BALLISTIC MOVEMENTS:

A UNIQUE APPROACH TO EMG NORMALIZATIONAND ITS EFFECT ON JOINT MOMENT ESTIMATION

by

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ABSTRACT

Modeling human movement using a computational based algorithm to predict muscle forces, ligament strains, and joint loads requires several inputs including normalized electromyographic (EMG) signals. Normalizing EMG reduces signal variability caused by electrode placement. EMG inputs scale the model's output muscle forces and employing normalized values greater than 100% implies muscles are generating supramaximal forces. Signals from maximum voluntary isometric contractions (MVIC) are typically used for normalization, even though ballistic tasks can produce larger EMG singals. Ballistic task EMG signals normalized to MVIC EMG maximums could yield EMG values greater than 100%. Normalizing to maximal values ensures EMG signals remain below 100%.

This thesis investigated dynamic signal repeatability and differences between MVIC EMG maximums and maximums from sprinting, jumping, and isokinetic movements to ascertain whether dynamic tasks produce significantly greater signals in search of a task which elicits reliable, maximal EMG. These maximum dynamic signals and the maximum MVIC values were used to normalize the model inputs. The model outputs of the two cases were compared.

The results showed a significant increase of peak EMG values between the dynamic and MVIC cases for the lateral gastrocnemius (LG), medial gastrocnemius (MG), soleus (SL), vastus medialis (VM), and vastus lateralis (VL). The peak EMG was significantly greater between the ballistic and MVIC cases for the LG, MG, and SL. Divergences of signal magnitudes were repeated across testing sessions. Intra-session analysis revealed reliability in almost all peak EMG values. The two EMG-driven model outputs established the MVIC normalization procedure as the superior

method. The r^2 value of the knee moment for the MVIC case was 0.94 while the sprint case was 0.91 and the MVIC case produced a significant decrease in peak knee moment error, with respect to the moment calculated using inverse dynamics.

This thesis identified MVIC EMG values as the better choice for normalizing dynamic EMG signals for the EMG-driven model and the results of this work establish a greater confidence in the model's applicability to highly dynamic activities. This thesis also determined dynamic tasks exist which produce significantly larger, repeatable peak EMG signals than the MVIC tasks.

Chapter 1

INTRODUCTION

Analysis of muscle activation produced during activities ranging from the daily life motions of walking and stair climbing to the athletic motions of running and jumping requires the collection of reliable EMG signals measured under dynamic conditions. EMG signals are recordings of the electrical activity that are associated with muscle contractions; these signals give insight into muscle function based on contraction timing and intensity. EMG collects signals from the muscle motor units in response to a neural command. Though EMG is representative of muscle activation, muscle force production and muscle activation do not equate to each other in a linear manner [Buchanan 2004]. Muscles' EMG signals are independent of one another and no inherent relation exists between them. EMG is measured as a voltage and is highly dependent on many factors other than contraction intensity, including electrode placement, skin impedance, joint angle [Farina 2001], and joint velocity [Bigland 1954]. Due to the factors affecting the signal, it is important to normalize the EMG values for comparison of muscle activations across muscles, testing sessions, subjects, and as inputs into an EMG-driven muscle model.

The EMG-driven muscle model is a computational based model which requires a series of inputs, including EMG, to predict individual muscle forces. The EMG inputs are used to determine the muscle activation intensity. The muscle activation intensity scales each muscle's output force based on its maximum producible force [Buchanan 2004]. The model has been tested repeatedly for low intensity movements, normalized to the consistent EMG values produced during maximum voluntary isometric contractions (MVIC). After normalization, the EMG values were less than 100% activation. Higher velocity, or ballistic, tasks often produce EMG magnitudes greater than those generated during MVIC tasks [Gazendam 2007]. Each muscle's maximum isometric force and cross sectional area is not uniquely input into the EMG-driven model for each subject. This requires the model to adjust the maximum isometric force of each muscle starting from the average values reported in literature [Winter 1990]. Therefore, normalizing to EMG values produced by isometric tasks could produce muscle activations much greater than 100% and lead to the incorrect conclusion that a muscle is producing forces outside of reasonable, physiological bounds. This mistake is avoided by normalizing to the maximal EMG value.

Finding an activity to yield a maximal EMG signal requires attention to many factors including joint angle, velocity, contraction magnitude, and fiber alignment. Muscle contraction intensity strongly influences EMG [Guo 2010]. The EMG signal is a result of the quantity of active motor units and the rates at which they are firing. As a contraction increases, the firing rate of the active motor unit increases. Once reaching a frequency threshold, an additional motor unit is recruited and this continues until the system reaches its maximum output [Person 1972], therefore a maximum contraction is needed to produce the maximal EMG. A MVIC is an example of a task requiring maximum contraction intensity while holding other variables constant leading to a muscle producing its maximum force [Hill 1938]. Holding other variables constant and exerting a maximum contraction is what causes MVIC to have an excellent reliability, which directly contributes to its use as a common normalization value. Previous studies have shown MVIC to be reliable during a variety of isometric tasks including knee and ankle position being fixed by a researcher, by a dynamometer, or by a stationary cycle [Lin 2008, Norcross 2010, Rouffet 2008]. The MVIC EMG value, because of the other influential factors, is unfortunately not maximal.

MVIC tasks produce reliable EMG data and yield the greater forces than concentric contractions of a muscle but this is not always the normalization value of choice, depending on the desired comparison. Ballistic EMG normalized to MVIC values could lead to EMG values above 100% activation [Burden 1999]. Since maximum force does not guarantee maximum EMG production [Buchanan 2003] and dynamic tasks exist that produce larger EMG signals, dynamic EMG has an important role in normalization. There is a wide variety of movements from which to record dynamic EMG signals ranging from controlled isokinetic contractions (ISOK) to the much more variable ballistic movements. Clinicians use ISOK movements for assessment [Maupas 2002] and muscle hypertrophy [Melegati 1997] and the EMG data produced by these movements are reliable [Larsson 2003]. Lower velocity dynamic movements, having unconstrained motion, have proved reliable as well. A single, shallow squat, to an approximately 45° knee flexion angle, produced reliable EMG from the quadriceps [Earl 2001]. EMG values collected from the gastrocnemius produced during self-selected walking gait were also found to be reliable [Karamanidis 2004]. Higher velocity dynamic trials provided reliable EMG signals in only limited instances.

Reliability varies depending on the dynamic movement and the muscle being tested. Ballistic tasks in previous studies of the leg muscles yielded repeatable

EMG in several cases including: the lower leg muscles during running at a fixed speed on a treadmill [Chapman 2009, Gollhofer 1990, Karamanidis 2004] and during rhythmic hopping at a single frequency [Gollhofer 1990], the quadriceps muscles during maximum vertical jumping [Goodwin 1999], and the quadriceps and hamstrings during sprinting into a 45° cut maneuver and drop jumps from a fixed platform height [Fauth 2010]. Sprinting and maximal vertical jumping have repeatable kinematics [Yoon 2006, Queen 2006] and requires peak exertions to perform which meets the criteria for finding a task which produces a reliable, maximal EMG signal.

Ballistic tasks differing from jumping and hopping to running and cutting have been studied briefly, but previous research has not compared the magnitudes of the resulting EMG nor have studies been performed on the primary muscles acting at the knee and ankle joints during a single collection of data. The EMG-driven muscle model needs to be processed with differing normalization techniques to look at the models sensitivity to normalization techniques. This work impacts those capable of performing maximal tasks, which can be applied as model inputs. The applications of computer based muscle models for use during rapid dynamic tasks are numerous and it is important to evaluate the model's output variation when altering normalizing methods of maximal and submaximal tasks.

1.1 Focus of the Thesis

The purpose of this study was to identify a task which generates reliable and maximal EMG signals for 9 lower leg muscles; the biceps femoris (BF), lateral gastrocnemius (LG), medial gastrocnemius (MG), rectus femoris (RF), soleus (SL), semimembranosus (SM), tibialis anterior (TA), vastus lateralis (VL) and vastus medialis (VM). These maximal signals, found during the ballistic task, were applied to the EMG inputs in order to evaluate the model's ability to predict joint moments of rapid dynamic movements.

1.1.1 Aim 1: Determine which task will produce the largest EMG signal

EMG signals were collected from 9 leg muscles during isometric, isokinetic, sprinting, and jump tasks. Maximum EMG values were found from a linear envelope created by filtering the signal. Planned comparisons were performed on the EMG signals to establish significant differences between each of the tasks. Our hypothesis was that the ballistic tasks would produce larger EMG signals than the MVIC and isokinetic tasks.

1.1.2 Aim 2: Determine whether the static and dynamic EMG maximums are reliable

Each of the subjects performed several trials for each of the tasks listed in Aim 1. The maximums of each trial were recorded and used to establish an intraclass correlation coefficient (ICC) to show reliability of the tasks in a single testing session. On a separate day, subjects returned and performed the protocol again. The maximum EMG signals from the first collection session were compared to the maximum signals of the second collection session. As an inter-day comparison of maximal tasks, the significant differences found of the first day were compared to the significant differences found on the second day. Our hypotheses are that the static tasks will produce repeatable EMG maximums and the sprint EMG maximums will be as reliable as the isokinetic EMG maximums. Additionally, the maximum EMG signals for each task will not be significantly different between test and retest sessions.

1.1.3 Aim 3: Determine whether EMG inputs greater than 100% activation affects joint moment outputs.

The EMG inputs for modeling the stance phase of a run trial were normalized to MVIC maximums and sprint maximums separately to produce two sets of inputs. These inputs were processed through the model to produce two separate knee moment outputs. The difference between these moment outputs was determined. Our hypothesis is the smaller input activations, determined by normalizing to larger EMG maximums, will produce small joint moment outputs from the EMG-driven model.

1.2 Significance of the Research

EMG signals are an excellent tool for the understanding of muscle contractions and relative activation magnitudes. Task assessment can be performed with EMG but an improper normalization technique used on the data can alter the conclusions [Benoit 2003]. Normalizing to a muscle's maximal EMG signal allows for calculating accurate co-activations and determining optimal exercises for peak muscle excitation [Ebben 2009]. To our knowledge, there have only been limited studies performed on few tasks and muscles examining the magnitudes of peak EMG signals during several tasks, static or ballistic. Our first aim was to understand which task a subject could perform in order to elicit a maximal EMG signal magnitude.

The maximal signal also impacts EMG-driven, musculoskeletal modeling. Musculoskeletal modeling allows researchers to understand the effects of pathologies [Winby 2009] and provide a more precise understanding of muscle kinetics which could indicate high risk kinematic movements [Lloyd 2001]. The EMG-driven model has been thoroughly tested for low velocity movements, such as walking, normalized to MVIC and has also been tested for EMG inputs normalized to the maximum EMG values of the task being modeled [Besier 2009]. The model has the potential to be used for much more dynamic tasks involving rapid contractions. This study creates a better understanding of the effects of normalizing to different values or processing the model with EMG values greater than 100%.

1.3 Thesis Outline

The next chapter will explain the source, background, and importance of EMG. Chapter 3 presents the data from the static and dynamic collections for comparison to establish a maximal and reliable signal. Chapter 4 establishes differences found in the outputs of the model after altering the inputs by varying EMG normalization techniques. Chapter 5 addresses each of the hypotheses of this study and remarks on possible future works for these maximal EMG data.

Chapter 2

BACKGROUND

2.1 EMG

EMG is the abbreviation for "electromyogram" or "electromyography," which is the signal, or the collection of signals, detected from the change in membrane potential of the muscle cells [Gasser 1930]. The detectable activation elevates with increased muscle fiber recruitment required to perform a task. The more strenuous the task, the more muscle fibers are required. The increased demand spurs the recruitment of additional activation of motor units which incorporates a greater number of fibers. As the task increases in difficulty, more motor units are recruited until all of the motor units are stimulated and the system is at its maximum output [Person 1972]. Muscle activations, determined with EMG signals, establish recruitment timing within a task [Winter 1983], the understanding of neurological and pathological deficits [Hallett 1975, Crenna 1998, Hubley-Kozey 2006], and contraction intensity to perform a particular task [Byrne 2007]. EMG allows researchers to identify which muscle is the primary contractor for a given task and identifies synergistic muscle activity in the co-contractors. The degree of co-contraction leads to the understanding of joint control and stability [Buchanan 1995, Zeni 2010]. The possible applications of EMG data are numerous, but the diversity of EMG utility is comparable to the variety of collection techniques and processing methods available which warrants the need for caution with the application of this tool.

2.1.1 Instrumentation

The EMG signal is collected through two different methods of signal detection. The first method is fine wire. Two fine wires are inserted into the muscle with a needle and detect activation within the muscle. The other method is much less invasive and does not affect subjects' movement patterns as drastically. This method is surface EMG. The electrical potential generated within the muscle is detected from the surface of the skin using two electrodes. The intensity of the signal from the surface electrodes is on the order of magnitude of millivolts [Nigg 2007] causing the need for the signal to be amplified. The amplification is at the researcher's discretion based on the equipment available. The placement of two separate electrodes can cause a discrepancy of inter-electrode distance so it is now common to use a single unit with with dual electrodes, fixing the collection area and preventing variations between muscles, subjects, and tests. The distance between the electrodes determines the muscle area from which the signal is being collected and based on that distance the signal can change significantly. Twenty (20) mm is the recommended spacing between the electrode poles [Farina 2002].

2.1.2 Anatomical variance

Electromyograms are sensitive to a variety of factors including the subject's physical condition, length of the muscle at the time contraction, and the location of the electrode in reference to the muscle's motor units. Surface EMG electrodes are non-invasive, but have the limitation of receiving a signal through the skin. Each layer of tissue presents an impedance which can vary within a subject over time [Nicander 2000]. Included in this layering of tissue is adipose, which greatly changes between subjects and recording location. Adipose tissue also increases the chance of the electrode moving in relation to the muscle during a collection, specifically during a dynamic activity, thus altering the signal [Farina 2006]. Muscle length change has been theorized to alter the signal in the same manner since a muscle contraction moves the muscle under the skin and not necessarily uniformly in relation to the electrode [Rainoldi 2000]. Others research has theorized the variability in EMG magnitude is a due to a neural inhibition caused when the joint is at a disadvantaged angle [Azegami 2007, Maffoietti 2003, Suter 1997]. Ideally, the line created by the two poles of the electrode should be aligned with direction of the muscle fibers being tested, but a muscle's fiber orientation changes with contraction [Maganaris 1998] which creates another anatomical shift in reference to the electrode. The rotational changes in fiber alignment in relation to the electrode during the contraction affect the output signals [Farina 2006].

2.1.3 Physiologic variance

Effectors of signal changes in the muscle during a contraction range from muscle contraction velocity to contraction intensity, which are independent of one another. Contraction intensity is paramount in the collection of EMG signals. Based on effort, the subject can control the magnitude of EMG elicited within a certain degree of accuracy [Guo 2010]. The rate of contraction at designated intensities also manipulates the signal. Ramping up effort during a static contraction creates a different signal than contracting in a very rapid, or ballistic, effort [Ricard 2005]. Motor units are rapidly recruited in response to a ballistic demand versus constant contraction intensities. The muscular response to the rapid motor unit recruitment produces a completely unique EMG signal to that of a steady, ramp up contraction [Semmler 2002]. In the same context, joint angular velocity has been correlated to

EMG magnitude changes. Specific muscles have shown a relationship between increased contraction velocity and elevated EMG signals [Bigland 1954, Soderburg 1984].

2.1.4 Fatigue

When collecting data on any mechanical or biomechanical system, fatigue must be taken into account and EMG data are no different. The effect of muscle fatigue depends greatly on the use of the muscle. If a muscle is sub-maximally activated, the signal acts differently over time than a muscle maximally activated. Once a muscle has been fatigued recovery time becomes a factor [Hultman 1983] and sufficient time must be allotted for rest to prevent fatigue effects on the signal.

Static contractions have been looked at for EMG fatigue because of their ability to be held for as long as the muscle can maintain a contraction. Dynamic cases, limited by range of motion, do not provide long contraction durations, nor show significant fatiguing effects [Babault 2006]. EMG signals during maximum effort static tasks behave differently compared to that of sub-maximal activities. Maximal static contractions demonstrate a large EMG signal at the commencement of contraction, but then quickly fall asymptotically with fatigue [Babault 2006, Hultman 1983, Stephens 1972]. Contracting sub-maximally leads the EMG signal to actually increase with time [Kuroda 1970, Adam 2005, Mottram 2005, Kirsch 1992, Thomas 2006, Lind 1979]. Initially, during a sub-maximal contraction, the EMG signal increases. The duration of this increase depends on the relative intensity compared to the maximal activation. If the sub-maximal contraction is held for enough time, the EMG signal actually begins to fall, similarly to the maximal contraction case. This phenomenon occurs because the initially active motor units, which are the higher frequency, lower amplitude units fatigue and the system responds by activating the lower frequency, higher amplitude motor units in order to maintain the same level of contraction [Lloyd 1971]. The EMG of the sub-maximal activation is not able to exceed the EMG signal of the MVIC case regardless of the duration of contraction. When the muscle is fatigued enough through the course of a sub-maximal contraction, the EMG curve begins to behave like the MVIC case with a similar level of exhaustion [Lind 1979]. The EMG increases produced through sub-maximal contraction also apply to compounding fatigue. Multiple sub-maximal contractions with limited recovery time cause the EMG signal to increase as though it were contracted statically for an extended period of time [Mendez-Villanueva 2009]. Appropriate rest for subjects during EMG collections is paramount to obtaining accurate results. Hultman et al. found EMG returning to its maximum levels during MVIC contractions after only 30 seconds of rest [Hultman 1983] which provides the minimum amount of rest recommended for EMG studies.

2.2 Processing

The processing method of the collected EMG data is equally important as the collection techniques. Since the data are returned in volts and with jagged, positive and negative values, several steps must be taken to obtain a functional signal for analyses. EMG signals yielded during dynamic movements, as previously stated, possess the potential error caused by skin motion artifact. This artifact occurs at 1-10 Hz and therefore it is recommended to pass the signal through a high pass filter with a cutoff of greater than 10Hz [De Luca 2010]. Since this study used rapid movements which potentially adds higher frequency artifact, 30Hz was the high pass filter limit.

The voltage signal from the EMG system is a positive and negative signal reflecting over a floating voltage caused by a lack of ideal grounding. It is common to place a grounding electrode on the subject to establish a reference voltage, but that does not guarantee the signal is based at 0.0 volts when it is compared to the actual ground on the analog to digital board. This requires a DC offset component to be removed to prevent a shift in signal magnitude which would skew EMG evaluations. This is done by finding the mean of the baseline noise and subtracting the offset from the entire signal, setting the voltage reference to 0.0 volts. After properly referencing the signal, it can be rectified. Since the signal emanating from the muscle is not directionally specific, the signal returns positive and negative voltages. The signals intensity is of interest and not the direction of the differential, therefore the negative component can be made positive.

Smoothing out the jagged EMG, which is riddled with local minima and maxima, allows for researchers to gain an understanding of the contraction patterns during common movement patterns. Low pass filtering of the EMG signals creates a linear envelope with a continuous shape for making activation timings more easily identifiable. The envelope is especially useful with a repeating pattern, such as gait, and can be averaged over several cycles. The filter cut off of choice is based on the primary component of the frequency spectrum. It is recommended that 95% of the signals power is contained in the filtered signal in order to avoid cutoff error [Shiavi 1998]. As the intensity of the activity increases, the power shifts towards the lower frequencies so more dynamic activity should use a low pass filter with a lower cutoff.

2.3 Summary

EMG is a multipurpose tool which is used in determining muscle timings, intensities, and co-activations in healthy and injured populations. This data can easily be misused and misinterpreted with the wrong collection and processing techniques. Proper collection locations should be established through an understanding of the target muscles' physical size, fiber orientation, line of action, and proximity to other muscles. Researchers must consider the muscle's function to gain a full understanding of the output and take into account contraction duration, compounding exhaustion, and resting time between trials when collecting EMG data. Following the collection, determining desired comparisons and knowing the frequency of the task leads to the application of suitable filters to the signals. The application of each of these details aids in the generation of a quality signal that can be used for comparing subjects, muscles, tasks, and many other important research details.

Chapter 3

COMPARING EMG MAGNITUDES OF STATIC AND DYNAMIC MOVEMENTS

3.1 Introduction

Electromyographic (EMG) signal recording is an essential tool for muscle activation assessment. EMG allows researchers and clinicians to establish muscle recruitment patterns, timing, and intensities during human movement [Ounpuu 1997, Prosser 2011]. A series of surface electrodes collecting EMG from various muscles identifies the contraction intensity needed to perform an action, but EMG signals are highly variable due to several differing anatomical traits including skin thickness and tissue conductivity [Farina 2001]. Electrode placement alone can result in a significant shift in the EMG values due to motor unit depth and fiber orientation relative to the surface EMG electrode [Campanini 2007]. Comparing across muscles, tasks, subjects, and testing sessions requires EMG signals to be normalized to a reference value [Knutson 1994, Mirka 1991, Burden 1999]. Normalizing signals to maximum voluntary isometric contraction (MVIC) values helps reduce the extrinsic variability in the EMG signal, but analyses between muscles and tasks within a subject may benefit from a more robust normalization method.

Whether the intent is to discover an injury mechanism, musculoskeletal response to pathology, or analyze elite performance, referencing to the maximal signal establishes a common relation between each of the active muscles during a task. Co-

contraction is the activation of an antagonist muscle group, opposite to the muscles acting in the contraction direction, to increasing joint stability [Lin 2007, Caty 2007] or in response to pathology [Hortobagyi 2005, Zeni 2010]. Assessing the effort required by the antagonists to create joint stability effectively is easily evaluated if each muscle's EMG signal is below 100% of its possible activation. The only way to ensure this fact is to normalize each muscle to its maximal signal. Besides establishing co-contraction, the result from normalizing to the maximal value determines the contribution of each muscle during a movement [Chapman 2008]. This information proves useful for comparing subjects of different performance abilities or tracking a subject's improvement over time. Normalized EMG is also used for the comparison of muscles between tasks and limbs during athletically demanding activities. Comparing activation magnitudes between these tasks requires a maximal EMG signal to differentiate between the more muscularly demanding exercises [Behm 2005] as well as the activation strategy each subject utilizes to perform the same motion.

Though important, the maximal EMG signal is a difficult value to collect. Many variables require the attention of the researcher. Anatomical factors such as fiber pennation angle, joint angle, and joint velocity impact the EMG signal magnitude [Farina 2001, Cresswell 1995, Bigland 1954]. Physiologic effectors act on the EMG signal repeatability as well. Rapid motor unit recruitment versus a slow increased recruitment pattern causes a significant alteration of the EMG magnitude [Ricard 2005]. The electrode can shift in location relative to the muscle body and motor units during a muscle contraction. The movement is an inherent factor of surface electrodes since the skin and muscle are not rigidly fixed to one another. This sliding motion can create signal alterations [Farina 2006]. Minimal publications have been presented on a subjects ability to elicit a maximal EMG signal.

The ability to reproduce a signal is paramount to its functionality as a reference value. Dynamic movements provide many hurdles to collecting reliable EMG signals. Each effect that causes a shift in EMG signal magnitude adds variability to the signal. Despite the added variability to the signals, previous studies have yielded within day repeatable EMG signals of the leg muscles in several ballistic tasks including: the lower leg muscles during running at a fixed speed on a treadmill [Chapman 2009, Karamanidis 2004] and rhythmic hopping [Gollhofer 1990], the quadriceps muscles during maximum vertical jumping [Goodwin 1999], and both the quadriceps and hamstring muscles during a sprint into a 45° cut maneuver and drop jumps from a fixed platform height [Fauth 2010]. Confidence in an EMG signal eliciting a maximum signal requires retest reliability in addition to within subject reliability and only a limited number of muscles have been compared for reliability of peak EMG signals either within day or between days [Ball 2010].

It is the purpose of this study to analyze the magnitude of EMG signals of 9 leg muscles during MVIC tasks, isokinetic (ISOK) tasks, and ballistic tasks, including counter-movement jumping and sprinting, to establish a task which reliably produces EMG signals greater than those produced by other activities. Since Rouffet and colleagues showed that EMG maximums elicited by dynamic tasks are greater than those produced during MVIC tasks for select muscles [Rouffet 2006], it is our hypothesis that the ballistic tasks will produce larger EMG signals than the static and controlled dynamic cases. Along with the findings of the EMG magnitude, within subject reliability and between day reliability will be established for a number of dynamic tasks.

3.2 Methods

3.2.1 Subjects

A sample of fifteen (15) healthy, uninjured males and females (mean age of 24.13±4.64 yrs, average height of 1.725±0.10m, and an average weight of 72.7±13.47kg) were recruited for study participation. Each subject participated in at least 50 hours per year of Level I, II or III activities as described by the International Knee Documentation Committee. Subjects were excluded from the study if they had any physical or neurological condition preventing them from walking, running or jumping normally. Subjects were also excused from the trials if they sustained any injury that would affect their movement patterns between testing sessions.

3.2.2 Instrumentation

The data for this collection was taken on a multitude of devices to collect specific data sets. The Biodex 3 System (Biodex Medical Systems, Shirley, New York) was used to fix joint angles and fix angular velocities during the MVIC and isokinetic (ISOK) tasks. EMG data were collected with a MA-300 EMG System (Motion Lab Systems, Baton Rouge, LA) for 9 leg muscles. The muscles included the medial and lateral vasti (VM, VL), rectus femoris (RF), biceps femoris (BF), semimembranosus (SM), medial and lateral gastrocnemius (MG, LG), soleus (SL) and tibialis anterior (TA). Bipolar silver/silver chloride EMG surface electrodes (Myotronics, inc., Kent, WA) were placed over these muscles in accordance with the *Anatomical Guide for the Electromyographer* [Perotto, 1994]. To ensure good conductivity, the electrode sites were shaved and lightly abraded with alcohol. Wires transmitting from the electrodes were connected to a receiver which was then connected to a long coaxial tether leading back to the data acquisition system. The receiver was placed on a backpack worn by the subject using Velcro. Electrodes were held in place with elastic wraps.

3D motion analysis was used for collecting joint angles, velocities, and ground reaction forces (GRF). Retro-reflective markers were placed on anatomical landmarks and marker motion was captured by an eight camera arrangement (Qualysis Motion Capture System, Gothenburg, Sweden). Rigid markers attached to shells were placed on the thighs, lower leg, and feet to establish a joint segment reference. Once a static trial is performed to gather calibration data, motion capture collected data for each of the dynamic movement trials.

3.2.3 Data acquisition and processing

EMG signals were collected at 1000Hz and motion data were collected at a frame rate of 200Hz. Data were processed using Visual 3D (C-Motion Inc., Bethesda, MD) and were visually inspected for excessive motion artifact or external signal noise. Signals were processed by high pass filtering using a 4th order Butterworth filter with a 30Hz cutoff frequency, removing DC offsets and full wave rectifying. A linear envelope was creating by low pass filtering the signals at 4Hz. The peak value of the linear envelope during the region of interest was the value used as the maximal signal of the trial.

3.2.4 Procedure

Subjects, informed of the experimental procedures, read and signed the informed consent document [Appendix B]. Following EMG electrode placement, subjects performed MVIC and ISOK trials on the Biodex 3 system. Contractions about the ankle and knee joint, at 90° and 60° of flexion respectively, were performed for the MVIC trial with the subject exerting maximal effort for three seconds of dorsiflexion, plantar flexion, knee flexion, and knee extension. The subjects performed ISOK tasks at 60° per second through the positive and negative directions of their full range of motion at each joint. Between each trial on the Biodex, the subject rested at least one minute to avoid fatigue. The subjects were given verbal encouragement throughout each of the contractions.

Following the completion of the Biodex trials, retro-reflective markers were placed on the subject and a standing calibration trial was performed followed by the dynamic trials of sprinting and counter movement jumping. The order of task was randomized to avoid compounding fatigue effects. The subject was permitted several practice trials to become comfortable with each dynamic task and then performed several acceptable trials. A Brower Timing Systems' timing system (Draper, Utah) and a 20-meter walkway with a floor level force plate was used in the laboratory for the dynamic trials. The sprint trial was performed at the subject's maximum running effort. Accepted trials required the speed to be within 5% of the other trials. The counter movement jump consisted of the subject moving from a standing position, into a squat, and then a maximal jump, using their arms to aid in reaching their maximum take-off force and height. A trial was accepted if the takeoff force was within 5% of the other trials. Subjects returned within a month for retesting.

3.2.5 Statistical Analysis

Analyses of the data were performed with the Statistical Package for the Social Sciences version 19. An ANOVA (alpha level set to $p<0.05$) with planned comparisons between the MVIC case, ISOK case, and the ballistic cases determined significant differences between peak EMG values. Intraclass Correlations Coefficients (3,1) were computed for each muscle during each task with in a single test session to establish reliability. Two-tailed paired t-tests (alpha level set to $p<0.05$) was used to find between day differences in EMG magnitudes.

3.3 Results

Statistical analysis of the data revealed several significant differences in peak EMG production across tasks of varying dynamic degree. The LG, MG, SL, VM, and VL all produced significantly larger signals during the dynamics cases, i.e. ISOK, sprinting, and jumping, compared to the MVIC case. The ballistic tasks were significantly larger than the controlled ISOK tasks for the triceps surae muscles, but the ISOK task was statistically larger than the ballistic tasks for the SM. Comparisons between the ballistic cases revealed sprint EMG maximums significantly exceeded those of the jump case for the SM and BF (Figure 3.1). Significance was established for only 5 muscles, but the mean EMG value for the MVIC case was the smallest condition for 7 of the 9 muscles.

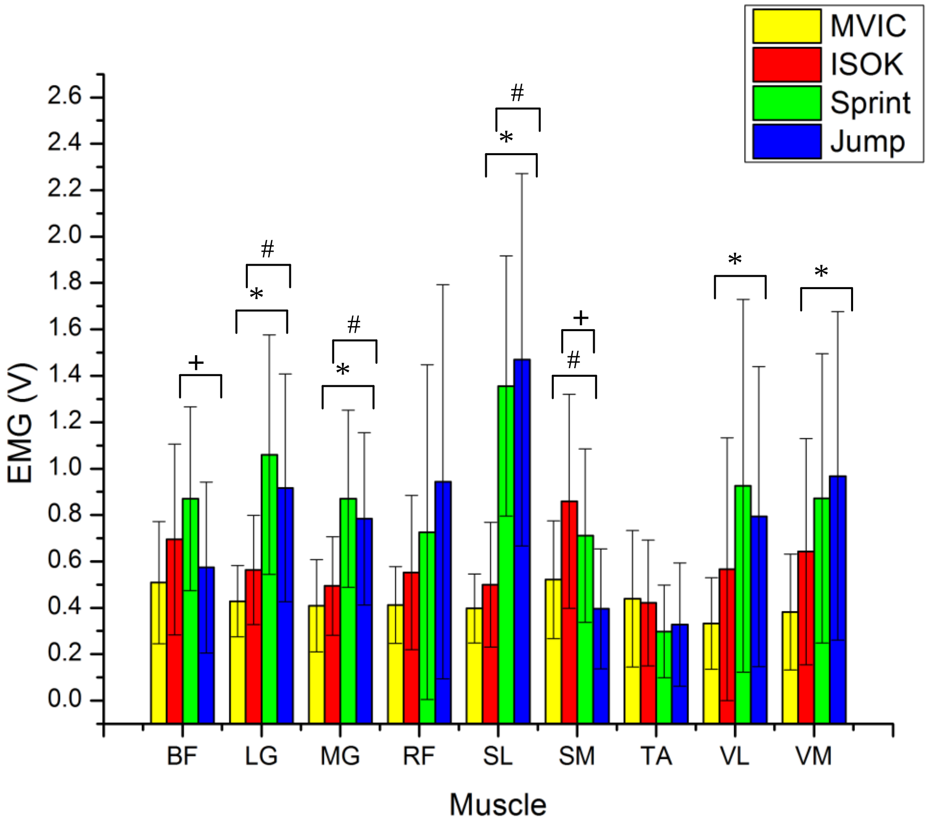

Figure 3.1. The mean peak EMG and standard deviation across the MVIC, ISOK, sprint, and jump tasks from the first collection session. The asterix (*) denotes a significant difference between the maximums of the MVIC and dynamic cases. The number sign (#) indicates a significant difference between the ballistic and the ISOK tasks and the plus sign (+) indicates a significant difference between the sprint and jump cases. Note: The MVIC is the smallest mean EMG value for 7 of 9 muscles.

The establishment of peak EMG reliability required the ICC values to be greater than 0.75 [Fleiss 1986] and that limit was exceeded for every muscle across all tasks except for a single muscle during a single task (Table 3.1). Only the SM was not able to reach the reliability threshold during the jump condition. It is worth noting that the EMG from the sprint method produced exceptionally reliable data with the signals from all muscles reaching an ICC value of 0.92 or greater.

Table 3.1. The within subject ICC values for each muscles' EMG maximum, with 0.75 being considered reliable. Note: The sprint trials' EMG signals were the most reliable and the jump trial was the only task with an unreliable signal (SM).

ICC	MVIC	ISOK	Sprint	Jump
BF	0.89	0.94	0.95	0.89
LG	0.86	0.85	0.96	0.95
MG	0.95	0.87	0.97	0.97
RF	0.92	0.97	0.92	0.95
SL	0.81	0.77	0.93	0.90
SM	0.91	0.88	0.94	0.68
TA	0.99	0.99	0.96	0.91
VL	0.93	0.86	0.95	0.95
VM	0.92	0.93	0.94	0.96

The inter-day EMG data set possessed only one significantly different alteration between peak EMG. The SL changed significantly across days for the ISOK task (Figure 3.2). The variability of all the muscles ranged from 2% to 38% between collection sessions for all of the mean peak EMG signals. The average difference between collection sessions across tasks was 11% for the MVIC, 22% for the ISOK, 17% for the sprinting, and 17% for the jumping conditions. The VL, on average, varied the least between days while the TA had the most day to day change.

Assessment of statistical significance of the second collection session determined similarities in maximal elicited signals between days. The LG, MG, and SL EMG values of the ballistic cases repeated the significantly larger magnitudes in relation to the MVIC and dynamic cases which were determined on day 1. (Figure 3.1, Figure 3.2). The second collection session maintained statistically increased maximums of dynamic cases versus the MVIC for the VL and VM (Figure 3.2).

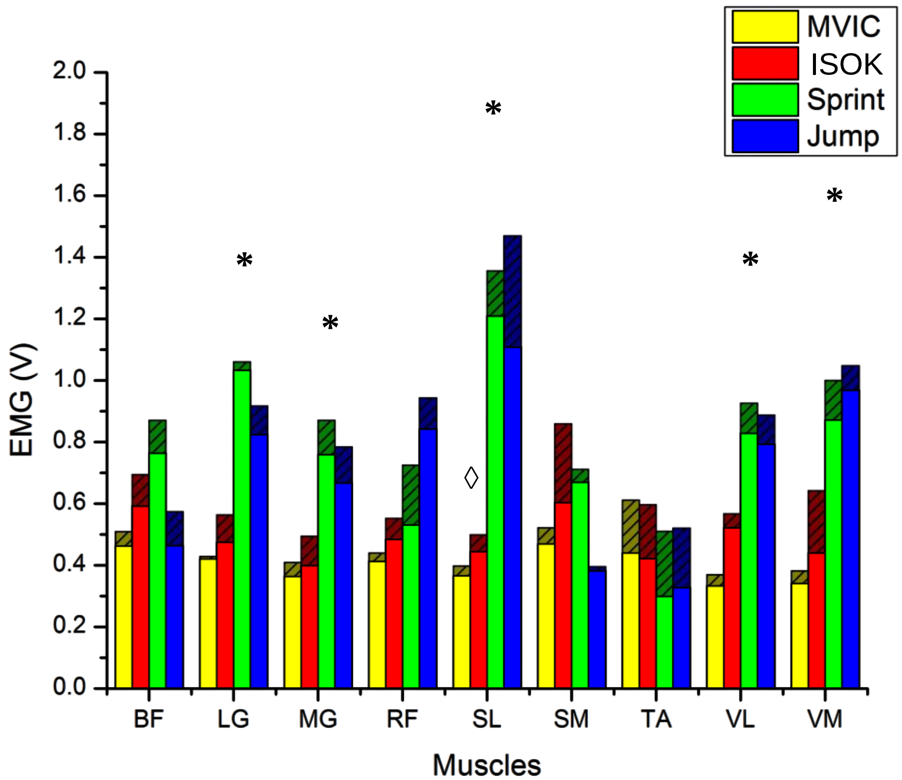

Figure 3.2. The differences in mean peak EMG signals collected on day 1 and day 2. The cross-hatched area represents the magnitude of change between the two collection sessions. The EMG collected from the soleus during the ISOK task the first day was significantly different than the maximum collected on the second day, denoted by the diamond (◊). The asterix (*) denotes the significant differences determined on the first day were same as the significant differences found on the second day. Note: The second day significant differences matched those of the first day for the LG, MG, SL, VL, and VM.

3.4 Discussion

3.4.1 Maximal task

The data clearly showed that the MVIC task does not produce the maximal EMG signal, which is supported by the literature. The varying anatomical and physiologic factors occurring during the ballistic tasks were enough to cause a significant increase in EMG. The muscles which yielded a significant increase in EMG during the ballistic tasks were the propulsive muscles used for push-off for both the sprint and the jump cases. The vasti are used exclusively as knee extensors and were recruited rapidly to generate a maximum speed for sprinting and jumping to a maximal height. The rapid muscle recruitment was accompanied by massive joint velocities (Table 3.2). To further understand the relationship between EMG and joint angular velocity, the percent difference between the maximum EMG of the MVIC and sprint cases were compared (Figure 3.3). The uniarticulate VM and VL provide an excellent joint angle to muscle length relationship, and therefore were analyzed during sprint task for comparison. The VM and the VL had a positive correlation between the EMG of the knee extensors and the joint velocity. The LG, MG, and SL during the ballistic tasks also produced significantly increased maximum EMG values so it would be logical to compare the ankle joint velocity to the EMG, but the anatomical dissimilarities of preventing a similar extrapolation. The ankle velocity during push off of both the jump and sprint phase were greater than the 60°/s of the ISOK case (Table 3.2), but the EMG/velocity plot does not correlate well to the shift in EMG from the MVIC to the sprint or jump cases. The lack of correlation may be attributed to the variable nature of the Achilles tendon length in loading and unloading based on force applied or operational velocity [Herbert 2002]. The Achilles tendon does not yield a uniform muscle fiber length change with joint angle change, thus preventing a reasonable regression fit.

Table 3.2 The mean kinematic data found for the ballistic trials. The angular velocities and range of motion (RoM) measurements were collected during the stance phase of sprinting and the takeoff phase of jumping. These values are supported by the findings of Dutto and Mann [Dutto 2004, Mann 1980].

Kinematics	Sprint	Jump
Knee Velocity (°/s)	151.44	373.01
Ankle Velocity (°/s)	197.45	538.50
Knee RoM (Degrees)	29.16	102.82
Ankle RoM (Degrees)	38.20	74.98

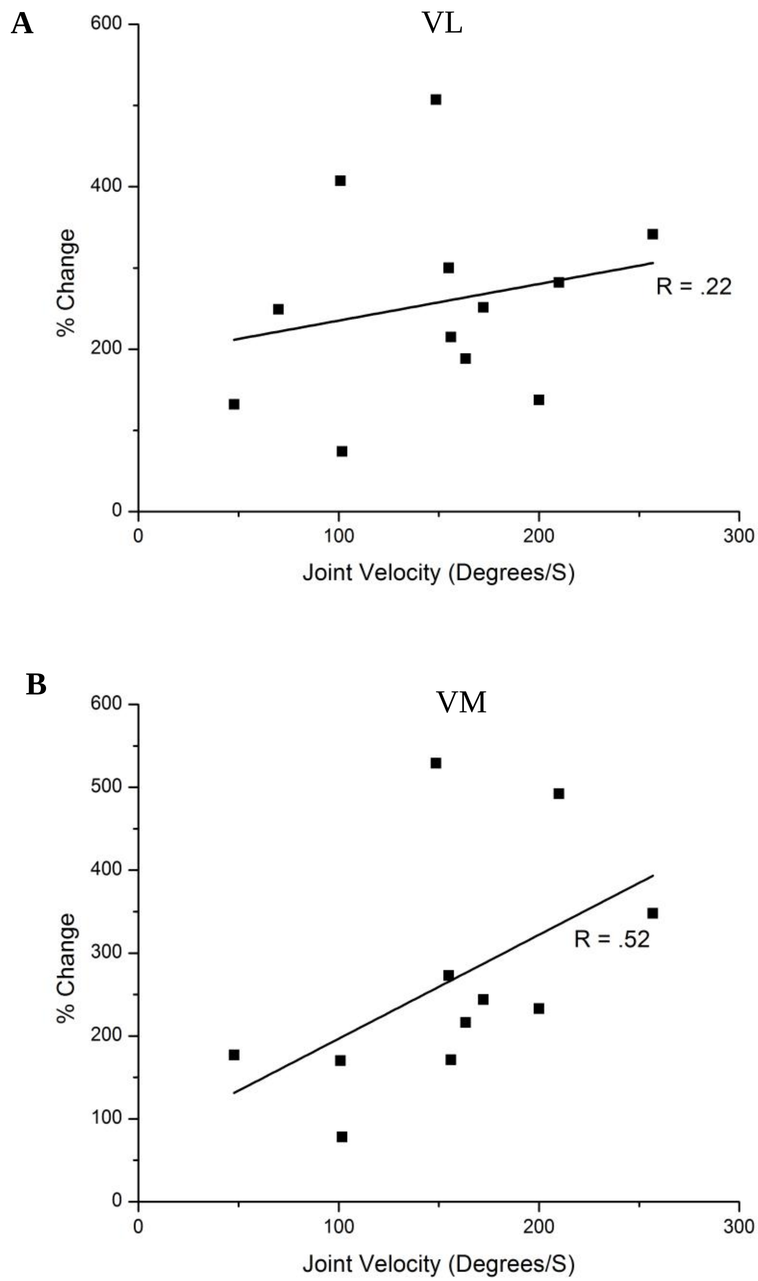

Figure 3.3. The percent change in maximum EMG from MVIC to sprint as compared to the angular velocity about the knee. Figure A is the EMG shift of the VL and figure B is the EMG shift of the VM. Note: The regression line slopes positively for both muscle groups.

The hamstrings, which operate primarily as knee flexors, were not significantly different between the static case and the dynamic cases. While the ISOK was consistently larger than the MVIC task, the sprint kinematic variability contributed to a large standard deviation and a lack of significance between the sprint maximums and the MVIC maximums. The subjects' hamstring recruitment deviated from one another because of the level of sprint training each subject had received. while the novice runners use the hamstrings as an eccentric, momentum reducing muscle during the swing phase, the elite runners use the hamstrings as a concentric, pulling muscle to aid in propulsion during stance. While novice and elite runners both use the quadriceps and triceps surae as propelling muscles, the variation between trained runners' hamstring recruitment patterns and those of the untrained runners [Novacheck 1998] lead to a large deviation in task maximum. The functionality of the hamstrings during the jump task was the cause of the decreased SM and BF EMG signals. The jumping task was collected during the push off phase in which the hamstrings are acting only as hip extensors and not knee flexors. The hamstrings are most active during leg flexion and therefore this constriction of functionality produced non-maximal EMG signals.

Previous studies have shown EMG magnitudes change as a function of joint angle during a task. The ISOK trials encompassed the subjects' entire range of motion and the jump trials were very similar, but the sprint had a very limited range of motion during the stance phase of gait (Table 3.2). Babault and colleagues showed the quadriceps producing their largest signals when the muscle was at its most disadvantaged, i.e. at its shortest length [Babault 2003]. With the ISOK and jumps trials acting over the entire joint range and the knee extensors and ankle plantar flexors at the end of travel during the sprint, these muscles were at their maximum EMG joint angles during each of dynamic activities. In the context of the signal collection, when the muscle was fully contracted and at its shortest length, the electrode was collecting from the maximum volume of that electrode location [Rainoldi 2000] also supporting the conclusion that the dynamic tasks were at the optimal joint angle for maximum signal output.

The maximal signal is a tool that will be used for a healthy and capable population, since that is the group which is able to sprint and jump at their maximal efforts, but the ability to perform the task is not the only factor limiting participation. The first points of the correlation graphs (Figure 3.3) are located at approximately 100%. If a subject is unable to perform a dynamic task maximally, it may imply they would not generate an EMG greater than the signal they could elicit during a static, constrained task. This removes the need for a ballistic collection when attempting to find a maximal EMG signal on an injured or incapable population.

3.4.2 Within subjects reliability

The within subjects reliability produced very acceptable results. The MVIC reliability for each muscle exceeded an ICC value of 0.80, as predicted by previous literature. Along with the static case, the reliability of the controlled ISOK case met expectations. Potential variability was introduced into the ballistic trials which could have drastically decreased reliability; however the added motion artifact, potential inconsistency of kinematics, and increase risk of fatigue did not alter the consistency of the ballistic EMG. The limited amount of motion artifact seen in the ballistic trials was able to be removed through high pass filtering of the signal. The mitigation of fatigue was achieved though the allowance of proper rest between trials, which successfully prevented altered kinematic performance and signal loss. The ICC values for 17 of the 18 ballistic cases successfully exceeded the acceptable 0.75 value for reliability. While other studies have reported reliability for similar tasks such as drop jumps, running, and cutting [Fauth 2010, Chapman 2009, Gollhofer 1990], it has not been previously demonstrated that sprinting and maximal vertical jumping are capable of being reliable for this number of muscles. The exceptional reliability produced by the sprinting tasks (ICC $\geq$ 0.92) showed the ability to successfully collect EMG during gait tasks at any velocity. The single muscle, which did not reach reliability, was the SM during the jump. These findings were similar to the results of Goodwin and colleagues [Goodwin 1999]. The SM, acting as a hip extensor and not a leg flexor, was not acting in its primary force-generating capacity during the jump trials; the resulting changes in jumping technique to produce major changes in signal output.

3.4.3 Inter-day repeatability

Significant increases of peak EMG values of the maximal task must be present across testing session for the value to be considered constructive. To establish reliability of the maximal values, the retest peak EMG values were assessed for significant differences as well. This resulted in the discovery that the significantly larger values of the first day also were large on the second day of testing. The signals of the LG, MG, SL, VL, and VM were significantly greater during the dynamic cases than the MVIC trials. The reliability of these signals can be attributed to the reliability in kinematics. Previous studies have demonstrated subjects able to consistently repeat a range of motion and joint velocity for both sprinting and maximal counter movement jumping [Yoon 2006, Queen 2006].

3.4.4 Conclusion

The EMG maximums of 9 muscles were compared across 4 tasks with maximal exertion held constant and joint velocity, joint range of motion, and various degrees of kinematic freedom. The limited range of motion of the sprint task did not prevent the propulsive muscles from emitting maximal signals that rivaled the maximum EMG signals of the jump task. Contrary to range of motion not being the definite cause of maximal EMG production, the joint velocity positively correlated to a shift in EMG magnitude, when analyzing a single joint muscle (VL and VM). The significant EMG maximums (Figure 3.1) demonstrated that ballistic movements can produce reliable peak EMG values (Table 3.1) that are close to maximal. Unfortunately, no single activity was identified as the producer of maximal signals, since neither ballistic EMG activity yielded a statistically greater peak EMG than the other. The small changes to the EMG means and the repeated significance between days combined with the inter-subject reliability creates a confidence in using data from the ballistic test for normalization and other comparisons. This understanding enables future studies to normalize to these values, compare muscles across activities, and compare co-activation between subjects and tasks.

Chapter 4

EMG NORMALIZATION EFFECTS ON AN EMG-DRIVEN MUSCLE MODEL

4.1 Introduction

Analysis of muscle activation produced during activities ranging from the daily life motions of walking and stair climbing to the athletic motions of running and jumping requires the collection of reliable electromyography (EMG) signals measured under dynamic conditions. EMG signals are recordings of the electrical activity that are associated with muscle contractions; these signals give insight into the function of the muscles based on contraction timing and intensity. EMG collects signals from the muscle body after being stimulated by the muscle's motor unit impulse, which is in response to a neural command. Though EMG is representative of muscle's activation, muscle force production and muscle activation do not equate to each other in a linear manner [Buchanan 2004]. An EMG signal of a single muscle is an independent value in regards to other EMG signals of separate muscles. EMG is measured as a voltage and is highly dependent on many factors other than contraction intensity, including electrode placement, skin impedance, joint angle [Farina 2001], and joint velocity [Bigland 1954]. Due to the factors affecting the signal, it is important to normalize the EMG values before the signal is input into an EMG-driven muscle model.

The EMG-driven muscle model is a computational based algorithm model which requires a series of inputs, including EMG, to predict individual muscle forces. The EMG inputs are used to determine the muscle activation intensity. The activation intensity scales each muscles output force based on its maximum producible force [Buchanan 2004]. The model has been tested repeatedly for low intensity movement inputs, which were normalized to the consistently repeatable EMG values produced during maximum voluntary isometric contraction (MVIC) tasks. After normalization, the EMG values of the lower intensity movements were below 100% activation. Higher velocity, or ballistic, tasks often produce EMG magnitudes greater than those generated during MVIC tasks [Gazendam 2007]. Subjects' individual maximum muscle forces and cross sectional areas are not input into the EMG-driven model. This requires the model to adjust the maximum isometric force of each muscle starting from the average values reported in literature [Winter 1990]. Therefore, normalizing to EMG values produced by isometric tasks could produce muscle inputs much greater than 100% and lead to the incorrect conclusion that a muscle is producing forces outside of reasonable, physiological bounds. This mistake is avoided by normalizing to the maximal EMG value.

Finding an activity to yield a maximal EMG requires attention to many factors including various joint angles, velocities and contraction magnitude, and muscle fiber alignment. Muscle contraction intensity and degree of contraction impulse strongly influences EMG. The EMG signal is a resultant of the number of active motor units and the rates at which they are firing [Guo 2010]. As a contraction increases, the firing rate of the active motor unit increases. Once reaching a maximum frequency threshold of the first motor unit, an additional motor unit is recruited and this continues until the system is at its maximum output [Person 1972], therefore a maximum effort contraction is needed to produce the maximal EMG. A MVIC is an example of a task requiring maximum contraction intensity while holding other variables constant leading to a muscle producing its maximum force [Hill 1938]. Holding other variables constant and exerting a maximum contraction is what causes MVIC to have an excellent reliability, leading to it becoming a common normalization value. Unfortunately, the MVIC value, because of the other influential factors, is not maximal.

MVIC tasks produce reliable EMG data and yield the maximum force of a muscle but this is not always the normalization value of choice. Ballistic EMG normalized to MVIC values can lead to EMG values above 100% activation [Burden 1999]. Since maximum force does not guarantee maximum EMG production [Buchanan 2004] and dynamic trials exist that produce larger EMG signals, dynamic EMG has an especially important role in the EMG normalization of high velocity movements being assessed in the EMG-Driven model. There is a wide variety of movements from which to record dynamic EMG signals ranging from controlled isokinetic contractions to much more variable ballistic movements. Ballistic tasks differing from jumping and hopping to running and cutting have been studied briefly, but previous research has only minimally compared the magnitudes of the resulting EMG and EMG-driven muscle models have not been assessed with differing normalization techniques.

The purpose of this study was to determine if EMG inputs of greater than 100% activation affect the EMG-driven muscle model. Our hypothesis was the smaller input activations, determined by normalizing to larger EMG maximums, would produce smaller joint moment outputs from the model. The applications of computer based muscle models for use during rapid dynamic tasks are numerous and it is important to evaluate the model's output variation between the normalizing methods of maximal and submaximal tasks when analyzing faster movements.

4.2 Methods

A sample of fifteen (15) healthy, uninjured males and females (mean age of 24.13±4.64 yrs, average height of 1.725±0.10m, and a weight of 72.7±13.47kg) were recruited for study participation. Each subject participated in at least 50 hours per year of Level I, II or III activities as described by the International Knee Documentation Committee. Subjects were excluded from the study if they had any physical or neurological condition preventing them from walking, running, or jumping normally. Subjects were also excused from the trials if they sustained any injury that would affect their movement patterns between testing sessions.

4.2.1 Instrumentation

The data for this collection was taken on a multitude of devices to collect specific data sets. The Biodex 3 System (Biodex Medical Systems, Shirley, New York) was used to fix joint angles and fix angular velocities during the MVIC and isokinetic tasks. EMG data were collected with a MA-300 EMG System (Motion Lab Systems, Baton Rouge, LA) for 9 leg muscles. Those muscles included the medial and lateral vasti (VM, VL), rectus femoris (RF), biceps femoris (BF), semimembranosus (SM), medial and lateral gastrocnemius (MG, LG), soleus (SL) and tibialis anterior (TA). Bipolar silver/silver chloride EMG surface electrodes (Myotronics, inc., Kent, WA) were placed on these muscles in accordance with the *Anatomical Guide for the Electromyographer* [Perotto 1994]. To ensure good conductivity, the electrode site was shaved and lightly abraded with alcohol. Snap wires transmitting from the electrodes were connected to a receiver which was then connected to a long coaxial tether leading back to the data acquisition system. The receiver was placed on a backpack worn by the subject using Velcro. Electrodes were held in place with elastic wraps.

Motion analysis was used for collecting joint angles, velocities, and ground reaction forces (GRF). Retro-reflective markers were placed on anatomical landmarks. Marker motion was captured by an eight camera arrangement (Qualysis Motion Capture System, Gothenburg, Sweden). Rigid markers attached to shells were placed on the thighs, lower leg and feet to aid in motion capture, which was used for each of the dynamic movement trials.

4.2.2 Data acquisition and processing

EMG signals were collected at 1000Hz and motion data were collected at a frame rate of 200Hz. Data was processed using Visual 3D (C-Motion Inc., Bethesda, MD) and was visually inspected for excessive motion artifact. Signals were processed by high pass filtering using a 4th order Butterworth filter with a 30Hz cutoff frequency, DC offsets were removed and the signal was full wave rectified. A linear envelope was created by low pass filtering the signals with a 4th order Butterworth filter with a 4Hz cutoff frequency. The peak value of the linear envelope during the region of interest was the value used as the maximal signal of the trial.

4.2.3 Procedure

Subjects informed of the experimental procedures read and signed the informed consent document. Following EMG electrode placement, subjects performed MVIC and isokinetic trials on the Biodex 3 system. Contractions about the ankle and knee joint, at 90° and 60° of flexion respectively, were performed for the MVIC trial with the subject exerting maximal effort for three seconds of dorsiflexion, plantar flexion, knee flexion, and knee extension. Between each trial on the Biodex, the subject rested at least 1 minute to avoid fatigue. The subjects were given verbal encouragement throughout each of the contractions.

Following the completion of the Biodex trials, retro-reflective markers were placed on the subject and a standing calibration trial was performed followed by the dynamic trials of sprinting and running. The subject was permitted several practice trials to become comfortable with each dynamic task and then perform several acceptable trials. A Brower Timing Systems' timing system (Draper, Utah) and a 20-meter walkway with a floor level force plate was used in the laboratory for the dynamic trials. The sprint trial was performed at the subject's maximum running effort. Accepted trials required the speed to be within 5% of the other trials. A series of running trials were performed at 4.0 m/s, within 5%, to create the model inputs. The model requires several inputs including EMG, knee and ankle moments, muscle tendon lengths and joint moment arms. The force and motion data were filtered using a low pass, 4th order Butterworth filter with a 6Hz cutoff frequency to remove the high frequency noise content. Knee and ankle moments were output from Visual3D. Motion data was processed with OpenSim (https://simtk.org/) to create the subject specific moment arms and muscle tendon lengths. To perform a comparison of normalization techniques, the model was executed with the same inputs except for normalization of the EMG to the MVIC and sprint cases, thus creating separate inputs.

4.2.4 Statistical Analysis

Analyses of the data were performed with the Statistical Package for the Social Sciences version 19. Paired t-tests, with a significance level $\alpha=0.05$, determined significant differences between peak knee moments of the normalization methods. Differences in the r-squared values of the curves, in relation to the inverse dynamics, from the output of the model were also compared.

4.3 Results

The model was able to match the inverse dynamic the knee joint moment with both normalization cases (r^2 values greater than 0.90). The MVIC case matched the inverse dynamic results better than the dynamic normalization case demonstrated by the larger r^2 values and lower error in peak moment (Figure 4.1). The average differences between the peak moments calculated by inverse dynamic case and the model predicted moments were 10.4 N-m from the MVIC case and 32.1 N-m from the sprint case. This represented a significantly different result between outputs.

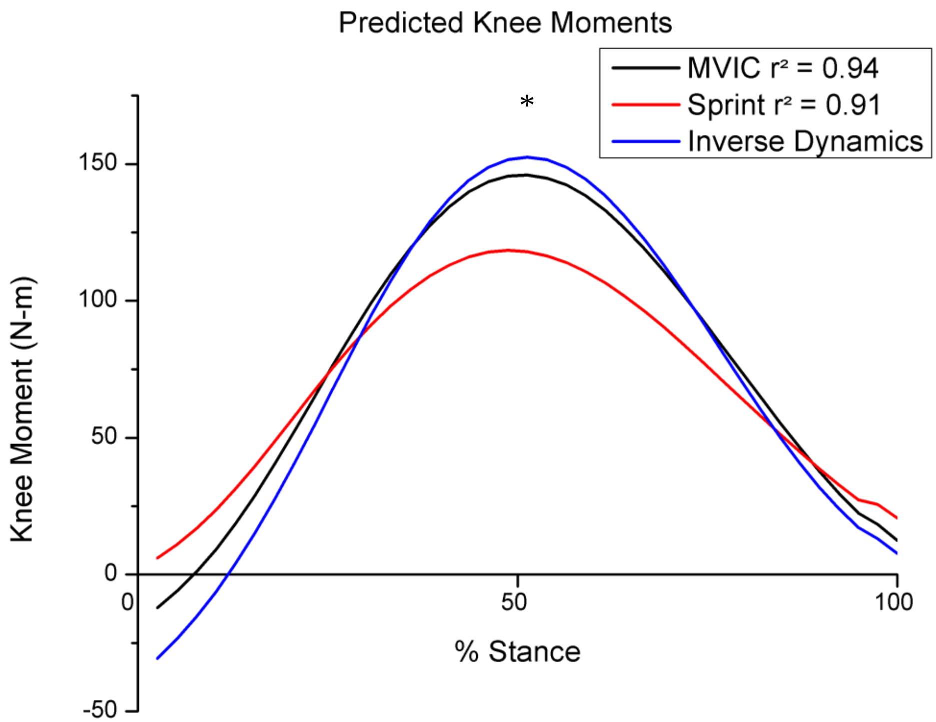

Figure 4.1. Predicted knee joint moments from the EMG-driven model with EMG inputs normalized to the MVIC and sprint maximums and the knee joint moment calculated through inverse dynamics. The asterix (*) denotes a significant decrease in error, with respect to the inverse dynamic value, of the peak knee moments from the two different normalization methods. Note: The curve matched better between the MVIC and inverse dynamic cases as identified by the larger mean r^2 values (located in the legend).

4.4 Discussion

The EMG-driven model generated knee moment curve in both the MVIC and sprint normalization cases matched the moment curve calculated by inverse dynamics. These finding were similar to those of Besier and colleagues [Besier 2009]. Our hypothesis was confirmed. The sprint EMG normalization produced a decreased joint moment compared to the MVIC case, and was less accurate as a result. The increased EMG inputs of the MVIC normalization case allowed the muscles to reach the necessary required level of estimated forces to match the knee moment assessed by the inverse dynamics. The muscle moment calculations involved individual joint moment arms multiplied by predicted muscle forces. The muscle force calculation combines several optimized musculotendon parameters, including knee flexor and extensor gain factors, and the EMG-driven muscle activations. The gain factors act a multiplier of the maximum isometric muscle force, determined from literature [Winters 1990], to adjust for individual subjects' differences in maximum strength. The muscle and tendon properties of the model were constrained to maintain acceptable physiologic values. The extensor and flexor gains were limited from 0.50 to 1.50 of the reported average maximum isometric force for each muscle. The tight gain limits prevented the sprint normalization case from obtaining a greater degree of accuracy during the moment prediction. During the optimization, the model reached the maximum gain limits of 1.50 which prevented the muscle forces from obtaining the necessary maximum forces required to reach the calculated inverse dynamic moments. The restricted forces predicted increased knee moments at heel strike and a lower overall peak knee moment at toe off (Table 4.1).

Table 4.1. The mean gain factors of the extensors and the flexors calculated during the EMG-driven muscle model optimization. Note: The sprint and MVIC cases were both reaching the extensor gain limit. The flexor gain limits were closer to reaching the maximums for the sprint case verses MVIC case.

	MVIC – Extensors	**MVIC - Flexors**	**Sprint - Extensors**	**Sprint - Flexors**
Gains	1.07	1.50	1.46	1.27

The gain regulates the increase or decrease in muscle force production. Understanding the effects of the gains on the model is paramount in knowing where to set the limits of the gain factors. A sensitivity analysis was performed on the 3 least matching trials by increasing the gains from 1.5 to 2.0 and 2.5 (Figure 4.2). As the gain factors were relaxed on the EMG-driven model, the sprint and the MVIC normalization cases were able to determine more accurate predictions. Since the sprint case accuracy was the lesser of the two, it produced greater improvements with each gain iteration, but was still not able to obtain the lesser deviation from the inverse dynamic obtained by the MVIC normalization case. (Figure 4.2).

A complete understanding of the better normalization method requires the need for an isometric force test which can isolate a group of muscles for the scaling of an individual's maximum, which is not a simple task due to the needed inclusion of co-activations and passive elastic tissue stresses. Since we are not able to determine the actual maximum of the subject, we rely on the models predictions and the limits of human physiology.

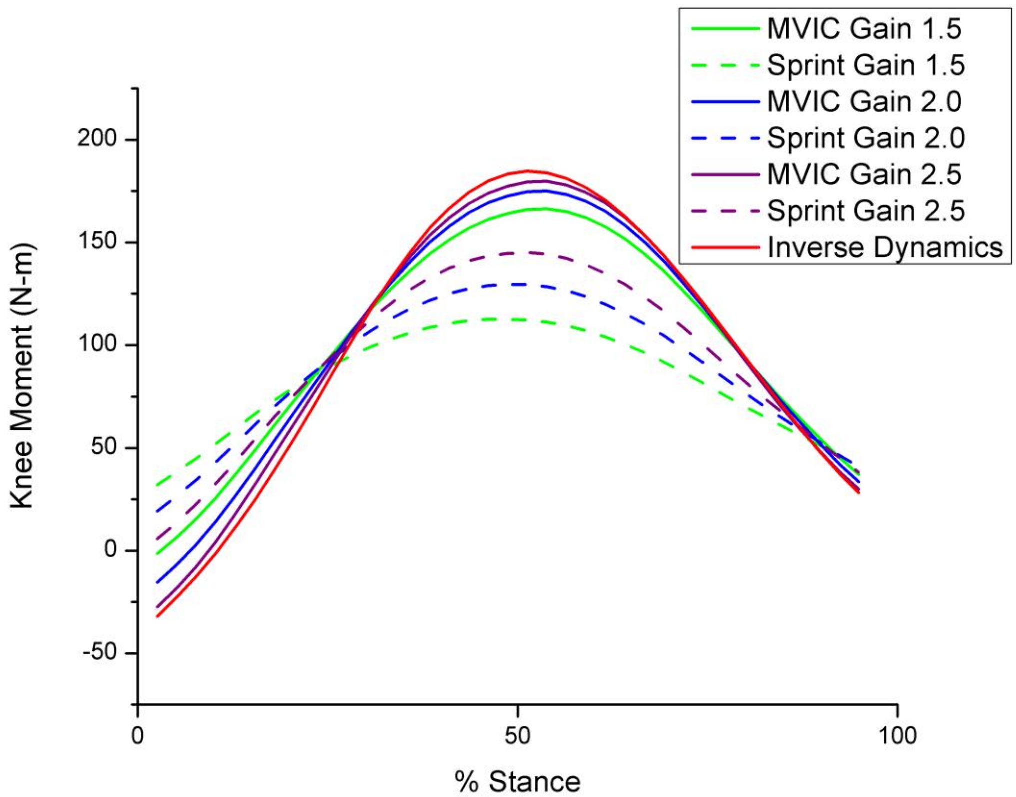

Figure 4.2. The mean knee extension moment of the 3 least accurate estimations during the stance phase of gait. Normalizations and gain parameters were changed to understand the effects of increasing the gain on both conditions. The sprint normalization method is plotted with dashed lines and the MVIC normalization method is plotted with solid lines. Note: As the gain was relaxed, i.e. the constraints were increased, the curves matched the inverse dynamic case better.

4.4.1 Conclusion

The MVIC normalization input produced a better match to the inverse dynamics without having to greatly increase the estimated muscle force gains. Normalizing dynamic EMG to a maximal EMG signal was not essential for predicting muscle forces. The EMG-driven model was able to adjust muscle force parameters leading to more accurate knee moments with the MVIC normalization method. Though our hypothesis was correct in the assumption of the decreased EMG would cause a smaller moment, it created too small of a moment to match the inverse dynamics without amplification of the subject's maximum isometric force to abnormal physiologic levels.

Chapter 5

CONCLUSION

This thesis focused finding a single task to produce a maximal EMG signal which was also reliable. To perform this test, subjects performed controlled dynamic and ballistic tasks while EMG signals were collected from 9 leg muscles. Significantly greater signals were found during the ballistic tasks, jumping and sprinting, than the static and ISOK tasks for the triceps surae. Dynamic tasks, including both the ballistic and ISOK tasks, were significantly greater than the MVIC case for the vasti, gastrocnemii, and soleus. It was also found that all the EMG signals produced during the ballistic tasks were reliable, less the SM during the maximal jumping case.

This thesis also assessed the normalization methods for modeling dynamic motion with an EMG-driven muscle force estimation model. The model's inputs, from the stance phase of a run task, were normalized with the near maximal sprint and sub-maximal MVIC EMG signals and the results were compared. The forward dynamic outputs from the model were compared with the calculated inverse dynamic knee moment. The MVIC case produced the more accurate moment results. Though only a single task was used for normalization, to obtain the true maximum of each muscle a series of tasks would need to be performed and the maximums would come from several tasks.

5.1 Major findings

5.1.1 Maximal EMG signals

The variable nature of EMG signals creates a difficulty in determining if a maximal signal was elicited by a subject. Typically, if a maximum signal is desired, researchers have subjects perform a MVIC task, but this is not the generator of maximal signals. This study showed that if a true maximum is desired, a ballistic task must be performed in which the desired muscles are acting in their primary functional manner. This permits the muscles to be fully activated and for the muscle to perform a burst contraction to reach its largest possible signal. The repeatable nature of these maximal signals was also established in this study. The performance of test and retest collections determined the ability of subject to maintain the same significant level of increased magnitudes between two separate days.

5.1.2 Dynamic signal reliability

Collecting EMG at high velocities introduces the possibility of increased signal altering factors including increased motion artifact, muscles movement beneath the skin relative to the electrode location, and the onset of muscle fatigue. Though there was the possibility of error in the signal, reliable EMG signals were still able to be collected with specific details of the collection process being addressed. Limiting motion artifact with adhesive electrodes followed by proper muscle wrapping reduced the skin artifact. The remaining low frequency component caused by motion artifact was removed by high pass filtering. Muscle movement was addressed through ensuring repeatable joint range of motion during each of the tasks. Fatigue was managed through sufficient rest for the subjects between trials. Attention to each of these details created reliable ballistic EMG signals.

5.1.3 Effects of EMG normalization

The EMG-driven model has produced estimations of muscle forces and knee moments for slower and more controlled movements, but the model is progressing towards predicting muscle forces and joint loading for more dynamic activities. The increased EMG signals from the more dynamic conditions called into question the typical method of normalizing to MVIC by producing activations above 100%. This study determined the increase in EMG signals did not negatively affect the model. To the contrary, the model produced knee moments closer to those of the inverse dynamics using the MVIC normalization method versus normalizing to a near maximal ballistic task without adjusting muscle parameters outside of reasonable physiological bounds.

5.2 Contributions

This thesis conclusively determined ballistic tasks exist which produce peak EMG signals greater than those produced by MVIC tasks as well as establishing that the relative magnitudes increases from the ballistic tasks could be repeated on a separate day. Previous studies alluded to this fact [Gazendam 2007] but no studies established significant magnitude changes between static and dynamic tasks nor have studies shown this fact for most of the major muscles in the leg. Significant differences found during ballistic tasks aids in the comparison of co-contraction and required activation to perform sub-maximal tasks. This study also showed the repeatability of a peak EMG value within a testing session of multiple dynamic tasks, confirming some of the findings of previous literature [Fauth 2010, Goodwin 1998, Gollhofer 1990], but on a much broader array of muscles. Finally, this study determined normalizing dynamic tasks to the MVIC case is an acceptable method when modeling muscle forces with the EMG-driven model.

5.3 Limitations

EMG studies are limited by the variability of subject's physical traits. Since this study required individuals to run faster than 4.0 m/s and jump maximally, a certain fitness level existed among our subject population. Much of our population consisted of active, young adults, which led to very low body fat. The decreased presence of adipose tissue permitted the collection of much cleaner signals. It reduced the amount of soft tissue error with less tissue to transmit EMG signals through and provided a lower skin motion artifact during the very dynamic activities. It is unclear whether the same repeatability would have been reached with a population with a higher body fat percentage.

The study also has a kinematic limitation. The joint angle range of motion for the sprint case was only assessed for the stance phase of gait. During the stance phase of gait, the plantar flexors and knee extensors were at their shortest. This was assumed to be when those muscles were at their peak EMG output, which was supported by the literature [Azegami 2007, Maffoietti 2003, Suter 1997]. Due to the limited motion capture volume we were not able to collect the range of motion for the knee flexors or the ankle dorsiflexors and therefore could not determine if the peak EMG values did not reach significance because of the task or because of a limited range of motion. The kinematic limitation also applied to the models force estimation. Since the model is tuned to the inverse dynamic knee moment, we were only able to calculate the knee moment during the stance phase of gait.

The final limitation of the study is due to the unknown forces in the individual muscles. The gain coefficients strongly influence the muscle force output and joint moment. Without knowing the maximum output force of an individual subject, it is difficult to determine the limits to apply to the gains. Our limits were set by previous subject research [Winters 1990].

5.4 Future work

The application of EMG signals to human motion analysis, future pathologies, and muscle modeling requires a great deal of understanding of the EMG tool itself. While this work was not able to determine a maximal signal between ballistic tasks, it was determined that maximal ballistic tasks produce much larger signals in the propulsive muscles of those tasks. This knowledge can be applied to finding a ballistic task which can produce maximal activity in the knee flexors and ankle dorsiflexors. Having a complete set of tasks producing near maximal signals will enable accurate muscle co-contractions and required muscle exertion within a task to be identified.

Reliability knowledge of a healthy population is the first step in using EMG for a pathologic population. Since sprinting is the highest velocity gait task, and therefore had the most variability introduced into the signal, it can be reasonably assumed that running at a slower speed, for healthy individuals, will be equally reliable. If this is the case, we can assess the variability of different pathologic populations to determine decreases in muscle activity and recognize muscle avoidance, inhibition, or decreased function. Knowing which muscles are consistently acting maximally during a task will allow us to make changes to those muscles and evaluate the kinematic response.

Understanding the more reasonable normalization value permits muscle force estimation with greater dynamic conditions. The model can now be utilized for rapid movements without the uncertainty of normalization task. The model can be utilized under dynamic conditions for task retraining, knee compartmental loading analysis, identifying risks of injury, and understanding recovery progression.

5.5 Summary

This thesis analyzed the maximum EMG signal output of BF, LG, MG, RF, SL, SM, TA, VL, and VM muscle during several dynamic tasks for comparison against the maximum peak EMG signal of the MVIC. It was determined that 5 of the 9 muscles yielded significantly greater EMG maximums during the dynamic tasks compared to the static. It was also seen that the MVIC produced the lowest average peak EMG for 7 of the 9 muscles. The peak dynamic EMG produced reliable signals, with the exception of a single case, within a testing session and maintained the large differences in reference to the MVIC across days. The ballistic and MVIC peak EMG values were used to normalize running EMG as an input for an EMG-driven muscle model. The MVIC case produced a better knee moment match to the inverse dynamics than the sprint normalization methods. The establishment of the better normalization task allows for future dynamic tasks to be optimized by the model with a greater level of confidence.

REFERENCES

Adam A and De Luca CJ, Firing rates of motor units in human vastus lateralis muscle during fatiguing isometric contractions. J Applied Physiol, 2005. 99(1):p.268-280.

Azegami M, Yanagihash R, Miyoshi K et al., Effects of multi-joint angle changes on EMG activity and force of lower extremity muscles during maximum isometric leg press exercises. J Phys Ther Sci, 2007. 19(1):p.65-72.

Babault N, Desbrosses K, Fabre MS et al., Neuromuscular fatigue development during maximal concentric and isometric knee extensions. J App Physiol, 2006. 100(3):p.780-785.

Babault N, Pousson M, Michaut A et al., Effect of quadriceps femoris muscle length on neural activation during isometric and concentric contractions. J Appl Physiol, 2003. 94(3):p.983-990.

Ball N and Scurr J, An assessment of the reliability and standardisation of tests used to elicit reference muscular actions for electromyographical normalisation. J Electromyogr Kines, 2010. 20(1):p.81-88.

Behm DG, Wahl MJ, Button DC et al., Relationship between hockey skating speed and selected performance measures. J Strength Cond Res, 2005. 19(2):p.326-331.

Benoit DL, Lamontagne M, Cerulli G et al., The clinical significance of electromyography normalization techniques in subjects with anterior cruciate ligament injury during treadmill walking. Gait Posture, 2003. 18(2):p.56-63.

Besier TF, Fredericson M, Gold GE et al., Knee muscle forces during walking and running in patellofemoral pain patients and pain-free controls. J Biomech, 2009. 42(7):p.898-905.

Bigland B and Lippold OCJ, The relation between force, velocity and integrated electrical activity in human muscles. J Physiol-London, 1954. 123(1):p.214-224.

Buchanan TS and Lloyd DG, Muscle-Activity Is Different for Humans Performing Static Tasks Which Require Force Control and Position Control. Neurosci Lett, 1995. 194(1-2):p.61-64.

Buchanan TS, Lloyd DG, Manal K et al., Neuromusculoskeletal modeling: estimation of muscle forces and joint moments and movements from measurements of neural command. J Appl Biomech, 2004. 20(4):p.367-395.

Burden A and Bartlett R, Normalization of EMG amplitude: an evaluation and comparison of old and new methods. Med Eng Phys, 1999. 21(4):p.247-257.

Campanini I, Merlo A, Degola P et al., Effect of electrode location on EMG signal envelope in leg muscles during gait. J Electromyogr Kines, 2007. 17(4):p.515-526.

Caty V, Aujouannet Y, Hintzy F et al., Wrist stabilisation and forearm muscle coactivation during freestyle swimming. J Electromyogr Kines, 2007. 17(3):p.285-291.

Chapman AR, Vicenzino B, Hodges PW et al., A protocol for measuring the direct effect of cycling on neuromuscular control of running in triathletes. J Sport Sci, 2009. 27(7):p.767-782.

Crenna P, Spasticity and 'spastic' gait in children with cerebral palsy. Neurosci Biobehav R, 1998. 22(4):p.571-578.

Cresswell AG, Loscher WN and Thorstensson A, Influence of gastrocnemius muscle length on triceps surae torque development and electromyographic activity in man. Exp Brain Res, 1995. 105(2):p.283-290.

Earl JE, Schmitz RJ and Arnold BL, Activation of the VMO and VL during dynamic mini-squat exercises with and without isometric hip adduction. J Electromyogr Kines, 2001. 11(6):p.381-386.

Ebben WP, Hamstring Activation During Lower Body Resistance Training Exercises. IJSPP, 2009. 4(1):p.84-96.

Farina D, Interpretation of the surface electromyogram in dynamic contractions. Exercise Sport Sci R, 2006. 34(3):p.121-127.

Farina D, Madeleine P, Graven-Nielsen T et al., Standardising surface electromyogram recordings for assessment of activity and fatigue in the human upper trapezius muscle. Eur J Appl Physiol, 2002. 86(6):p.469-478.

Farina D, Merletti R, Nazzaro M et al., Effect of joint angle on EMG variables in leg and thigh muscles. IEEE Eng Med Biol, 2001. 20(6):p.62-71.

Fauth ML, Petushek EJ, Feldmann CR et al., Reliability of surface electromyography during maximal voluntary isometric contractions, jump landings, and cutting. J Strength Cond Res, 2010. 24(4):p.1131-1137.

Fleiss J. The Design and Analysis of Clinical Experiments. New York: Wiley; 1986.

Gasser HS, Contractures of skeletal muscle. Physiol Rev, 1930. 10(1):p.35-109.

Gazendam MGJ and Hof AL, Averaged EMG profiles in jogging and running at different speeds. Gait Posture, 2007. 25(4):p.604-614.

Gollhofer A, Horstmann GA, Schmidtbleicher D et al., Reproducibility of electromyographic patterns in stretch-shortening type contractions. Eur J of App Physiol O, 1990. 60(1):p.7-14.

Goodwin PC, Koorts K, Mack R et al., Reliability of leg muscle electromyography in vertical jumping. Eur J of App Physiol O, 1999. 79(4):p.374-378.

Guo JY, Zheng YP, Xie HB et al., Continuous monitoring of electromyography (EMG), mechanomyography (MMG), sonomyography (SMG) and torque output during ramp and step isometric contractions. Med Eng Phys, 2010. 32(9):p.1032-1042.

Hallett M, Shahani BT and Young RR, Emg Analysis of Patients with Cerebellar Deficits. J Neurol, 1975. 38(12):p.1163-1169.

Herbert RD, Moseley AM, Butler JE et al., Change in length of relaxed muscle fascicles and tendons with knee and ankle movement in humans. J Physiol-London, 2002. 539(2):p.637-645.

Hill AV, The heat of shortening and the dynamic constants of muscle. Proceedings of the Royal Society of London Series B-Biological Sciences, 1938. 126(843):p.136-195.

Hortobagyi T, Westerkamp L, Beam S et al., Altered hamstring-quadriceps muscle balance in patients with knee osteoarthritis. Clinical Biomechanics, 2005. 20(1):p.97-104.

Hubley-Kozey CL, Deluzio KJ, Landry SC et al., Neuromuscular alterations during walking in persons with moderate knee osteoarthritis. J Electromyogr Kines, 2006. 16(4):p.365-378.

Karamanidis K, Arampatzis A and Bruggemann GP, Reproducibility of electromyography and ground reaction force during various running techniques. Gait Posture, 2004. 19(2):p.115-123.

Kellis E and Baltzopoulos V, Gravitational moment correction in isokinetic dynamometry using anthropometric data. Med Sci Sport Exer, 1996. 28(7):p.900-907.

Knutson LM, Soderberg GL, Ballantyne BT et al., A Study of Various Normalization Procedures for within Day Electromyographic Data. J Electromyogr Kines, 1994. 4(1):p.47-59.

Larsson B, Karlsson S, Eriksson M et al., Test-retest reliability of EMG and peak torque during repetitive maximum concentric knee extensions. J Electromyogr Kines, 2003. 13(3):p.281-287.

Lin CF, Chen CY and Lin CW, Dynamic Ankle Control in Athletes With Ankle Instability During Sports Maneuvers. Am J Sport Med, 2011. 39(9):p.2007-2015.

Lin YC, Walter JP, Banks SA et al., Simultaneous prediction of muscle and contact forces in the knee during gait. J Biomech, 2010. 43(5):p.945-952.

Lind AR and Petrofsky JS, Amplitude of the Surface Electromyogram during Fatiguing Isometric Contractions. Muscle Nerve, 1979. 2(4):p.257-264.

Lloyd AJ, Surface Electromyography during Sustained Isometric Contractions. J App Physiol, 1971. 30(5):p.713-&.

Lloyd DG, Rationale for training programs to reduce anterior cruciate ligament injuries in Australian football. J Orthop Sport Phys, 2001. 31(11):p.645-654.

Lloyd DG, Buchanan TS and Besier TF, Neuromuscular biomechanical modeling to understand knee ligament loading. Med Sci Sport Exer, 2005. 37(11):p.1939-1947.

Maffiuletti NA and Lepers R, Quadriceps femoris torque and EMG activity in seated versus supine position. Med Sci Sports Exer, 2003. 35(9):p.1511-1516.

Maganaris CN, Baltzopoulos V and Sargeant AJ, In vivo measurements of the triceps surae complex architecture in man: implications for muscle function. J Physiol-London, 1998. 512(2):p.603-614.

Mann RA and Hagy J, Biomechanics of Walking, Running, and Sprinting. Am J Sport Med, 1980. 8(5):p.345-350.

Maupas E, Paysant J, Datie AM et al., Functional asymmetries of the lower limbs. A comparison between clinical assessment of laterality, isokinetic evaluation and electrogoniometric monitoring of knees during walking. Gait Posture, 2002. 16(3):p.304-312.

Mendez-Villanueva A, Baudry S, Riley ZA et al., Influence of rest duration on muscle activation during submaximal intermittent contractions with the elbow flexor muscles. J Sport Med Phys Fit, 2009. 49(3):p.255-264.

Mirka GA, The Quantification of EMG Normalization Error. Ergonomics, 1991. 34(3):p.343-352.

Mottram CJ, Jakobi JM, Semmler JG et al., Motor-unit activity differs with load type during a fatiguing contraction (vol 93, pg 1381, 2005). J Neurophysiol, 2005. 94(1):p.901-901.

Murley GS, Menz HB, Landorf KB et al., Reliability of lower limb electromyography during overground walking: A comparison of maximal- and sub-maximal normalization techniques. J Biomech, 2010. 43(4):p.749-756.

Shiavi R, Frigo C and Pedotti A, Electromyographic signals during gait: criteria for envelope filtering and number of strides. Med Biol Eng Comput, 1998. 36(2):p.171-178.

Soderberg GL and Cook TM, Electromyography in Biomechanics. Phys Ther, 1984. 64(12):p.1813-1820.

Stephens JA and Taylor A, Fatigue of Maintained Voluntary Muscle-Contraction in Man. J Physiol-London, 1972. 220(1):p.1-&.

Thomas CK, Johansson RS and Bigland-Ritchie B, EMG changes in human thenar motor units with force potentiation and fatigue. J Neurophysiol, 2006. 95(3):p.1518-1526.

Winby CR, Lloyd DG, Besier TF et al., Muscle and external load contribution to knee joint contact loads during normal gait. J Biomech, 2009. 42(14):p.2294-2300.

Winter JM, Woo SL, Multiple muscle systems: Biomechanics and movement organization. New York: Springer-Verlaag; 1990. 717-773 pp.

Yoon SH and Challis JH, The variability of maximum vertical jumps. J Hum Movement Stud, 2005. 48(2):p.147-156.

Zeni JA, Rudolph K and Higginson JS, Alterations in quadriceps and hamstrings coordination in persons with medial compartment knee osteoarthritis. J Electromyogr Kines, 2010. 20(1):p.148-154.

APPENDIX

Appendix A

MODEL PARAMETERS

Table 6.1. The physiological limits within the optimization criteria of the EMG drive, muscle force estimation model.

	Parameter	Description	Limits
ℓ_o^m	optimal fiber length	The length at which the sarcomeres are aligned for maximal force	±20%
ℓ_s^t	resting tendon length	The amount of tendon stretched before tension acts as a passive force	±20%
D	electromechanical delay	Time from EMG signal arrival to muscle contraction	10 - 100ms
A_{len}	non-linear shape factor	A relationship to account for the non-linearity of EMG to force relationship	0.01 - 0.12
γ_1, γ_2	recursive filter coefficients	Coefficients to calculate activation from EMG with 2nd order dynamics	$-0.9 < \gamma_{1,2} < 0.9$
G_e	extensor strength gain	Strength range of extensors based on isometric values	0.5 – 1.5
G_f	flexor strength gain	Strength range of flexors based on isometric values	

Appendix B

SUBJECT INFORMED CONSTENT

UNIVERSITY OF DELAWARE

RESEARCH OFFICE

210 Hullihen Hall
University of Delaware
Newark, Delaware 19716-1551
Ph: 302/831-2136
Fax: 302/831-2828

DATE:	November 9, 2010
TO:	Stephen Suydam, BSME
FROM:	University of Delaware IRB
STUDY TITLE:	[187870-1] Consistency of peak dynamic EMG in leg muscles
SUBMISSION TYPE:	New Project
ACTION:	APPROVED (E. Peloso & C. Galloway)
APPROVAL DATE:	November 9, 2010
EXPIRATION DATE:	November 8, 2011
REVIEW TYPE:	Expedited Review

Thank you for your submission of New Project materials for this research study. The University of Delaware IRB has APPROVED your submission. This approval is based on an appropriate risk/benefit ratio and a study design wherein the risks have been minimized. All research must be conducted in accordance with this approved submission.

This submission has received Expedited Review based on the applicable federal regulation.

Please remember that informed consent is a process beginning with a description of the study and insurance of participant understanding followed by a signed consent form. Informed consent must continue throughout the study via a dialogue between the researcher and research participant. Federal regulations require each participant receive a copy of the signed consent document.

Please note that any revision to previously approved materials must be approved by this office prior to initiation. Please use the appropriate revision forms for this procedure.

All SERIOUS and UNEXPECTED adverse events must be reported to this office. Please use the appropriate adverse event forms for this procedure. All sponsor reporting requirements should also be followed.

Please report all NON-COMPLIANCE issues or COMPLAINTS regarding this study to this office.

Please note that all research records must be retained for a minimum of three years.

Based on the risks, this project requires Continuing Review by this office on an annual basis. Please use the appropriate renewal forms for this procedure.

- 1 -

Generated on IRBNet

If you have any questions, please contact Clara Simpers at 302-831-2137 or csimpers@udel.edu. Please include your study title and reference number in all correspondence with this office.

UNIVERSITY OF DELAWARE
POSTURE AND MOVEMENT BIOMECHANICS LABORATORY
INFORMED CONSENT FORM

Study Title: Consistency of peak dynamic EMG in leg muscles.

Principal Investigators: Thomas S. Buchanan PhD, Stephen Suydam, BS

PURPOSE AND BACKGROUND

You are being asked, along with approximately twenty other subjects, to participate in a study that will evaluate leg muscle activity (EMG) recorded using surface electrodes, during several different tasks. Muscle activity will be recorded during static (non-moving) and dynamic (moving) tests. The following tasks will be examined: walking, running at a fixed speed, running at a fixed speed and cutting diagonally to one side, running as fast as you can, and jumping as high as you can.

Participation in this research study is entirely voluntary and refusal to participate will involve no penalty. You may withdraw from the study at any time. You must be at least 18 years of age to participate in this study.

All testing will take place in the Posture and Movement Biomechanics Laboratory at the University of Delaware. The entire testing session will take approximately 2.5 hours to complete. This is a test-retest experiment and you will be required to return one time within one month following the first testing session. This second testing session will take an additional 2.5 hours to complete.

PROCEDURES

Recording Muscle Activity

Surface electrodes will be placed on the skin of your lower leg and thigh. In addition, reflective markers will be placed on easily identifiable points of reference on your body. These reflectors make it easy for the video cameras to collect your motion with a large degree of accuracy.

All electrodes will be taped to your skin to prevent movement and then plugged into a small (6” x 4” x 3”) transmitter box that will be attached to the back of a vest, which you will be wearing, with Velcro. Muscle activity will be collected so we can determine when the muscles are working during the activities. These measurements will be taken at the same time as the dynamic testing and the motion analysis testing.

Strength Testing

Strength will be measured of your upper and lower legs while you are seated in a device that limits your motion and measures how much force you can produce. You will have your leg placed in a fixed position and you will perform a push and pull movement as hard as you can with both your lower leg and foot individually. Following that, your movement will be restricted to a speed of 60 degrees per second and you will again push and pull as hard as you can with both your lower leg and foot with verbal encouragement from the researcher. Each maximum force trial will be repeated three times. The test will measure the strength of your lower leg and thigh muscles and you will be provided with 2 minutes rest between each trial.

Walk, Run, Sprint and Jump Testing

Markers will be placed on your skin and shoes (both legs) using adhesive skin tape. Shells with markers will be placed on your lower back, thigh, calf, and foot and will be held in place with elastic wraps. Video cameras will be used to track the markers on your leg during the activities.

You will first perform walking trials at your self selected speed. The next test will be a running test at 9 miles per hour. After that test, a 45^{o} angle step will added to the middle of the run. Following the run/cut trial, you will perform a sprint in which you will accelerate to your maximum speed for a distance of 15 meters. During each of these trials you will step on a force plate to calculate the force of your landing. Lastly, you will perform a counter movement jump from the force plate. This involves swinging your arms, squatting down and then jumping as high as you can. Seven trials will be performed for each task. These trials will give us information about the way your legs move during these activities.

Prior to the first collection of data for each type of movement, you will be allowed to perform several practice trials with stand-by assistance to allow you to get comfortable with the activity and to make sure that you can maintain your speed and consistency of each of the tasks. You will be given time to rest between each trial so you do not feel fatigued.

Risks/Discomforts

The risks associated with participation in this study are minimal. You may experience discomfort from the removal of tape holding markers and EMG electrodes in place. If you have an allergy to any adhesives, please let the researcher know. This may exclude you from the trial. There will be wires and cameras within the running field, but a specific path as been laid out to prevent injury and damage as well as having a researcher watching you and the wires to prevent tripping. The strength, sprint and jump testing can cause local muscle soreness and fatigue. Following testing your muscles may feel as if you exercised vigorously.

Benefits

The results of this study may allow for clinicians to use the muscle signals from moving trials as a tool for identifying muscles used during movement tasks in future research studies; however there will be no direct benefit to the current participants.

Compensation

You will receive no compensation for participation in this research study

Confidentiality and Records

All subjects will be identified by number. Only the investigators will have access to the data. Neither your name nor any identifying information will be used in publication or presentation resulting from this study. A statistical report which may include pictures that do not identify you may be disclosed in a scientific paper. Data will be archived indefinitely. There will be no identifying information of you stored with the data.

Study Title: Reliability of Peak Dynamic EMG of the Leg for Test-Retest

Principal Investigators:

Thomas S. Buchanan, PhD
Stephen Suydam, BS

Subjects Statement:

I have read this consent form and have discussed the procedure described above with the principal investigator. I have been given the opportunity to ask questions regarding this study, and they have been answered to my satisfaction.

In the case that I am injured or experience an acute medical emergency during the study, I will be provided with immediate first aid. Any additional care will be at my own expense. I have been fully informed of the above described procedures, with its possible risks and benefits, and I hereby consent to the procedures above.

______________________________ ________________________________

Subject's Name Subject's Signature Date

____

Witness Date

If you have any questions concerning the rights of individuals who agree to participate in research, you may contact the Chair of University of Delaware IRB, at (302) 831-2137.

Further questions regarding this study may be addressed to:
Thomas S. Buchanan, PhD
(302) 831-2401
Posture and Movement Biomechanics Laboratory
Department of Mechanical Engineering
Center for Biomedical Engineering Research