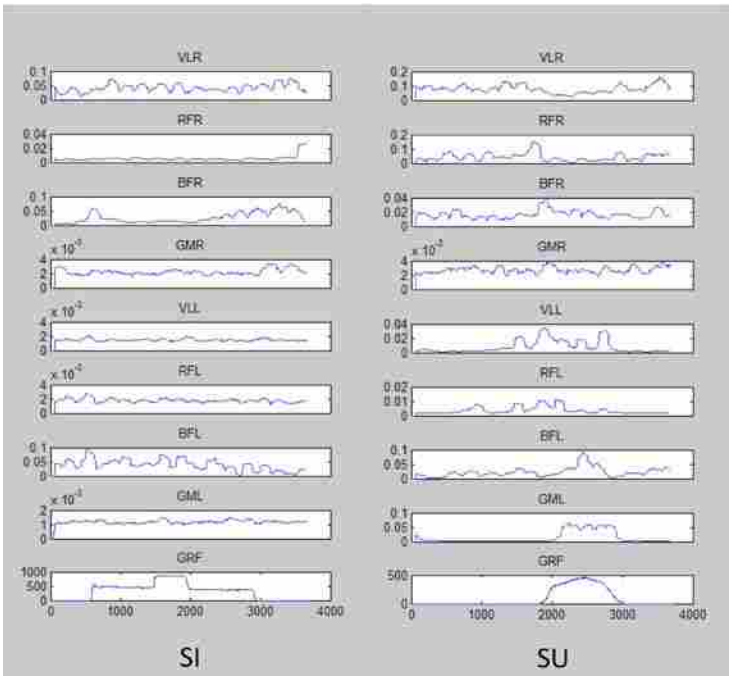

Figure E--2: EMG for an involved (SI) and uninvolved limb (SU) knee scooter trial for the representative subject.

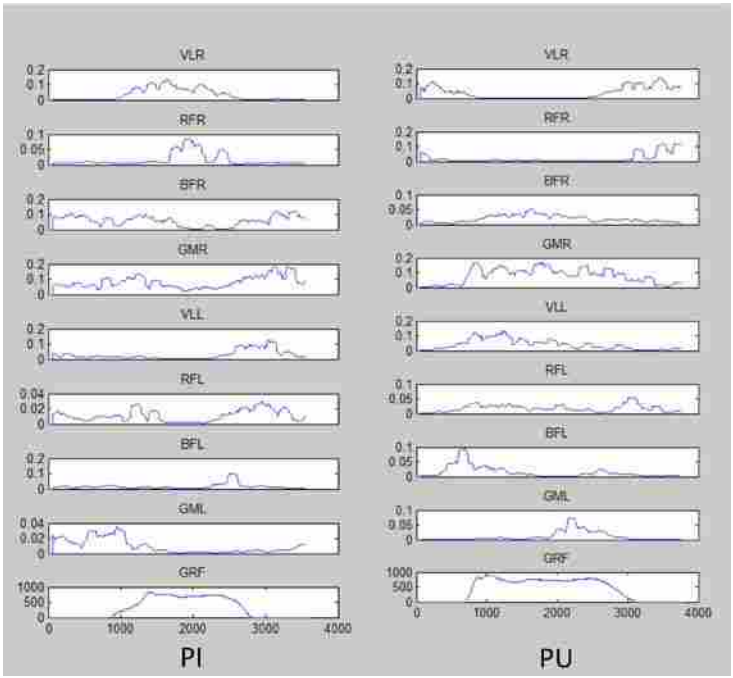

Figure E--3: EMG for an involved (PI) and uninvolved limb (PU) TI prosthesis trial for the representative subject.

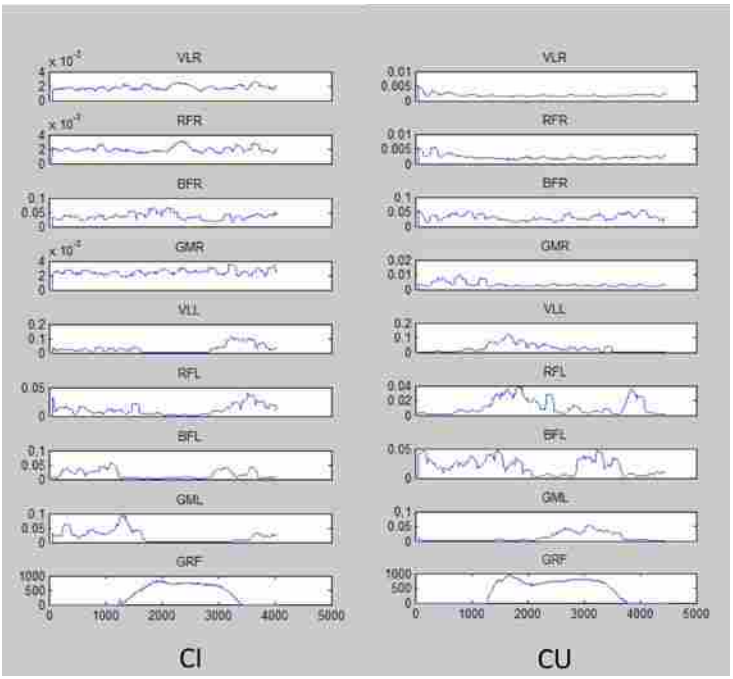

Figure E--4: EMG for an involved (CI) and uninvolved limb (CU) crutches trial for the representative subject.

APPENDIX F. CHECKLISTS FOR DATA GATHERING

The following checklists are the actually checklists referenced during data collection. In the case of the subject data gathering checklist, the simplified checklist was printed for each subject and each item was checked off as it was completed.

F.1 Subject data gathering checklist

General Notes:

1. **Let your subjects know what to expect**. Explain what you'll do before you do it. It may also help to explain why.
2. Have **everything ready to go** sitting out on the table for the subject.
3. **Wipe down equipment** between each use with alcohol wipes and towel.
4. **Watcher checklist:** correct foot or device hits the plate fully and exclusively. The subject is comfortable and maintaining speed all the way to the opposite cone.
5. Ensure that the **lab door is closed** so we don't have people walking in and out.
6. Two people place sensors so it goes faster.
7. If there is anything that would be uncomfortable and it is reasonable to have them do it have them do it. For example, pulling their shorts up to place a sensor and moving them back down.
8. Tell them that **if anything feels odd or uncomfortable let us know**.

Simplified checklist:

1. Informed **consent, photo & video release**, **screening** check. Can exit study anytime.
2. **Fitting, Training** and practice with the **devices.** They may have muscle soreness.
3. Change into **spandex** clothing
4. Prepped for **EMG** and place sensors
5. **Manual muscle test**
6. Place **motion markers**
7. Marker **static test** and **static photos** (standing and on devices)**.** Check **zero force plate.**

8. **NOTE:** treatment order is randomized, check the "subject demographics and device settings" spreadsheet for order.
9. 5 trials **right (involved)** leg then **check data**
10. 5 trials **left (uninvolved)** leg then **check data**
11. **Repeat** previous 2 steps for all devices/treatments
12. **IF DATA LOOKS GOOD**, **remove sensors** and markers
13. Change back to **regular cloths**
14. Subject is done and is **paid**. They **sign sheet.**
15. **Save data to external source (Google drive)**

Extended Checklist:

1. **Informed consent, photo & video release**, **screening check**
    a. At the beginning of the subject's appointment, they are screened based on the exclusion criteria. All questions and concerns about the informed consent document and the procedures are explained. If they still wish to participate in the study, informed consent is obtained along with photo and video release forms. A copy is offered to them as well.
    b. Collect demographics: height, weight, gender, age, dominant leg. Put in spreadsheet.
2. **Fitting, Training and practice with the devices.**
    a. Each of the devices will be demonstrated by a member of the research team so it is perfectly clear how to safely and effectively use each one and instructions will be given on proper use. A device is demonstrated and the fit to the subject. Subject is allowed to practice.
    b. Fit the devices. Crutches should be set such that there is approximately 1 hand width between armpit (axilla) and top of crutch pad. hand grip is set such that there is an approximate 20º angle at the elbow. Follow manufacturers recommendations for other devices.
    c. Ensure that each subject places their involved limb fully against the stop on the flexleg and scooter.
    d. Subjects will then be assisted by a member of the research team as they practice using the devices.
    e. Record device settings in spreadsheet.
    f. Use a 1-10 scale for device practice with normal walking being a 10 and not confident enough to use the device is a 1. Subject is asked to rate their level of confidence on this scale as they begin practicing and instructed to inform a member of the research team once they feel they can rate themselves at a 7 or higher. Once they rate themselves at 7 or higher and we agree they are proficient enough, data gathering proceeds. Subjects will not be permitted to continue to the next step of the study without being deemed proficient and comfortable. Extra time to practice is allowed as needed.
3. **Change into spandex clothing**

a. Once subjects are comfortable on each device, they will be instructed to change into spandex exercise clothing and provided a private location to do this. Show them where to change and where they can keep their stuff.
b. Women wear sports bra, spandex shorts and tennis shoes with ankle socks.
c. Men wear spandex shorts and tennis shoes with ankle socks.
d. BE SURE TO CHECK FOR REFLECTIVE SHOES. Cover the reflective portion with tape if there is any.

4. Prepped for EMG and place sensors

a. See "F.2 EMG setup checklist" in appendix F. FOLLOW EXACTLY THE INSTRUCTIONS FROM SENIAM (use a tape measure). Subjects are prepped for surface EMG using standard established practices (e.g. shaving of the sensor location area if necessary, slight abrasion of skin at that site and cleansing of the site with alcohol to reduce impedance of the skin).
b. Location for sensor placement is determined using standards such as those found at seniam.org. Muscles to be investigated are the vastus lateralis (VL), rectus femoris (RF) biceps femoris long head (BF), gastrocnemius medialis (GM). Sensors will be appropriately placed bilaterally on each of these muscles.
c. Sensor assignments: #1 through #4 are on the involved limb (R designates right leg). #1=VLR, #2=RFR , #3=BFR, #4=GMR.
d. Sensor assignments: #5 through #8 are on the uninvolved limb (L designates left leg). #5=VLL, #6=RFL, #7=BFL, #8=GML.
e. It will be best to have the subject strap on the FlexLeg or Kneel on it so you know where to place markers that won't interfere. For example the clusters can go on the front of the upper leg in between the FlexLeg cradles.
f. Don't pull the power flex tape too tight when covering the EMG on the first pass because it can dislodge the sensor. You can make it tighter on the second and third passes.
g. Turn on the sensors as you are placing them to ensure they are working, before you go covering them up with power flex tape.

5. Manual muscle test

a. do this before you wrap the sensor in power flex tape. You want to know it is working before you cover it up. Have the subject flex each muscle as instructed and ensure it is looking reasonable in the nexus software.

6. Place motion markers

a. See "F.3 Motion capture checklist" in appendix F.
b. When placing markers on devices, mark on the device where it was placed in case it falls off (done).

7. Marker static test and static photo. Check zero force plate.

a. Ensure the force plate has been zeroed.
b. You will have them stand on the force plate facing the wall (opposite of the door) and put their arms up until their right fist touches their right clavicle bone and

their left fist touches their left clavicle (the intend being that they don't occlude any of the markers).

c. Create new trial and name subject number, static (ex: p01static)
    i. Have them hold still for the length of the captured data (3-4 s)
    ii. click capture and wait 4 seconds
d. Take actual pictures while they are in this static stance of all four sides of them.
e. Static test is done for motion tracking software. Also, Static photographic images are taken for use as references in scaling model in OpenSim if needed.

8. Walking trials: 5 trials right (involved) leg then check data

a. subject is instructed to walk naturally, start at one mark (orange cone) and go to the other (orange cone) and to continue walking this loop until instructed to stop or unless they become uncomfortable doing so. They will step forward with whichever leg is most natural for them. They should be instructed to not slow down until they reach the other cone. They should be instructed to not target the plate.
b. Adjust the distance of the cone from the force plate until they are naturally hitting the force plate as part of the set up for that particular subject.
c. we are collecting data each time they enter the force plate zone. A watcher is watching to ensure the **right leg strikes the plate first** and correctly. Computer person is manually incrementing the trial number according to the naming convention. If there is a bad, additional trials are recorded. Bad trials are noted in the digitizing log. Naming convention:
    i. Method: Subject (p#), treatment (w = walking, c = crutches, s = scooter, p = temporary-injury prosthesis), involved (r) or not involved (l), trial (#)
    ii. Example: p1sr1. This would be subject 1 on a scooter measuring the involved limb on the force plate and the first trial.
d. Check data after first trial to ensure the following:
    i. Trackers are tracking
    ii. Force plate is working and not exhibiting and offset
    iii. All data looks good (EMG etc)

9. Walking trials: 5 trials left leg then check data

a. Same as above, but the **left leg should strike the plate**.

10. Device trials: 5 trials right (involved) leg then check data

a. For the FlexLeg this would be the device foot striking the plate. For the scooter it would be the scooter rolling across the plate. For crutches it would be the crutches themselves landing on the plate.

11. Device trials: 5 trials left leg then check data

a. This is the weight bearing leg in every case.

12. Repeat previous 2 steps for all devices

13. IF DATA LOOKS GOOD, remove sensors and markers

a. DO NOT remove sensors until we are certain the data is good. It will be a pain to get them back in on another day and another $40 in compensation.

14. Change back to regular cloths

a. Subjects can change back into regular clothes and gather up their stuff. They'll come back into the lab to get their money.

15. Subject is done and is paid

a. They sign a document that states they received the payment.

F.2 EMG setup checklist

*Indicates that these are the same steps you follow for doing motion tracking studies

System Setup*

- □ **Turn on** Vicon black box.
- □ **Turn on** the amps (blue boxes with orange switches)
- □ **Launch Nexus** software 2.1(updated version) (this is the software interface for EMG, motion tracking and force plate hardware)
- □ **Click go live** if it is not in live mode (if we stay offline longer the computer runs more smoothly)
- □ In nexus software on the left pane >system tab>drop down and **select "msanders2"**
- □ **Ensure Delsys (EMG software) is running** as well as Nexus (open program, but wait to turn on EMG to save battery)

Set up devices (done only once)*

in the system tab (left side of the screen)

- □ Drop down menu – select msanders2
  - o if you don't have EMG devices in there yet do the following:
    - ▪ Right Click "devices">add analog EMG
    - ▪ Right Click "voltage">add component
- □ Devices > #1 EMG > voltage (you can right click and add component to add more muscles). Name it and look at what pin it is.
  - o EMG starts at pin 25. Sensors are in order (e.g. #1 sensor is 1st in list, #9 sensor 2nd in list (#2 is broken), #3 is 3rd in list, …)
  - o you can replace a bad sensor with another sensor by doing the following:
    - ▪ in delsys ensure that all sensors are stopped
    - ▪ click on the pair button of the sensor number you want to use. for example, if 6 was bad and you want to replace it with 15 you would press "pair" on sensor 6 in delsys. Then a message will pop up saying it is waiting for some action.
    - ▪ You then hold down the button on the sensor for 3 seconds or until another message pops up asking about calibration.
    - ▪ click the box on the right (auto calibrate).
    - ▪ In delsys sensor 6 should now be active.
    - ▪ When you are done you can set the sensor back how it was.

Calibrate Force Plates*

Do this only if you plan to use the force plates in your study

- □ Zero out force plates
  - o Push autozero on blue box.
  - o Right-click #3 east > zero level
  - o Right-click #4 west > zero level

Sensor placement and setup

- □ **Locate site for EMG placement on body**
  - o Seniam.org is a great resource for this. Reference the google doc that Kenzie put together titled "EMG placement".
  - o Number EMG according to the muscles you want to put it on. The Dominant leg (right leg) is the involved limb for our study. So...
    - ▪ #1 through #4 are on the involved limb (R designates right leg). #1=VLR, #9(#2)=RFR (this muscle may be eliminated and we'd shift the numbers accordingly), #3=BFR, #16(#4)=GMR.
    - ▪ #5 through #8 are on the uninvolved limb (L designates left leg). #5=VLL, #6=RFL, #7=BFL, #8=GML.
    - ▪ Make sure we label the EMG signals in the left-hand pane or check them to ensure they are correct. I believe this is done just once per profile we create. We created a profile titled msanders2
- □ **Skin prep** (helps lower impedance)
  - o Shave hair (vacuum up after you are done)
  - o Roughen the skin with sand paper to clear off dead skin
  - o Clean skin with alcohol to get all oils and other contaminants off.
- □ **Orient arrow with muscle fiber orientation**.
  - o Need to know muscle anatomy and structure to do this properly. However for our study the fiber orientation is in parallel with the leg for all but the vastus lateralis. That has a slight angle. On that note, it is better to error on the side of being more parallel with the leg rather than more angled.
- □ **Turn on sensor**
  - o Little button on sensor. Green light comes on.
  - o In Delsys software window you should see that the EMG sensors are active.
  - o If they are not active, be sure to click run or start on the delays software.
- □ **Set up live graphs window**
  - o In Nexus click on window > open new floating window and pull onto second screen.
  - o In the new window click on dropdown box (it will likely be set as 3D perspective) and change it to graph.
    - ▪ You can right-click mouse and drag left or right and up and down to scale the graph.
    - ▪ Click both mouse buttons to pan.
    - ▪ Click on the different muscles in the first window in the system tab to select which muscles the graphs will display. You can select multiple muscles.

- □ **Manual muscle test:** Stick on sensors and check graphs while subject flexes different muscle groups per your instruction.
- □ If there is errant pulsing in the data, turn off the treadmill with the key and hit the big red kill switch.
- □ **Tape up:** If graphs look good, tape on with power flex tape.
- □ **Clean sensors.** When study is complete and you have checked the data you can remove the sensors. When done with sensors, pull off the stick backs and clean with alcohol swabs.

Setup Data Storage*

- □ **Set up subject, study and session**
  - Right pane>Communications button (bottom of screen)>Click the file data management button (looks like a folder), tells the computer where to store it.
  - New top level (study): green button
  - New subject (person): yellow button
  - New session (session for that person): gray button.
  - **Our naming convention** is as follows:
    - Method: Subject (p#), treatment (w = walking, c = crutches, s = scooter, p = temporary-injury prosthesis), involved (r) or not involved (l), trial (#).
    - Example: p01sr1. This would be subject 1 on a scooter measuring the involved limb on the force plate and the first trial.
  - **Editing subject names (as needed).** You have to click on the study to then edit subject name in the most right-hand pane or click on the subject to edit the gray button name. You can right-click on it in the most right-hand pane and then there is an option to edit.
- □ **Name Trial and set duration**
  - Click movie clip icon
  - Type trial name (see naming convention above).
  - For some studies it may be important to have the same number of frames for all trials and all subjects. Because our study uses devices that make the speed of travel across the plate much more variable. We stop and start manually to capture frames from before initial ground contact with the force plate until after final ground contact. If you want to set the duration you can accomplish this by check the box for "stop after duration". Select a duration that makes sense and keep it consistent across all subjects and trials.
    - For each trial of each subject, use the blue arrow handles in the viewing window of nexus software to clip the data down to your desired number of frames. We are currently using 200 frames.
- □ **Click "start"** under capture to capture data when you are ready to begin
- □ **Check data before moving on** to the next set of trials to make sure you got good data and what you needed.
  - Go "offline" and see steps outlined below in "check data and exporting data".
- □ **Check all data** before letting your subjects take of instrumentation and leave. You don't want to need to have them come back. see below for checking data.

Check Data and Exporting data*

- **Processing data and Checking it**
  - May need to click "go offline" to do the following
    - It will allow the computer to run better
  - Click communications tab (bottom of right pane)
  - browse to your folder, subject and trial
  - Select it and click "reconstruct" pipeline button in top left of screen (looks like luggage label). This processes the data.
  - once it's done click the save icon. A green circle with a P will appear next to it if it core processed correctly.
  - you can then click on the various devices in the left pane and view the data visually
    - for motion you are checking to see there is no flickering of the markers in the motion capture zone (you can see them all)
    - for emg you are checking for no clipping (signal goes up and then flatlines and then comes back down) or slipping (entire signal mean will shift down after a slip and the slip event is visible in the graph)
    - for force you are looking for absolutely no signal before heel strike on the z direction. basically that the plates are zeroed.
- **Exporting data**
  - click on tools
  - click pipeline icon (looks like a gear)
  - select your pipeline profile (msanders2) from current pipeline dropdown.
  - Ensure that the following options are checked
    - delete unlabeled trajectories (only pertains to motion tracking markers) **NOTE: ENSURE ALL MARKERS ARE LABELED BEFORE EXPORTING WITH THIS CHECKED OR IT WILL DELETE YOUR MARKER DATA. See notes on how to label trajectories in motion study checklist document for more information. The data is not gone forever and can be recovered.**
    - export ASCII
    - export C3D
  - click play button
  - go check that files are in the correct folder
    - c drive/vicon data/michael sanders atrophy study
- **Repeat these steps for each trial**. Good to check after the first one and make sure you get a good spreadsheet like you want.

F.3 Motion capture setup checklist

Motion Camera Setup

Don't spend too much time on this upfront (you will be making adjustments as needed after running through these steps. The test is if you can see all the markers without flickering when in the motion capture volume).

- □ Set up cameras: 6 on the wall, 4 on the floor (put 2 in front and 2 in back). If some cameras are not showing up in the system tab of the left hand pane you can *try restarting the software. If that doesn't work try turning on and off the black boxes*. This is because the old cameras sometimes don't play nice with the new cameras and system.
- □ Place wand in center of area (on the force plate is good).
- □ In nexus software be sure you selected "msanders2" as your profile name in nexus.
- □ Drop down menu in middle pane and select camera view.
- □ Select all the cameras and clear all masking with white X in middle pane.
- □ Window drop down menu>new floating workspace. You can then drag and drop this to the second screen.
- □ Adjust cameras to point at field
    - o Do this by clicking on them in system panel in nexus software. A blue light on the camera will light up.
    - o point the cameras at the wand. You'll go through each camera and ensure you can see all the markers.
    - o Use the ladder to adjust cameras on the wall
- □ Nexus software Left hand pane. Go through each camera (by clicking on it) and adjust the centroid fitting (two sliders) until you can see all the markers and they are not flickering (best to start with threshold and then do minimum circularity). The lower the threshold number the more noise (artifacts) on the screen. Remove or cover any object that you can that are showing up.

Masking Cameras

- □ Mask cameras: before you do this remove the wand and the markers you put down to define the space. You can select all the cameras and click "start" to mask the cameras and it will do it automatically, when it looks like it has masked everything, click stop. If that doesn't catch something then you can go through each camera and remove or mask false markers (could be other cameras, reflectors, metal, etc. done in the nexus software).
- □ Place wand back in space. Go through each camera and check threshold again. Sometimes this can be messed up after masking.
- □ Check that cameras can see all markers. You can select multiple cameras and view all at once.

Calibrating Cameras

- □ Calibrate cameras:
    - o Confirm that you have selected the profile "msanders2" previously in the left window pane. The cameras should be running at 100 hertz.
    - o Check the following settings in the right pane by clicking system preparations button (top of the right hand pane):
        - ▪ 5 marker wand and L-frame
        - ▪ full calibration
        - ▪ all cameras
        - ▪ 1500 (this defines how many images each camera must obtain of the wand before it stops capturing)
        - ▪ 1500

- Click start under calibrate cameras once you are ready to do the next step.
- Paint the entire study area you are interested in with the wand during the entire calibration time (it will stop automatically once every camera has 1500 captures).
- Check Image error. Image error for every camera be below .2 or you should to try it again. After speaking with Vicon, a threshold above .3 is probably okay. You can increase the number of captures to 2000 and see if that helps you can also switch cameras locations and see if that helps. Neither of those helped this time. You can call Vicon on speed dial if you get stuck.

□ Place the wand on the force plate in northeast corner of the so that the "L" lines up with two sides of the plate. Level the wand.
□ Click "start" under "Set volume origin" and click "set origin".
□ Check camera orientation and set up. In middle pane>click drop down>select 3D perspective. Move wand around and make sure it doesn't flicker while in the motion capture volume you defined. If it does, then adjustments to the cameras and recalibration may be necessary.
□ NOTE: if a camera gets bumped you will need to recalibrate the cameras.

Placing markers on Subject

□ Shave any areas that have hair before placing the marker on them
□ Overview of marker placement (11 individual plus 2 rigid body marker sets per leg)
- 4 markers on foot (dorsum, head of fifth metatarsal, third distal phalanx, and posterior calcaneus)
- 2 markers on ankle (lateral and medial malleolus)
- 2 markers on knee (lateral and medial femoral condyle)
- 2 markers on hip bone (anterior superior iliac spine and posterior superior iliac spine)
- 1 marker on hip joint (greater trochanter)
- 2 rigid body marker sets: one on the anterior thigh and one on the distal lateral shank

□ Placement of Rigid body Marker set
- An exact location for placement on leg is not critical so long as they stay put.
- Have the subject strap into the flexleg when determining the location of these so they don't interfere with the device.
- A four marker rigid body is placed is placed distally and laterally on the lower leg such that it does not interfere with the use of the flexleg or the scooter.
- A four marker rigid body is placed anteriorly on the upper leg and in a location such that it does not interfere with the use of the flexleg or scooter.

□ **Remove tape from markers**. When study is complete and you have checked the data you can remove the markers. Be careful not to touch the markers because it gets them dirty and then they don't work as well.

APPENDIX G. SUBJECT DEMOGRAPHICS, DEVICE ORDER, AND DEVICE WEIGHT

The following table contains subject demographics and the order in which devices were used to collect data for a particular subject (order was randomized; walking (w), knee scooter (s), TI prosthesis (p), crutch (c)). The various device weights in kilograms are as follows: knee scooter (9.81), TI prosthesis (2.65), medium crutches total for both (1.63), and large crutches total for both (1.94).

Table G--1: Subject Demographics and Device Order

	Demographics					Device order
subject ID	height	weight	gender	age	dominant leg	
3	5'8"	130	f	19	r	s,c,p,w
4	6'3"	182	m	27	r	w,s,p,c
5	6'2"	160	m	22	r	p,c,s,w
6	6'1"	160	m	21	r	w,p,s,c
7	5'8"	140	m	22	r	s,w,p,c
8	5'11"	150	m	19	r	w,p,s,c
9	6'2"	165	m	22	r	p,w,c,s
10	5'4"	140	f	18	r	c,p,w,s
11	5'6"	130	f	18	r	c,w,p,s
12	6'2"	180	m	22	r	c,w,p,s
13	5'4"	125	f	22	r	w,s,p,c
14	5'8"	120	f	21	r	c,p,s,w
15	5'1"	100	f	20	r	c,w,s,p
16	6'2"	137	m	25	r	s,w,p,c
17	5'4"	143	f	20	r	p,c,s,w
18	6'8"	198	m	23	r	c,p,w,s

APPENDIX H. IRB DOCUMENTATION

The following includes the actual documents approved by the IRB and used to recruit subjects and obtain informed consent and photographic release.

H.1 Informed consent

Consent to be a Research Subject

Introduction

Fig. 1 Crutches, knee scooter, and a temporary-injury prosthesis.

This research study is being conducted by Michael Sanders and Anton Bowden, Ph.D., at Brigham Young University to determine the effects of different mobility aids (see figure 1) on muscle adaptations of the leg. Every year there are approximately 3 million visits to the emergency room for below the knee injuries. These injuries are usually followed by a period of disuse of the injured leg, which requires the person to use a mobility aid for 6-8 weeks typically. Within this time a great deal of muscle loss and weakness usually occurs. Regaining this muscle and strength can take as long or longer than the period of disuse. This can be detrimental to a person's health and ability to return to normal and necessary activates as soon as they need or would like.
You were invited to participate because you are a healthy adult between the ages of 18-30 without any previous or current biomedical disorders that could affect the results or put you at higher risk for participating. Specifically you can answer **NO** to the following:

- ☐ Participation in an organized strength-training within the past 6 months.
- ☐ Current or previous musculoskeletal injuries affecting the legs, arms, or shoulder complex.
- ☐ Current or previous musculoskeletal disease.

- ☐ Chronic disease with regular clinical treatment.
- ☐ Regular drug, alcohol or tobacco intake.
- ☐ Any metabolic or hormonal disorder.

Procedures

If you agree to participate in this research study, you will be asked to do the following:

- Confirm your eligibility and desire to participate in the study (i.e. answer no to the questions above and sign this consent document). Provide some personal information such as name, age, weight, height and gender.

- You will then be instructed on proper use of each of the three devices. Each of the devices will be demonstrated by a member of the research team so it is perfectly clear how to safely and effectively use each one. You will then be assisted by a member of the research team as you practice using the devices. Your proficiency on each device will be determined by both yourself and a member of the research team. You will not be permitted to continue to the next step of the study without being deemed proficient and comfortable.

- Once you are comfortable on each device, you will be instructed to change into spandex exercise clothing and provided a private location to do this. Men will wear only spandex shorts and athletic shoes, women will wear only spandex shorts, sports bras and athletic shoes.

- You will then be prepped for surface electromyography (EMG) using standard established practices. This involves shaving of the sensor location area if necessary, slight abrasion of skin at that site and cleansing of the site with alcohol to reduce impedance of the skin. These sensors simply sit on top of the skin and can detect electrical signals from the muscles when they contract.

- The location for the placement of each sensor is determined using standards like those found at seniam.org. Sensors will be placed on the front of the thigh (vastus lateralis), the back of the thigh (biceps femoris) and on the back of the calf (gastrocnemius medialis). These muscles were chosen because of the role they play in normal ambulation. In order to correctly place these sensors, a member of the research team will use their hands on the upper and lower leg to feel for notable landmarks of the leg. For example around the ankle, knee and hip. These landmarks are used to then find where the sensors should be placed. Sensors will be placed in this way on both legs. Sensors are stuck to the skin using stickers that are made for the sensors. These sensors are small, wireless and light weight.

- Motion markers will be placed with a two sided tape that is commonly used for this purpose. Markers are placed on the lower half of the body in several locations that aid in the

identification of joints and segments of the leg. The markers are small, lightweight and are reflective in nature.

- A motionless test is performed as part of motion tracking set up. Photos are taken during this test for use in creating the models in the analysis software that we'll use (OpenSim).
- You will be instructed to walk at a comfortable pace across the force plate (this is a plate in the ground that is like a weight scale).
- You are then instructed to use standard crutches (medline) to go across the force plate at a comfortable pace.
- After using crutches, you will then go across the force plate using the temporary-injury prosthesis (TIP) (flexleg TIP from flexleg.com).
- Finally, you will be instructed to go across the force plate using the knee scooter (Roscoe Medical ROS-KSBG Knee Scooter from amazon.com).
- All sensors and markers will be removed and you will be provided a private location to change back into your regular clothes.
- This marks the completion of your involvement in the study. Nothing further is required.

All data collection will take place in the biomechanics lab (124 Richards building) on Brigham Young University campus in Provo, Utah. It is expected that the total time spent for the visit is 1.5 hours. The visit will take place at a time that is convenient for you.

Risks/Discomforts and Efforts to Reduce Those Risks

Risks/Discomforts

1. Stickers on the skin for gathering EMG and motion data. These stickers and tape are routinely used for EMG sensors and motion markers.

2. Very mild and temporary muscle soreness from using a mobility aid. This is equivalent to effects following a mild strength training workout. This potential muscle soreness is due to a change in the muscles you used as a result of using a pair of crutches, knee scooter or temporary-injury prosthesis (TIP).

3. Minimal fall risk from using a mobility aid. This risk is no greater than might be expected from using a medically prescribed mobility aid for recovery from a lower-leg non-weight bearing injury. These sorts of device are and have been used by many people over the past several years.

4. Some individuals may feel uncomfortable wearing spandex exercise clothing.

5. Discomfort due to skin preparation for EMG sensors. In order to get a good signal, the skin in the area where the sensors are placed is shaved if necessary, lightly abraded and cleansed with alcohol. This is all done in accordance with routinely accepted practice for surface EMG use.

Efforts to Reduce Those Risks

1. You will be informed of the use of these adhesives and that you can exit the study at any time.
2. You are instructed on the proper use of each mobility aid and are informed of the potential for mild muscle soreness. Being that you are a healthy 18-30 year old individual, recovery from this type of soreness, if any occurs, will be rapid.
3. Training will be provided on how to safely use each device as might be expected for a patient who would be using it for an injury. One or more members of the research team will be there during the entire time of your use of the device. Additionally, you are instructed that you can exit the study at anytime if you feel uncomfortable using the device.
4. You will be made aware of the need to use this clothing during the study. All data will be gathered within the lab thereby minimizing contact with others while wearing the spandex clothing. Further, you will be informed that you may leave the study at anytime if you feel uncomfortable in the clothing.
5. You will be informed of this procedure and all skin preparation will be performed by a trained member of the research team.

If at any time, or for any reason, you wish to discontinue participation, you may inform the researchers of this and withdraw yourself from this study. You will have no other negative consequences except for the loss of the stated study compensation.

Benefits

There will be no direct benefits to you. However, it is hoped that the understanding gained from this research will aid clinicians in their prescription of these mobility aids as well as guide others towards better designs that are more conscientious of atrophy and its prevention.

Confidentiality

Subjects will be assigned a unique study ID and subject data will be associated with that study ID. The key between the subject's name and the data will be kept in a locked filing cabinet, and only those study personnel immediately involved in the research study will have access to the key. The data, which does not contain any personal identifiers, will be stored on lab-associated computers for analysis. The data will be stored for a minimum of three years. Only group data will be published, and it will be published without any personal identifiers. When the data analysis has been completed and three years have passed (whichever happens last), the data will be deleted. All photographs and/or video recordings obtained during the study will not identify you by name.

Compensation

Upon completion of your participation in this study you will receive $40 in cash. This compensation will not be prorated.

Participation

Participation in this research study is voluntary. You have the right to withdraw at any time or refuse to participate entirely without jeopardy to your standing (if applicable) with the university.

Questions about the Research

If you have questions regarding this study, you may contact Michael Sanders at (801) 921-9377 or michaelsanders.me@gmail.com for further information. You may also contact the faculty advisor for this research, Dr. Anton Bowden at (801) 422-4760 or abowden@byu.edu.

Questions about Your Rights as Research Participants

If you have questions regarding your rights as a research participant, contact IRB Administrator at (801) 422-1461; A-285 ASB, Brigham Young University, Provo, UT 84602; irb@byu.edu.

Statement of Consent
I have read, understood, and received a copy of the above consent and desire of my own free will to participate in this study.

Name (Printed): _____

Signature:
_____Date:_____

H.2 Classroom announcement

Classroom Announcement:

The classroom script may be accompanied by a powerpoint slide of the IRB approved flyer that we previously submitted. The script will read as follows:

Research is being conducted to understand how various mobility aids effect skeletal muscle adaptations of the lower limb during a period of non-weight bearing on the limb.

Subjects will be asked to perform specific movements while using the various mobility aids. Motion and EMG data will be collected as the patient performs these movements. A single visit will take place in the biomechanics lab (124 Richards Building) on BYU campus and should last approximately 1.5 hours. Upon completion of the study, participants will be compensated $40 in cash for their time and effort.

To Participate: Must be between the ages of 18-30 years, free from past or current musculoskeletal disease or injury and be physically fit.

For more information, contact Michael Sanders.

michaelsanders.me@gmail.com
801-921-9377

This research is conducted under the direction of Professor Anton Bowden, Mechanical Engineering Department.

H.3 Recruitment flyer

VOLUNTEERS NEEDED

FOR RESEARCH STUDY!

We need participants for a research study:

Mobility Aids and Muscle Adaptations

Description of the project: Research is being conducted to understand the effects of various mobility aids (see figure 1) on skeletal muscle adaptations of the lower limb. Data will be gathered through EMG and motion tracking on participants.

Fig. 1 Crutches, knee scooter, and a temporary-injury prosthesis.

Motion and EMG data will be collected as the participant performs specific movements using the various mobility aids. Participants will come for a single visit to a lab on campus (124 Richards Building) that will last approx. 1.5 hours. If you complete the study, you will be compensated $40 in cash for your time and efforts.

To participate: Must be between the ages of 18-30 years, free from past or current musculoskeletal disease or injury and are physically fit.

For more information, contact:

Michael Sanders
michaelsanders.me@gmail.com
mobile: (801) 921-9377

This research is conducted under the direction of Professor Anton Bowden, Mechanical Engineering Department.

H.4 Photographic release

Photographic Release Form

As part of this project, I will be taking photographs of you during your participation in the research. These photographs will mainly be used to gather data concerning your position and motion, as they are reviewed by members of the research team. We may desire to use select photographs for other purposes and request your permission for the specific uses. Please indicate what uses of these photographs you are willing to permit, by initialing next to the uses you agree to and signing at the end. This choice is completely up to you. I will only use the photographs in the ways that you agree to. In any use of the photographs, you will not be identified by name. In any publishing or public display of any photograph, your participation will be kept anonymous by concealing (through Photoshop® or image cropping) your face and any other identifying marks (e.g., scars, tattoos) that you specifically state below.

_____ Photographs can be reviewed by the research team.

_____ Photographs can be used for project illustrations.

_____ Photographs can be used for classroom presentations.

_____ Photographs can be used for academic conference presentations.

_____ Photographs can be used for fundraising presentations/proposals.

_____ Photographs can be used for newspaper or magazine publication.

_____ Photographs can be used for journal publication.

_____ Photographs can be posted to a website.

Please specify any identifying marks (e.g., scars, tattoos) that you would like to be concealed during any public display of the photographs: _____

_____

_____

_____

I have read the above descriptions and give my express written consent for the use of the photographs as indicated by my initials above.

Name (Printed):_____

Signature:_____ Date: _____

APPENDIX I. OPENSIM AND MUSCLE FORCE ANALYSIS

While the following can be re-derived independently, it seemed prudent to include a summary of resources and information to make processing of data in the future easier.

I.1 OpenSim resources for model and simulation setup

Several videos were produced illustrating how we used opensim to process motion capture data to produce muscle force predictions and can be obtained for reference (folder "OpenSim instructional videos"). There is also a companion reference document "Q & A on OpenSim from Spencer's Tutorial Videos" that can be found in the same folder. In particular, the videos show that checkboxes for "preserve mass distribution during scaling" and "marker data from measurements" were selected. Also important to note is that when scaling a model, each segment is scaled to maintain its relative mass to the overall model. Therefore, the model needs to be scaled for a particular subject before adding the weight of a device to a particular segment. If this is not done, then the weight that was added for the device will be scaled along with the rest of the model. This also means that for every subject, there will be a model for each device that required mass to be added to the model.

I.2 Determining range of muscle force data

The follow is taken from the spreadsheet "graphs All EMG and all muscle force p09wr05 (plus proto and notes)" and is instructive in plotting the correct segment of muscle force with the corresponding EMG.

- Open the file containing EMG data for a particular trial. Find IGC with its associated GRF and then find FGC with its associated GRF. Note those.
- Then open the grf.mot file that is created by the lee-son toolbox (eg "p09wr05_fixed_grf.mot"). Column i contains the GRF from the imported data (knee scooter trials are column c). You can then search that column for the IGC GRF you found from the EMG data. Then note the time that that occurs (column A).
- Then search column i for the FGC GRF you found from the EMG data. Note the time that that occurs (column A).
- You can now go open the spreadsheet that was exported for muscle data (eg "p09wr05 (all muscles, data exported)") and find the corresponding time for IGC and FGC (it will be within a 100^{th} or 1000^{th} of a second, not exactly the same typically). This is the range of muscle force you want to graph in the EMG-muscle force overlay graph.