THE SURFACE ELECTROMYOGRAPHY-FORCE RELATIONSHIP DURING ISOMETRIC DORSIFLEXION IN MALES AND FEMALES

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ABSTRACT

This study evaluated sex-related differences in the tibialis anterior (TA) surface electromyography (EMG) to force relationship. One-hundred participants (50 males and 50 females) performed three isometric contractions at 20, 40, 60, 80, and 100% of maximal voluntary contraction (MVC) in an apparatus designed to isolate the action of the dorsiflexors. The surface EMG signal was amplified (1000×), band-pass filtered (10–500Hz), and sampled at 2048 Hz. The load cell signal was low-passed filtered at 100 Hz and sampled at the same rate. Males were stronger than females ($P < 0.05$). However, there was no significant difference in root-mean-square (RMS) values between sexes ($P < 0.05$). Both sexes exhibited a quadratic increase in RMS across force levels ($P < 0.05$). The mean power frequency (MNF) for males was greater than for females ($P < 0.05$). Males and females exhibited a linear increase in both frequency measures up to 80% of MVC ($P < 0.05$). Between 80 and 100% MVC, the frequency values for the females plateaued while males showed a decrease ($P < 0.05$). The magnitude of the difference in MNF between males and females was consistent with sex-specific TA physiology. In general, the pattern of means for RMS and MNF between males and females revealed no sex-related differences in the surface EMG/force relationship. We therefore conclude that there are no sex-related differences in the gradation of muscle force.
ACKNOWLEDGEMENTS

I would like to express my deep gratitude to Dr. Gabriel for his pragmatic wisdom that filled all parts of my life during my studies. There were many times his unerring perception was able to boil things down so simply. David is a true master craftsman of his trade and knows that all good things come from rolling up your sleeves and having at it. For every Batman, there is a Robin and for David, there is Greig. I am pleased to have met Greig not only for his humour and wit, but also for all the help with technical issues and presentations. I wish Greig the best of luck on his future endeavours. To all the hands that have helped make the work lighter, namely Jenn, Ed, Tim, and Kyle. I am grateful for all the time and effort you put in, none of this could be possible without your teamwork. To my special lady friend Rose, you help keep me on an even keel everyday making anything possible. Your support is the rock of my life.
Table of Contents

CHAPTER I DEVELOPMENT OF THE PROBLEM .............................. 1
   Introduction ........................................................................ 1
   Statement of Purpose ...................................................... 3
   Hypotheses ......................................................................... 3
   Significance of the Study .................................................... 4
   Basic Assumptions ............................................................ 5
   Glossary ........................................................................... 5

CHAPTER II REVIEW OF THE LITERATURE .............................. 7
   Physiological Considerations for the tibialis anterior .............. 7
      Muscle-tendon architecture ............................................. 7
      Motor units ...................................................................... 10
      Origin of the surface electromyography pattern ................ 11
   Measures of Surface Electromyographic Activity .................. 16
      Root-mean-square amplitude ......................................... 16
      Mean power frequency ................................................... 22
   Sex Differences .................................................................. 25
      Muscle Strength ............................................................ 25
      Electromyography .......................................................... 26

CHAPTER III METHODS .......................................................... 30
   Experimental Design ........................................................ 30
   Recording surface electromyographic activity and force ........ 32
   Data reduction and analysis .............................................. 35

CHAPTER IV RESULTS ........................................................... 37
   Pre and Post Contractions .................................................. 37
   * denotes significance set at the 0.05 level .......................... 38
   Force ............................................................................... 38
   Root-mean-square ........................................................... 39
   Mean power frequency ..................................................... 40

CHAPTER V DISCUSSION ......................................................... 42
   Force ............................................................................... 42
   Electromyography ............................................................ 43
   Summary - Conclusions ..................................................... 48

References ............................................................................ 50

Appendix A ........................................................................... 61
   Subjects ............................................................................ 61

Appendix B ........................................................................... 63
   Physical Activity Questionnaire ......................................... 63
CHAPTER I

DEVELOPMENT OF THE PROBLEM

Introduction

The relationship between surface electromyography and force has been used to study how the nervous system regulates the generation of force in skeletal muscle. The amplitude of the surface electromyographic signal is a function of motor unit recruitment and firing rate statistics (Lawrence & DeLuca, 1983). As more motor units are recruited and/or these motor units increase in firing rate the amplitude or magnitude of the electromyographic signal increases. The frequency content of the surface electromyographic signal increases with the recruitment of higher threshold motor units, which are associated with larger diameter muscle fibers that have faster conduction velocities (Broman et al., 1985; Kupa et al., 1995). Motor unit firing rate statistics also affects the frequency content of signal, but to a lesser degree (DeLuca, 1997). The interaction between recruitment range and firing rate accounts for differences in the relationship between force and surface electromyography between large and small muscle groups (Lawrence & DeLuca, 1983). However, differences between studies for the same muscle group are usually attributable to the physical characteristics electrode detection system used (Farina et al., 2004).

Direct measurement has demonstrated that the proportion of type I fibers positively correlates with the mean frequency, whereas the proportion of type II fibers positively correlates with the root-mean-square surface electromyographic amplitude (Gerdle et al., 2000). Since males and females are nearly identical with respect to the
fiber type ratio for the tibialis anterior (Holmstäck et al., 2003), it might be expected that they should exhibit the same pattern of change in the root-mean-square and mean power frequency measures of the surface electromyographic signal with increasing relative force levels. Cioni et al. (1994) provide data that partially support this hypothesis. Monopolar detected root-mean-square magnitude for the tibialis anterior was nearly identical for males and females from 10 – 60% of maximal voluntary contraction; males then diverged to achieve higher values until 100% of maximal voluntary contraction. The same pattern of means was observed for bipolar detected root-mean-square magnitude, but the separation in means occurred after 30% of maximal voluntary contraction. The median power frequency data were mixed. Although males had a higher median frequency than females, sex differences (linear versus nonlinear) were also dependent on electrode configuration.

The data by Cioni et al. (1994) further highlights the fact that technical factors related to the signal detection system affect the surface electromyographic-force relationship. Cioni et al. (1994) used an inter-electrode distance of 3 cm, which is quite large relative to the distribution of motor points in the tibialis anterior (Roy et al., 1986). Electrode placement relative to the motor points is known to have an impact on the magnitude of the surface electromyographic amplitude and frequency estimates (Roy et al., 1986). When a signal detection system is placed on either side of a motor point (neuromuscular junction), the result is a reduction in the magnitude of the myoelectric signal due to waveform cancellation. Furthermore, as a signal detection system moves further from a motor point, there is also a reduction in the magnitude of the myoelectric signal due to a muscles’ inability to maintain an action potential as well as a neuron. A
larger inter-electrode distance also increases the risk of cross-talk contamination from adjacent muscles (DeLuca and Merletti, 1988; Winter et al., 1994).

**Statement of Purpose**

The purpose of this thesis was to re-examine the surface electromyographic-force relationship in the amplitude and frequency domains using a bipolar configuration with a selective inter-electrode distance, placed away from electrically identified motor points. It is assumed that sex differences in recruitment and firing strategies exist; they will manifest themselves in the shape of the statistically defined surface electromyographic-force relationship (i.e., linear, quadratic, or cubic). Although the males and females are similar with respect to tibialis anterior muscle morphology, there is still a distinct difference in the physiological cross-sectional area that will impact the absolute magnitude of surface electromyographic activity. Thus, it is hypothesized that males and females will exhibit the same surface electromyographic-force relationship differing only in absolute magnitude.

**Hypotheses**

The following experimental hypotheses will be tested in this study:

1. There are no significant difference between sexes with respect to dorsiflexion strength,
2. There are no significant difference between sexes in the root-mean-square amplitude of the surface electromyographic-force relationship,
3. There are no significant differences between sexes in the mean power frequency of the surface electromyographic-force relationship.
Significance of the Study

Males and females are similar with respect to tibialis anterior muscle morphology. Further, when specific strength (strength/physiological cross-sectional area) is calculated for maximum dorsiflexion strength, there are no significant differences between males and females (Holmbäck et al., 2003). Since the motor apparatus involved in the generation of maximum dorsiflexion strength appears to be scaled between males and females, there is no reason to expect differences in the gradation of muscle force. However, this basic physiological information is only partially supported. The gradation of force is important because it dictates how the muscle functions during submaximal activities. The proposed research will provide data that will contribute to the understanding of potential differences (or lack thereof) between males and females with respect to the gradation of muscle force.

The proposed study also has practical applications. The tibialis anterior is critical for control of the foot during gait and for balance control (Winter, 1991; Winter, 1995). If there is a decrement in tibialis anterior activation due to stroke or neuromuscular disorder, gait and balance control can be nearly impossible, depending on the severity of the disorder specific to just this one muscle (Malanga & DeLisa, 1998). The efficacy of restorative therapies and neuroprostheses depends on knowledge of the relationship between muscle activation as revealed by surface electromyography and force (Amankwah et al., 2006; Guiraud et al., 2006).
Basic Assumptions

The following assumptions affect the interpretation of the results:

1. The interference pattern recorded at the skin surface represents the activity of motor units within the recorded volume of tissue,

2. Dorsiflexion force is dominated by the tibialis anterior without substantial contribution of the toe extensor muscles (extensor digitorum longus and extensor hallucis longus),

3. The maximal voluntary contraction generated by participants in the study is truly representative of their maximum effort,

Glossary

The candidate realizes that there may be variations to the definitions of the following terms; however, the definitions given are those relative to the proposed study.

Isometric contraction. Excitation and contraction coupling of actin and myosin resulting in force transmission to the bone through connective tissues. The muscle shortens during an isometric contraction, but there is no overt movement of the limb.

Mean power frequency. The frequency corresponding to the mean power of the signal is the mean power frequency.

Median frequency. The frequency associated with the power that divides the total power of the signal into lower and upper halves.

Motor unit. A single alpha motor neuron and all the muscle fibers that it innervates.

Motor unit recruitment. The recruitment of additional motor units progressing in an orderly fashion from low threshold to high threshold motor units.
Muscle fatigue. Muscular fatigue is defined as a reversible state of a muscle characterized by a temporary decrease in maximum isometric strength associated with the intensity and duration of the activity being performed.

Ramp contraction. A ramp contraction is defined as a single on-going contraction with linearly increasing force output from the muscle, usually from 0 to 100% of maximal voluntary effort.

Rate coding. Change the impulse frequency of motor unit firing to meet the demands of the task.

Root-mean-square. Each datum of the surface electromyographical signal is first squared. The sum of the squared data taken over a specific number of data points (N) is then divided by N to obtain a mean square. The square root detector is then applied to bring the magnitude of the signal back to its original scale.

Step contraction. A step contraction requires that the participant exert a force from a relaxed state to a target force level, defined as a percent level of maximal voluntary effort. The target must be achieved as quickly and as accurately as possible so that the force profile resembles a “step”.

Surface electromyography. The summations of motor unit action potentials both in number and frequency generate an interference pattern that is volume conducted through the intervening tissue and detected at the skin surface.
CHAPTER II

REVIEW OF THE LITERATURE

Physiological Considerations for the tibialis anterior

Muscle-tendon architecture

The tibialis anterior originates on the lateral surface of the tibia and the portion of the interosseous membrane spanning the tibia and fibula (see Figure 1). The distal portion of the tibialis anterior runs along medial and plantar surface of the medial cuneiform bone and inserts at the base of the first metatarsal (Rasch & Burke, 1978). Although the muscle lies along the length of the anterior surface of the tibia, the lower section of the muscle becomes tendonous (Luttgens & Wells, 1982). The muscle fibers appear to run longitudinally, but insert obliquely into the tendon, which is held down by the superior extensor retinaculum (Rasch & Burke, 1978). The tibialis anterior is a bi-pennate muscle (Maganaris et al., 2001). Although pennate muscles do not generate as much force per fiber about a joint, the fibers are usually packed much more tightly than fusiform muscles, and therefore have more fibers per area unit. As a pennate muscle contracts in an isometric fashion, it is important to understand how the area of that muscle is affected.

Maganaris and Baltzopoulos (1999a) assessed the predictability of ultrasound based changes in the tibialis anterior pennation angle from rest to a 100% isometric dorsiflexion at maximal voluntary contraction using a planimetric model, assuming constant thickness between the aponeuroses and straight muscle fibers.
Dorsiflexion contractions were performed on an isokinetic dynamometer with ankle angles at -15 degrees in the plantarflexion direction, 0, +15, and +30 degrees. When comparing the maximal voluntary contractions to resting muscle length at any given angle, pennation angle was larger (62 – 71%, P<0.01), fiber length smaller (37 – 40%, P<0.01), and muscle thickness was unchanged (P<0.05). The tibialis anterior architecture was also identical between the two unipennate parts and along the scanned length at a given ankle angle and state of contraction.

Maganaris et al. (1999b) tested the hypothesis that the tibialis anterior moment arm increases during maximum isometric dorsiflexion, as compared to the muscle in a resting state. Many studies assume that the moment arm does not change or is unaffected by contractions of muscles enclosed by retinacular systems, such as the tibialis anterior. Maganaris and colleagues (1999b) used the same experimental paradigm as the previous study. They observed the tibialis anterior tendon moment arm during a maximal voluntary contraction of isometric dorsiflexion increased by 0.9 – 1.5 cm, compared with its length in the resting state. This change in moment arm was primarily attributed to a displacement of both the tibialis anterior action line by 0.8 – 1.2 cm and the instantaneous center of rotation, which moved 0.3 – 0.4 cm distally in relation to its original starting position. As the tibialis anterior contracts near maximum force, an immediate observation is that the talus bone inserts into a notch between the tibia and fibula, which changes the center of rotation. Unlike the gastrocnemius and soleus, the tibialis anterior muscle thickness does not change in the transition from rest to maximal voluntary contraction (Maganaris et al., 1999b). The result is a stretch of the retinaculum during
static dorsiflexion, which displaces the tibialis anterior tendon compared to its resting state.

Marsh and colleagues (1981) examined the influence of joint position on ankle dorsiflexion in humans. Stimulation of the tibialis anterior at 30 to 40 Hz showed that the optimum length of the muscle corresponded to approximately 10 degrees of plantarflexion. Maximum torque was developed at 10 degrees of plantarflexion, and it was severely compromised when the ankle was dorsiflexed beyond 5 degrees. However, the departure from maximal force was not significant between 10 degrees of plantarflexion and the neutral ankle joint angle. An important point relevant to the proposed study is that motor-neuron excitability did not appear to be influenced significantly by changes in the joint angle. Therefore, the electromyographic amplitude and frequency measures are most likely unaffected.

**Motor units**

Feiereisen et al. (1997) showed that 90% of new motor units in the tibialis anterior are recruited by 60% maximal voluntary contraction and that all motor units have been recruited by 88% of maximal voluntary contraction. Therefore, motor unit firing rates should have a stronger influence on the surface electromyographic-force relationship from 60% of maximal voluntary contraction onward. However, if the tibialis anterior follows the motor unit recruitment pattern similar to that of other muscles, then each new motor unit recruited would have an increasing population of muscle fibers that are innervated (Keenan et al., 2005).
Henriksson-Larsén et al. (1983) examined whether small biopsy samples are representative of the whole human muscle or whether the different fiber types are unevenly distributed at different depths of the muscle. The researchers examined cross-sections of whole tibialis anterior muscles. The total number of fibers varied greatly between individuals, from 96,000 to 162,000 in five individuals. There was a gradual increase in type II fiber occurrence from the surface of the muscle, about 10 – 25%, to the deeper regions of the muscle, about 30 – 50%. Concurrently type I fibers are predominately located in the anterior portion of the muscle (75-90%). Therefore, due to the superficial location of type I fibers, they will represent the greatest contribution of new motor units recruited and firing rates measured from surface electrodes.

**Origin of the surface electromyography pattern**

A motor unit is defined as a single alpha-motor-neuron and all of the muscle fibers that it innervates (Basmajian & DeLuca, 1985). Two motor units and their associated muscle fibers, which are intermingled within the whole muscle, are illustrated in Figure 2. There are three muscle fibers for motor unit ‘A’ and two muscle fibers for motor unit ‘B’. Each muscle fiber generates its own muscle fiber action potential. Innervation of a given motor unit then results in summation of its associated muscle fiber action potentials to produce a motor unit action potential. Repetitive motor units firing are defined as a motor unit action potential train. The basic building block of the surface electromyographic interference pattern is the summation of two or more motor unit action potential trains (Figure 3). It is important to recognize that the potentials travel bi-directionally from the innervation zone, and can be detected a great distance away by surface electrodes through fascia, fat, and skin (Figure 4). Fascia, fat, and skin all
combine to filter the electromyographic signal that is volume conducted from the muscle fiber. The result is a decrease of power in the higher frequency components compared to electrodes that are inserted directly into the muscle.
Figure 2. Surface electromyographic recordings from muscle fibers of two motor units.

Figure 3. A graphical representation of an electromyographic signal by adding generated motor unit action potential trains. Basmajian, J.V., and De Luca, C.J. (1985). *Muscles Alive: Their function revealed by electromyography* (5th Ed.). Baltimore, MD: Williams & Wilkins, Figure 3.9, page 81.
Figure 4. Detection of the electromyographic signal at the skin surface. Basmajian, J.V., and De Luca, C.J. (1985). *Muscles Alive: Their function revealed by electromyography* (5th Ed.). Baltimore, MD: Williams & Wilkins, Figure 2.17, page 54.
Measures of Surface Electromyographic Activity

Root-mean-square amplitude

To calculate the root-mean-square amplitude of the surface electromyographic signal, each datum (x) within a specific time-period (T) is first squared. The sum of all the squared values is then divided by the number of data points that make the sum, resulting in a mean-square. The square-root is performed to return the magnitude of the mean-square value back to the original scale of measurement:

\[ RMS = \sqrt{\frac{1}{T} \int_{t}^{t+T} x^2(t)dt} \]

The root-mean-square amplitude is mathematically related to the half-power of the surface electromyographic signal, and it is considered a better measure than the average rectified value for monitoring changes in muscle activity because it is not affected by cancellation of the motor unit action potentials (De Luca & Van Dyk, 1975). It is important to emphasize that specific motor unit behavior cannot be inferred from amplitude changes alone.

The root-mean-square amplitude of the surface electromyographic signal is indicative of the amount of neural drive to the muscle. The type of muscular contraction (i.e., isometric, isotonic, or isokinetic) has an effect on the nature of the relationship between force and root-mean-square amplitude of the surface electromyographic signal. Isometric contractions are most often used in investigative work as a methodological control to minimize changes in muscle force due the stretch-shortening cycle and associated variations in moment arm length (Kamen & Caldwell, 1996). The validity of
surface electromyographic analysis depends on the stationarity of the signal. Stationarity means that the mean and standard deviation of the signal remains constant. Any measure calculated on a stationary signal is thought to be representative of the time-period under consideration. If muscle force remains stable during the time-period of analysis, the magnitude of the surface electromyographic signal is assumed to be stationary (Bilodeau et al, 1997).

Robertson et al. (2004) stated that an increase in the root-mean-square amplitude is caused by an increase in active motor units firing and the frequency of activation of those motor units. The amplitude of the electromyographic signal has been shown to be related to the size of the force tension exerted, where a linear relationship has been observed (Beck et al., 2005; Metral & Cassar, 1981; Moritani & de Vries, 1978; Philipson & Larsson, 1988). Philipson & Larsson (1988) described the root-mean-square as the most reliable descriptor of force from 0 – 100% of maximal voluntary contraction. These researchers also strongly supported the use of monopolar surface electrodes when attempting to provide the most valid detection of the electromyography amplitude. While needles showed a greater variation around the mean, bipolar electrodes yield spatially filtered action potentials because they are differentially amplified (Philipson & Larsson, 1988). Fatigue can also affect the observed linear relationship between force and electromyography. As fatigue becomes prominent in the muscle of interest, it has been observed that the root-mean-square amplitude will increase (Merletti et al., 1990). A quadratic relationship could be erroneously observed as a linear one or a real linear plot may be shifted to one showing a positively accelerated curvilinear relationship (Moritani & de Vries, 1978).
Different electrode configurations have been used to study the surface electromyography-force relationship. Using a four-bar electrode array which resulted in a double-differential signal, Sbricolli et al. (2003) observed a curvilinear relationship in the biceps brachii surface electromyographic amplitude during ramp contractions. Using a Laplacian electrode configuration, Ollivier et al. (2005) also observed a curvilinear relationship. Dupont et al. (2000) and Gabriel and Kamen (In press) observed a linear relationship for both ramp and step isometric contractions, but used bipolar electrodes. It is difficult to know whether or not the disparate findings are due to electrode configuration or other technical factors. For example, when Cioni et al. (1994) compared unipolar and bipolar electrode arrangements on the electromyography-force relationship in the tibialis anterior, they observed curvilinear relationships in both configurations although the absolute values were different; however, when the results were normalized, they were nearly identical.

A historical summary of results concerning the surface electromyography-force relationship are summarized in Table 1. Not only are there differences in the electrode detection system, but the actual measure used to quantify surface electromyography activity. Integrated electromyographic activity has been used by many of the authors listed in Table 1. This measure is not often defined the same way by different investigators, and it is seldom true mathematical integration.
<table>
<thead>
<tr>
<th>Author</th>
<th>Year</th>
<th>Muscle Reviewed</th>
<th>Contraction</th>
<th>Relationship</th>
<th>Measures of EMG</th>
<th>Electrode Configuration</th>
</tr>
</thead>
<tbody>
<tr>
<td>Inman et al.</td>
<td>1952</td>
<td>Biceps, Pectoralis, Triceps, Tibialis, Anterior</td>
<td>Isometric</td>
<td>Linear</td>
<td>Integrated electromyography</td>
<td>Surface bipolar, needle, and wire</td>
</tr>
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<td>Lippold</td>
<td>1952</td>
<td>Soleus, Gastrocnemius</td>
<td>Isometric</td>
<td>Linear</td>
<td>Integrated electromyography</td>
<td>Surface bipolar</td>
</tr>
<tr>
<td>Bigland &amp; Lippold</td>
<td>1954</td>
<td>Calf and finger extensor</td>
<td>Isotonic</td>
<td>Linear</td>
<td>Integrated electromyography</td>
<td>Surface bipolar</td>
</tr>
<tr>
<td>Bergstrom</td>
<td>1959</td>
<td>Forefinger Abductor</td>
<td>Non-isometric</td>
<td>Linear up spike freq 500 cps</td>
<td>Integrated electromyography</td>
<td>Surface bipolar</td>
</tr>
<tr>
<td>De Jong &amp; Freund</td>
<td>1967</td>
<td>Adductor Brevis, Pollicis (Evoked)</td>
<td>Isometric</td>
<td>Linear</td>
<td>Amplitude of the action potential</td>
<td>Needle electrodes, disc electrodes</td>
</tr>
<tr>
<td>Bouisset &amp; Goubel</td>
<td>1973</td>
<td>Biceps Brachii</td>
<td>Dynamic</td>
<td>Linear</td>
<td>Integrated electromyography</td>
<td>Surface bipolar</td>
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<td>Bigland-Ritchie</td>
<td>1974</td>
<td>Quadriceps</td>
<td>Isotonic</td>
<td>Linear</td>
<td>Integrated electromyography</td>
<td>Surface bipolar</td>
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<td>Milner-Brown &amp; Stein</td>
<td>1975</td>
<td>First Interosseus, Dorsal</td>
<td>Isometric</td>
<td>Linear</td>
<td>Mean rectified electromyography</td>
<td>Surface bipolar</td>
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<tr>
<td>Author</td>
<td>Year</td>
<td>Muscle Reviewed</td>
<td>Contraction</td>
<td>Relationship</td>
<td>Measures of EMG</td>
<td>Electrode Configuration</td>
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<tr>
<td>Fuglsang-Frederiksen &amp; Mansson</td>
<td>1975</td>
<td>Triceps Brachii, Biceps Brachii, Brachioradialis</td>
<td>Isometric</td>
<td>Linear up to 30-50% of maximal voluntary contraction</td>
<td>Number of turns and mean amplitude</td>
<td>Concentric electrode needle</td>
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<td>Komi &amp; Viitasalo</td>
<td>1976</td>
<td>Rectus Femoris</td>
<td>Isometric</td>
<td>Quadratic with increase of muscle force</td>
<td>Integrated electromyography</td>
<td>Surface bipolar</td>
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<tr>
<td>Moritani &amp; deVries</td>
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<td>Quadratic with increase of muscle force</td>
<td>Integrated electromyography</td>
<td>Surface bipolar</td>
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<td>1983</td>
<td>First Interosseus, Dorsal Interosseus, Deltoid, Biceps Brachii</td>
<td>Isometric Abduction (FDI, Deltoid), Isometric Flexion (Biceps)</td>
<td>Quasilinear (FDI), Non-linear (Deltoid, Biceps)</td>
<td>Root-mean-square</td>
<td>Surface bipolar</td>
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<td>Kranz et al</td>
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<td>Biceps Brachii</td>
<td>Isometric</td>
<td>Non linear (Root-mean-square); Linear (Corrected surface electromyographic signal)</td>
<td>Root-mean-square</td>
<td>Surface bipolar</td>
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<tr>
<td>Author</td>
<td>Year</td>
<td>Muscle Reviewed</td>
<td>Contraction</td>
<td>Relationship</td>
<td>Measures of EMG</td>
<td>Electrode Configuration</td>
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<tr>
<td>Cioni et al.</td>
<td>1994</td>
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<td>Surface bipolar</td>
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<td>Onishi et al.</td>
<td>2000</td>
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<td>Linear</td>
<td>Average Integrated electromyography</td>
<td>Wire bipolar</td>
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<td>Dupont et al.</td>
<td>2000</td>
<td>Biceps Brachii</td>
<td>Isometric</td>
<td>Linear</td>
<td>Root-mean-square</td>
<td>Surface bipolar</td>
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<td>Karlsson &amp; Gerdle</td>
<td>2001</td>
<td>Quadriceps (VL, RF, VM)</td>
<td>Isometric</td>
<td>Linear</td>
<td>Root-mean-square</td>
<td>Surface bipolar</td>
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<tr>
<td>Babault et al.</td>
<td>2001</td>
<td>Rectus Femoris</td>
<td>Isometric</td>
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<td>Root-mean-square</td>
<td>Surface bipolar</td>
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<td>Bilodeau et al.</td>
<td>2003</td>
<td>Quadriceps (VL, RF, VM)</td>
<td>Isometric ramp</td>
<td>Linear</td>
<td>Root-mean-square</td>
<td>Surface bipolar</td>
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<tr>
<td>Ferrario et al.</td>
<td>2004</td>
<td>Masseter and Temporalis Anterior</td>
<td>Isometric</td>
<td>Linear</td>
<td>Root-mean-square</td>
<td>Surface bipolar</td>
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<tr>
<td>Beck et al.</td>
<td>2005</td>
<td>Biceps Brachii</td>
<td>Isometric/Isokinetic</td>
<td>Linear</td>
<td>Root-mean-square</td>
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<td>Root-mean-square</td>
<td>Surface monopolar, bipolar, Laplacian</td>
</tr>
<tr>
<td>Gabriel et al.</td>
<td>2007</td>
<td>Biceps Brachii</td>
<td>Isometric</td>
<td>Linear</td>
<td>Root-mean-square, mean square amplitude</td>
<td>Surface Bipolar</td>
</tr>
</tbody>
</table>

Key: IEMG = integrated electromyographic activity; RMS = root-mean-square; MSA = mean spike amplitude.
Mean power frequency

The surface electromyographic signal can be viewed as a combination of sinusoids of different frequency that when summed together yield the original signal. However, the weight of the contribution of each particular sinusoid to the overall signal is different. Fourier analysis is based on a least squares fit of sinusoids of different frequencies to the overall signal wherein the weight of each particular frequency is analogous to a regression coefficient. Consider a plot wherein the x-axis is the frequency of sinusoids that constitute the signal and the y-axis is the magnitude of the weighting coefficient to denote the strength of the contribution of each particular frequency. The plot would show the "spectrum" of frequencies within the signal and their relative contribution. The weighting coefficient is really the "power" of a particular frequency and their sum gives the total power present in the electric signal.

The normalized "power spectrum" of frequencies for the surface electromyographic signal is presented in Figure 5. In this case, the frequency with the strongest contribution has a power of 1.0. The power spectrum is analogous to a frequency distribution curve that has a slightly positive skew, with the same measures of central tendency, i.e., mean and median. The range of frequencies has functional meaning. Motor units generally fire at rates from 10 to 40 Hz, and therefore strongly represent this portion of the power spectrum. The overall shape of the power spectrum is directly related to the shape of the motor unit action potentials within the muscle. There is generally no power in the signal beyond 400 Hz (Basmajian & DeLuca, 1985; DeLuca, 1997).
The mean frequency is the value that is the numerical mean of the frequencies present within the power density spectrum from the detected electromyographic signal, where \( P(f) \) is the power density spectrum of the surface electromyographic signal, the frequency of the mean power is:

\[
MF = \frac{\int_{0}^{f} f \times P(f)df}{\int_{0}^{f} P(f)df}
\]

The mean frequency is sensitive to a variety of physiological processes that happen during a muscular contraction. For example, as higher threshold motor units are recruited as part of the force gradation process, the mean frequency is increased due to the increased contribution of muscle fibers with higher conduction velocities (Sbriccoli et al, 2003). Muscle fiber conduction velocity has an impact on the general shape of the motor unit action potential. Faster conduction velocities decrease the duration of the motor unit action potential and increase the frequency content of the surface electromyographic signal (Lindström & Magnusson, 1977; Tanzi & Taglietti, 1981).
Figure 5. Depicts an ideal power spectrum of an electromyographic signal. 
The phenomenon of spectral compression is a shift to the left in the power spectrum, indicating that lower frequencies are dominating the surface electromyographic signal. Spectral compression has been observed under two conditions. A decrease (=5 to 10 Hz) in mean frequency has been observed at higher levels of maximal voluntary contraction (i.e., > 60%) in the absence of fatigue, and is believed to be due to motor unit synchronization which increase the force output of the muscle (Gabriel & Kamen, in press). A decrease in mean frequency is most often associated with muscle fatigue. During local muscular fatigue associated with intense contractions, there is an increased reliance upon anaerobic glycolysis for the production of adenosine triphosphate for metabolic energy. The production of lactic acid results in a decrease in the pH of the muscle and blood (Bouissou et al., 1989). In turn, the resulting ionic imbalance impairs the propagation of action potentials along the muscle fiber (Brody et al., 1991). The motor unit action potentials become longer in duration, which in turn decreases the frequency content of the signal (Lindström & Magnusson, 1977; Tanzi & Taglietti, 1981).

**Sex Differences**

**Muscle Strength**

Miller et al. (1993) examined strength and muscle characteristics in the biceps brachii and vastus lateralis of eight women and eight men. A significant correlation was found between strength and muscle physiological cross-section area. Men had significantly larger mean fiber areas than the women. However, no significant sex differences were found in the ratio between strength and physiological cross-sectional...
area or any motor unit characteristics (motor unit amplitude and muscle fibers per motor unit). The authors concluded that the greater strength of the men was primarily due to larger fibers, greater physiological cross-sectional area, and simply more lean muscle mass.

Holmäck et al. (2003) investigated determinants of ankle dorsiflexion strength and size in men and women. Muscle physiological cross-sectional area was assessed with magnetic resonance imaging, and biopsies were taken from the tibialis anterior to determine total numbers, areas and properties of type I and II fibers, and the relative content of myosin heavy chain isoforms, 1, 2, and 2x. Anthropometric measures were also obtained on the participants. The ratio of type II to type I fibers was significantly greater in men than women. Women had lower concentric and eccentric strength, smaller cross-sectional area, and smaller areas of type I and type II fibers. The eccentric strength and physiological cross-sectional area ratio was lower in women than men, but the concentric strength and physiological cross-sectional area ratio was similar for both sexes. Building upon the findings of Miller at al. (1993), it was concluded that physiological cross-sectional area was to a great extent determined by the body weight and fiber size.

**Electromyography**

The surface electromyographic power spectrum is sensitive to many variables that differ between studies. Electrode placement, inter-electrode distance, fatigue, fiber-type, fiber-type distribution in the muscle, blood flow, tissue filtering and skin preparation all have significant impacts on the shape of this curve (DeLuca, 1997; Kamen & Caldwell,
Depending on which muscle is being examined, the amount of subcutaneous fat over a muscle can vary greatly. A recent experimental study by Bilodeau and colleagues (2003), as well as a modeling study by Gabriel and Kamen (In Press), have both concurred that skinfold differences between sexes are insufficient to account for differences in electromyographical measures in the amplitude and frequency domain. Secondly, skinfolds taken from the tibialis anterior have much smaller values compared to most muscles. Concurrently, skinfolds from the tibialis anterior are not generally collected to calculate total body fat measurements with three, five, seven, or nine site tests, indicating the muscle has a tendency to be a lean muscle.

Cioni and colleagues (1994) examined sex-differences in surface electromyographic spectral parameters in the tibialis anterior during isometric dorsiflexion at different levels of force. The influence of electrode configuration (monopolar versus bipolar) was also investigated. Torque contractions were completed from 10 to 100% of maximal voluntary contractions in 10% increments.

The relationship between root-mean-square amplitude of the surface electromyographic signal and force was curvilinear for both males and females. Moreover, the root-mean-square values were nearly identical at lower submaximal levels of force, i.e., less than 60% of maximal voluntary contractions for monopolar electrodes and less than 30% of maximal voluntary contractions for bipolar electrodes. The median frequency values for the females were lower than males. Females also had a slower rate of increase in the median frequency than males. The different electrode configurations resulted in a different relationship between the surface electromyographic frequency and force. The median frequency values for the monopolar leads showed a linear relationship
across force levels for males. This same relationship shifted to a quadratic and curvilinear relationship with the bipolar leads, as well as an overall higher frequency composition. Females exhibited a curvilinear relationship for both electrode configurations, but the shape changed from concave to convex for monopolar versus bipolar configurations, respectively.

Bilodeau et al. (1992) examined sex-related differences in the surface electromyographic power spectrum during ramp contractions from 0 – 100% of maximal voluntary contractions in the biceps brachii, triceps brachii, and anconeus muscles. Males had greater mean and median frequency values than females across all force levels. The researchers suggest that the sex-difference is primarily due to skinfold thickness and fiber type characteristics. It was observed that women had a greater skinfold thickness than males. The low-pass filtering effects of the additional subcutaneous tissue can cause a shift in the power spectrum towards the lower frequency bands, as well as a marked decrease in the amplitude of the surface electromyographic signal. The higher frequency content for the surface electromyographic signal in males may also be due to muscle fiber type differences. It has been observed that type II fibers have a greater firing rate, as well as faster muscle conduction velocity, compared to type I fibers (Kupa et al, 1995).

Gabriel and Kamen (in press) recently reported significant negative correlations between skinfold thickness at the electrode recording site and the amplitude and frequency of the biceps brachii surface electromyographic signal. However, when used as a covariate, skinfold thickness failed to account for significant differences in surface electromyographic activity. This finding provides indirect support for the idea that the sex-differences may reside primarily on some aspect of muscle physiology.
Pincivero et al. (2001) examined the influence of sex on the median frequency in three quadriceps muscles: the vastus lateralis, vastus medialis, and rectus femoris. The vastus lateralis exhibited the largest change in median frequency as a function of contraction intensity, and males had a significantly higher median frequency than females. The investigators suggested that the vastus lateralis may possess a higher percentage of type II fibers which would give rise to higher frequency electromyographic signals with increasing contraction intensity. Bilodeau et al. (2003) supported these findings by demonstrating that males had a more pronounced increase in mean frequency of the vastus lateralis. However, Bilodeau et al. (2003) also observed significant differences in the vastus medialis and rectus femoris in both sexes as well. These differences may likely be due to the specific location of the electrodes on the muscles which would reflect the variability of fiber characteristics across individuals and within a given muscle.
CHAPTER III

METHODS

A minimum of 36 subjects, 18 females and 18 males, were needed to participate in the investigation and complete the necessary trials (see Appendix A). In the present study 50 males and 50 females were evaluated because the true population means and standard deviations are unknown and could possibly have been greater than those calculated. Participants were recruited from undergraduate and graduate students attending Brock University. The subjects were screened by a fitness questionnaire (PAR-Q) to determine their ability to participate in the study. The subjects were right foot dominant, free of lower body disabilities, and were not undertaking any lower body weight training. The subjects signed an informed consent form that complies with Brock University Ethics Board (REB-02-284) before participating in the study.

Experimental Design

Age, height, weight and the dimensions of the lower leg of each participant were taken upon arrival to the Electromyographic Kinesiology Laboratory (see Appendix B, Anthropometric measurements). Before beginning the experiment, participants performed three maximal isometric contractions of the ankle dorsiflexors. The contractions were 5-seconds in duration with a 3-minute inter-trial rest interval. A target line on an oscilloscope was presented to each participant. The target represented 110% of peak value of the previous three contractions. If a participant was able to reach the
target line during a fourth trial, the maximal voluntary contraction value was updated (Baratta et al., 1998).

The maximal voluntary contraction was used to set submaximal target forces of 20, 40, 60, and 80%. After three minutes of test, three trials at each percentage of maximal voluntary contraction were performed. A total of twelve 5-second step-contractions, each at 3-minute intervals were completed. Presentations of the submaximal conditions were randomized. The required force level was presented as a horizontal line on an oscilloscope (Hitachi, VC-6525) placed in front of the subject. Another three trials at 100% of maximal voluntary contraction were performed three minutes after the submaximal contractions to test for fatigue. Following the contractions, measurements for electrical impedance and skin temperature were taken again. A graphical representation of the experimental design is given below.

**Experimental design – graphical representation**

- Five seconds per contraction
- Three minutes rest between each contraction
Recording surface electromyographic activity and force

The motor points of the tibialis anterior were identified using a stimulating cathode probe. The train rate on the stimulator (Grass S88, Astro-Med Inc., West Warwick, RI) was set at 5 pulses per second and the stimulus duration was set at 1 msec. The cathode was moved around the tibialis anterior to locate the motor point. The motor point is a surface region wherein the lowest level of stimulation would produce a minimal muscle twitch (Christie et al, 2005). The surface electromyographic electrodes were placed on the tibialis anterior, away from the electrically identified motor points, towards the distal tendon.

The recording surface of the lower leg was shaved, abraded with NuPrep®, and then cleaned with isopropyl alcohol prior to electrode placement. The electrode-skin input impedance was always less than 10 kΩ (Grass EZM5, Astro-Med Inc., West Warwick, RI). Two Ag/AgCl recording electrodes (Grass F-E9M, Astro-Med Inc., West Warwick, RI) were then placed in a bipolar configuration with an inter-electrode distance of 1cm. The ground electrode (CF5000, Axelgaard Manufacturing Co., Ltd, Fallbrook, CA) was placed on the patella.

Participants were placed inside a Faraday cage to reduce electrical inputs from outside sources. Participants sat in an adjustable apparatus designed to isolate the action of the dorsiflexors in an isometric contraction (see Figure 6). The hip, knee, and ankle were placed at 90 degrees of flexion in the sagittal plane. The waist and knee were secured to minimize extraneous movement. A metal plate over the metatarsals was used to secure the bare foot to the load cell assembly. The load cell (JR3 Inc., Woodland, CA)
was located underneath the footplate and was aligned with the first metatarsophalangeal joint.
Figure 6. Experimental set-up and subject test position (left panel). Bipolar electrodes are located on the tibialis anterior. The knee and ankle are strapped in at 90° with the load cell located directly below the foot clamp. Representative force and surface electromyographic activity traces are shown for one pilot subject (right panels). The force and surface electromyographic traces were generated during isometric dorsiflexion at 20, 40, 60, 80, and 100% of maximal voluntary contraction as planned in the experimental design.
The surface electromyographic signal was band-pass filtered (10-500Hz) and amplified (Grass P511, Astro-Med Inc., West Warwick, RI) to maximize its resolution on the 16 bit analogue-to-digital converter (NI PCI-6052E, National Instruments, Austin, TX). All signals (force and surface electromyographic) were sampled at 2048 Hz, with a computer-based data acquisition system (DASYLab, DASYTEC National Instruments, Amherst, NH). The data was then stored on a Pentium III PC for off-line processing (Seanix Technology Inc., Blaine, WA).

**Data reduction and analysis**

Prior to data reduction and analysis, power spectral analysis was performed on the baseline signals to detect the presence of power line noise and other forms of interference in the surface electromyographic signal. A program in MATLAB (The Math Works Inc., Natick, MA) was used to calculate the force, root-mean-square, mean frequency, and median frequency from a one-second epoch of data during the first-half of the contraction. The location of the data window is designed to avoid any initial overshoot and represent the first stable portion of the force curve within the target area (±2.5%) (Gabriel et al, 2007).

The means of the three trials for each force level were used for statistical analysis. To account for individual differences in sEMG associated with physiological cross-sectional area of the muscle, electrode location, and subcutaneous tissue, the values generated during the post-test 100% MVC condition were used as a covariate in a two-factor repeated measure analysis of variance (ANOVA). The analysis of covariance is preferred over other normalization methods, which can change the ranking of subjects.
and the variance of the distribution (Tanner 1949; Lindquist 1953). There was one between group factor (males versus females) and one within group factor (force levels). Orthogonal polynomials were used to identify trends (i.e., linear, quadratic, or cubic) in the means across force levels. The statistical procedures were performed in SYSTAT (SPSS Inc., Chicago, IL) with alpha set at the 0.05 probability level.
CHAPTER IV

RESULTS

Pre and Post Contractions

The 100 % maximal voluntary contractions that were completed immediately before and after the experiment were analyzed to determine the presence of fatigue. The mean difference in force from the pre and post maximal voluntary contractions was less than 2 N (230 N and 228 N, respectively), and was not significantly different \((P<0.05)\). The mean difference in root-mean-square from the maximal voluntary contractions pre and post was 0.0008 mV (0.52 mV and 0.52 mV, respectively), and was not significantly different \((P<0.05)\). The mean difference in mean power frequency from the maximal voluntary contractions pre and post was 0.58 Hz (108 Hz and 108 Hz, respectively), and was not significantly different \((P<0.05)\). The mean difference in median power frequency from the maximal voluntary contractions pre and post was 0.84 Hz (88 Hz and 87 Hz, respectively), and was not significantly different \((P<0.05)\).

Table 2 shows the raw mean values for the anthropometric measurements and EMG data observed during the study.
Table 2 – Subject Characteristics
Values are means ± standard deviations of anthropometric measurements, force (N), root-mean-square amplitude (RMS), and mean power frequency (MPF) at 100% of maximum voluntary contraction.

<table>
<thead>
<tr>
<th>Measure</th>
<th>Males (n=50)</th>
<th>Females (n=50)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Mean ± SD</td>
<td>Mean ± SD</td>
</tr>
<tr>
<td>Age (years)</td>
<td>22.8 ± 3.7</td>
<td>20.7 ± 2.0</td>
</tr>
<tr>
<td>Mass (Kg)</td>
<td>84.6 ± 10.8</td>
<td>64.9 ± 6.9</td>
</tr>
<tr>
<td>Height (cm)</td>
<td>180 ± 6</td>
<td>164 ± 16</td>
</tr>
<tr>
<td>BMI (kg/m²)</td>
<td>26.1 ± 1.5</td>
<td>24.1 ± 1.9</td>
</tr>
<tr>
<td>Leg length (cm)</td>
<td>38 ± 2</td>
<td>35 ± 2</td>
</tr>
<tr>
<td>Leg circumference (cm)</td>
<td>38 ± 3</td>
<td>36 ± 3</td>
</tr>
<tr>
<td>Dorsiflexion force (N)</td>
<td>269 ± 61*</td>
<td>191 ± 51*</td>
</tr>
<tr>
<td>RMS SEMG (mV)</td>
<td>0.524 ± 0.363</td>
<td>0.508 ± 0.277</td>
</tr>
<tr>
<td>MNF SEMG (Hz)</td>
<td>115 ± 27*</td>
<td>102 ± 27*</td>
</tr>
</tbody>
</table>

* denotes significance set at the 0.05 level

Force

Males were stronger than females at all force levels (Figure 7) with respect to dorsiflexion force output \((P<0.05)\). The mean difference \((78 \text{ N})\) from \(20 - 100\%\) of maximal voluntary contraction was significantly different between sexes \((P<0.05)\). Both groups did exhibit linear trends in dorsiflexion force \((P<0.05)\) which accounted for over 99\% of the total variance \((P<0.05)\). There was a significant interaction between the groups in terms of the linear trend \((P<0.05)\). The rate of increase in force was greater for males than for females \((P<0.05)\).
Figure 7. Dorsiflexion force (values are means ± standard deviations). Males achieved 269 N (± 61) and females achieved 191 N (± 51).

Root-mean-square

The root-mean-square measure was similar in males and females across all force levels (Table 2), differing by only 0.016 mV at 100% of maximum voluntary contraction ($P<0.05$), see Table 2. The repeated measures analysis of covariance also revealed no significant differences between the groups ($P<0.05$). Figure 8 presents the least squares means and standard deviations computed after adjusting for the significant covariate effect ($P<0.05$). There was a significant linear increase in root-mean-square amplitude for males and females that accounted for a majority of the variance in means across force levels namely, 97.5% for males and 93.9% for females. The means also followed a curvilinear pattern that resulted in a significant quadratic trend ($P<0.05$). The quadratic
trend component accounted for 2.5% and 6.5% of the variance in means for males and females, respectively.

![Tibialis Anterior Surface EMG Amplitude](image)

Figure 8. Surface electromyographic amplitude (values are means ± standard deviations). Males achieved 0.524 mill volts (mV) (± 0.36) and females achieved 0.508 mV (± 0.28), after correcting for the effect of the covariate.

**Mean power frequency**

The mean power frequency for males was 13 Hz greater ($P<0.05$) than for females at 100 % of maximal voluntary contraction, (Table 2). Figure 9 presents the least squares means and standard deviations computed after adjusting for the significant covariate effect ($P<0.05$). The repeated measure analysis of covariance failed to show any significant differences between groups ($P>0.05$). Both groups exhibited a significant
linear increase in the mean power frequency until 80% of maximal voluntary contraction; this linear component accounted for 67.6% and 86.9% of the variance in means across force levels for males and females, respectively. The pattern of means also resulted in a significant quadratic trend component ($P<0.05$). There was a plateau in the mean power frequency means between 80 and 100% of maximal voluntary contraction for females, while males exhibited a decrease. The quadratic trend component accounted for a greater percentage of the variance for males (25.9%) than for females (12.7%). However, the group by force levels interaction term for the quadratic trend component was not significant ($P>0.05$).

Figure 9. Surface electromyographic mean power frequency (values are means ± standard deviations). Males achieved 115 Hz (± 27) and females achieved 102 Hz (± 27).
CHAPTER V

DISCUSSION

This study re-examined the tibialis anterior surface electromyographic-force relationship between males and females. It was assumed that differences between males and females with respect to motor unit recruitment and firing rate strategies are manifested in the surface electromyographic – force relationship (Lawrence and DeLuca, 1983). It was hypothesized that males and females would exhibit the same surface electromyographic-force relationship because of the following reasons: (1) both sexes have a similar proportion of type I and II fibers in the tibialis anterior (Jaworowski et al, 2002); and (2) sex differences in the tibialis anterior cross-sectional area are not as great as other well-studied muscles (Holmbäck et al, 2003). To date, there has only been mixed support for this hypothesis (Cioni et al., 1994). The current study demonstrated that the pattern of change in root-mean-square and mean power frequency surface electromyographic activity across force levels was the same for both males and females. These results will be discussed in further detail.

Force

Males in the present study had a maximal isometric dorsiflexion force of $269 \pm 61$ N. This is similar to $262 \pm 19$N reported by Kent-Braun and Ng (1999) and $250.9 \pm 8.4$ N observed by Patten and Kamen (2000). In contrast, female participants in the current study had a much greater maximal isometric dorsiflexion force ($191 \pm 50$ N) than that observed by both Kent-Braun and Ng (1999) and Patten and Kamen (2000), $136 \pm 15$ N.
and 150.9 ± 4.2 N, respectively. Males in the current study were (26%) stronger than females. This is consistent with Holm bäck and colleagues (2003) who concluded that muscle cross-sectional area was the principal determinant of dorsiflexion strength (Miller et al., 1993). In general, the cross-sectional area of the tibialis anterior in males is only 20% larger than that for females (Jaworowski et al., 2002; Holm bäck et al., 2003).

Electromyography

The study by Cioni et al. (1994) is the only one to have measured the root-mean-square and median power frequency of surface electromyography for increasing levels of dorsiflexion force in a manner similar to the present study. Cioni et al. (1994) did not provide specific values for root-mean-square and median power frequency, but the means at 100% of maximal voluntary contraction estimated from their figures are very similar in magnitude to those reported in Table 1. The current study corroborates the finding Cioni et al. (1994) for the non-linear increase in root-mean-square amplitude for both males and females. However, the pattern of the median power frequency in the current work only partially supports Cioni et al. (1994). Cioni et al. (1994) reported a similar quadratic relationship that was observed in the current study for males, but the females in their study exhibited a linear increase across force levels.

The surface electromyographic-force relationship is affected by both the electrodes and physiological factors related to the gradation of muscle force (Farina et al., 2004; Lawrence and DeLuca, 1983). The quadratic increase in root-mean-square amplitude is a robust finding as Cioni et al. (1994) observed the same pattern of change for monopolar and bipolar electrode configurations. The only difference between the two
electrode configurations was a lower surface electromyographic magnitude for the bipolar electrodes. Lower root-mean-square magnitudes can be expected because bipolar electrodes act as a spatial filter for the electromyographic signal (Lindström and Magnusson, 1977).

The magnitude of the surface electromyographic signal is also dependent on the amplitude of the motor unit action potentials closest to the electrodes (Lawrence and DeLuca, 1983). Since type I fibers are situated towards the anterior of the tibialis anterior muscle (Henriksson-Larsén et al, 1983) and since males and females are similar with respect to the proportion of type I fibers (Holmbäck et al, 2003), the non-significant difference in root-mean-square magnitude between sexes is not surprising. This is particularly true since the selective inter-electrode distance (1cm) used in the current work focused the pick-up volume towards the anterior region of the tibialis anterior.

The slightly greater RMS surface EMG magnitude for males may be explained by the interaction of two additional factors. First, the larger cross sectional area for the TA in males (Jaworowski et al. 2002; Holmbäck et al. 2003) contributes to greater signal strength. Second, the amplitude of the muscle fiber action potential is proportional to \(d^{1.7}\), where \(d\) is the diameter of the muscle fiber (Rosenfalck 1969). Henriksson-Larsén (1985) showed that the TA muscle fibers for females are in general smaller than that for males. Thus, motor unit action potentials for the same muscle fiber type with the electrode pick-up area will be smaller for females. This explanation assumes that high threshold, larger motor units-distributed towards the posterior of the TA, are too distant to contribute significantly to the overall magnitude of the surface EMG signal. However, the ratio of type II to type I fiber area for the TA is between 13 and 16% greater for males.
than females (Jaworowski et al. 2002; Holmbäck et al. 2003). Therefore, the contribution of more type II fibers to the surface EMG signal for males cannot be eliminated. A difference in subcutaneous fat has also been tendered as a plausible explanation for sex-related difference in mean power frequency (Bilodeau et al. 1992). However, significant difference in skin-fold thickness have been shown to be insufficient in accounting for a greater RMS and mean power frequency surface EMG for male participants (Gabriel and Kamen, In press; Bilodeau et al., 2003).

The larger muscle fiber diameter for males also explains why the mean power frequency for males was greater than that for females. Muscle fiber diameter affects the frequency spectrum of the surface EMG signal through conduction velocity. Conduction velocity increases with nerve diameter and gives rise to a surface EMG signal with higher frequency content (Lindström and Magnusson 1977). However, muscle fiber type also affects the frequency spectrum of the surface EMG signal through muscle fiber diameter and conduction velocity. Higher threshold motor units have larger muscle fiber diameters and therefore faster conduction velocities (Gerdle et al., 2000; Lindström and Magnusson 1977; Kupa et al., 1995; Broman et al., 1985). It may be reasonably assumed that the anterior distribution of type I fibers in the TA and the highly selective inter-electrode distance used in the current work minimized the impact of sex-related fiber-type differences.

The surface electromyographic-force relationship for any given muscle is also governed by the combination of motor unit recruitment and rate-coding specific to that muscle (Lawrence and DeLuca, 1983). The recruitment range for a muscle is the percent maximal voluntary contraction beyond which no more motor units are recruited.
Modeling studies have shown that muscles with a narrow recruitment range (40-60% maximal voluntary contraction) depend on rate coding for the gradation of muscle force and have a non-linear surface electromyographic-force relationship (Fuglevand et al., 1993; Zhou and Rymer., 2004; Keenan et al., 2005). Muscles that have a broad recruitment range (80-90% of maximal voluntary contraction) rely primarily upon the recruitment of additional motor units for the gradation of force and have a more linear surface electromyographic-force relationship (Fuglevand et al., 1993; Zhou and Rymer., 2004; Keenan et al., 2005). The motor unit data of Feiereisen et al. (1997) indicate a recruitment range of 90% of maximal voluntary contraction for the tibialis anterior. However, the physiological data indicates that motor unit size increases exponentially (Feinstein et al., 1955; Keenan et al., 2005). The size of motor unit action potentials would therefore increase dramatically towards the end of the recruitment range. Waveform cancellation would prevent a highly non-linear increase in surface electromyographic amplitude, but may contribute to the mild quadratic trend component observed here (Keenan et al., 2005). Alternatively, or in some combination, rate-coding at higher force levels may contribute to the superposition of motor unit action potentials and result in a non-linear increase in the electromyographic amplitude.

Males and females exhibited a linear increase in mean power frequency mean values from 20 to 80% of maximal voluntary contraction. The pattern of the mean power frequency mean values exhibited by both groups is consistent with the findings of several investigators who have reported that an increase in the frequency content of the signal reflects the orderly recruitment of high threshold motor units with larger fiber diameters and faster conduction velocities (Moritani and Muro, 1987; Broman et al., 1985;
Solomonow et al., 1990; Kupa et al., 1995). The mean power frequency mean values for females plateaued between 80 and 100% of maximal voluntary contraction, while that for males they exhibited a decrease. The result was a significant quadratic trend component for both groups. It is logical that the mean power frequency means would plateau after 80% of maximal voluntary contraction as the recruitment range for the tibialis anterior is up to approximately 90% of maximal voluntary contraction. The decrease in the mean power frequency mean values for males is more difficult to explain. High firing rates and/or motor unit synchronization increases the probability of temporal overlap between motor unit action potentials. The resulting summations in the surface electromyographic interference pattern would have larger amplitude spikes with longer durations which decrease the frequency content of the signal (Fuglevand et al., 1993; Yao et al., 2000; Gabriel and Kamen, In press). At present, there is no experimental evidence which suggests there are sex-differences in either peak motor unit firing rates or levels of synchronization.

Bilodeau et al. (2003) reported similar findings for the vastus lateralis during isometric knee extension from 10 to 90% of maximal voluntary contractions. There was a quadratic increase in root-mean-square magnitude across force levels. The mean power frequency means exhibited a linear increase up to 50% of maximal voluntary contraction. There was then a decrease between 70 and 90% of maximal voluntary contraction that resulted in a quadratic pattern for the mean power frequency means. Males had higher root-mean-square and mean power frequency values than females. Bilodeau et al. (2003) linked their results with physiological data that showed males have larger type II muscle fibers in the vastus lateralis than females. They also observed a significant negative
correlation ($r=-0.74$) between skin-fold thickness over the vastus lateralis and root-mean-square, but not the mean power frequency. Since female subjects in the study also had significantly greater skinfold thicknesses, this could explain the lower root-mean-square amplitude.

Schick and colleagues (2002) compared the efficacy of imaging and spectroscopy techniques on measuring lipid content in the tibialis anterior. The males and females used in the study had nearly identical body mass indexes, 23.9 ± 3.3 (kg/m$^2$) and 23.3 ± 4.6 (kg/m$^2$), respectively. In comparison, the present study found that males had greater body mass indexes than females (Table 2). Schick and colleagues (2002) observed with the imaging technique, which was concluded to be more sensitive, that females did possess greater lipid content than males, 5.4% and 3.0% respectively. However, the overall lipid profile of the tibialis anterior is extremely low (Schick et al., 2002) and quite different than the vastus lateralis. Since there was no significant difference between sexes in the root-mean-square measure, it is reasonable to assume that either 1) the skinfold thickness is nearly identical between sexes, or 2) the differences in the lipid profile between sexes in the tibialis anterior is not great enough to produce sex differences in the myoelectric signal.

Summary - Conclusions

Males were slightly greater in strength than females, but the difference is consistent with the greater physiological cross-sectional area for the tibialis anterior in males. Therefore, we reject the hypothesis that males and females will not be different in strength. There was no difference in root-mean-square amplitude or patterns of means across force levels. This is primarily attributable to an equal ratio of muscle fiber types.
between sexes in the tibialis anterior as well as a function of the highly selective inter-electrode distance used. Therefore, we accept the hypothesis that males and females will have similar trends in the root-mean-square amplitudes across force levels. Males had a higher mean power frequency of the surface electromyographic signal than females, which is purported to be caused by the larger muscle fiber diameter of males. However, the pattern of means for the mean power frequency revealed no sex differences between males and females in the surface electromyographic-force relationship. Therefore, we accept the hypothesis that males and females will have similar trends in the mean power frequency measure across force levels. The significance of this research is related to demonstrating the importance of methodological controls in the application and location surface electrodes. Careful methodological controls revealed no significant differences between males and females with respect to the force gradation process.
References


Appendix A

Subjects

It is necessary to determine the number of subjects that will allow the investigator a reasonable opportunity to reject the null hypothesis when it is false. Sample size estimation procedures were conducted in accordance with Cohen’s (1988) Case 0 formula for independent means. This requires the a priori establishment of: (1) the level of significance; (2) the appropriate power value; (3) the mean (M) and standard deviation (SD) of the criterion measure; and (4) the effect size (ES), which is deemed important or non-trivial by the investigator. Cohen (1988) suggested that a reasonable risk of Type II to Type I error is 4-to-1, meaning that a mistaken rejection of the null hypothesis is considered four times as serious as mistaken acceptance. The 4-to-1 risk ratio is satisfied by choosing an $\alpha=0.05$ and a $\beta=0.20$, where power is 0.80.

There are no studies that report the mean and standard deviation of a measure relevant to the proposed study. Van Cutsem and colleagues (1998) provide data for the root-mean-square amplitude of the surface electromyography from the tibialis anterior during isometric dorsiflexion at 100% of maximal voluntary contraction. Unfortunately, the male and female data were combined, which resulted in the magnitude of the sex-differences being obscured. The next closest data available for sample size estimation was reported by Gabriel and Kamen (In press), who reported the finding of sex-differences in biceps brachii surface electromyographic activity. The mean root-mean-square surface electromyographic amplitude for females was $\mathcal{X}_1 = 1.078mV$, for males it
was $\chi_2 = 1.63 \, mV$, and the pooled variance was $\sigma_p = 0.573 \, mV$. To calculate the effect size, i.e., Cohen’s (1988) tabled “d” value:

$$d = \frac{(\chi_2 - \chi_1)}{\sigma_p} = \frac{1.63 \, mV - 1.078 \, mV}{0.573 \, mV} = 0.97$$

Linear interpolation is needed to estimate sample size from Cohen’s (1988) tables (page 55). Here, $d_1$ is $d=0.80$ and the tabled $d$-value is $v_1=26$, and $d_2$ is $d=1.00$ and the tabled $d$-value $v_2=17$. The following calculation indicates 18 subjects ($n=18$) per group, for a total of 36 subjects.

$$N = v_1 + \frac{v_2 - v_1}{d_2 - d_1} \times (d - d_1) = 18.35$$
Appendix B

Physical Activity Questionnaire

Personal Contact Information

Please fill out your personal information in case the researchers need to contact you after completing the collection process.

First name: ______________________
Surname: ______________________
Age: ______
E-mail: ______________________

Physical Activity Background

1. How many hours a week do you spend weight training? ______________________

2. How many years of weight training experience do you have? (minimum 1yr)

3. How many hours a week are you physically active (aside from weights)? ______

4. What is the most frequent mode of exercise you engage in? ______________________
PAR – Q

Electromyographic Kinesiology Laboratory

Brock University, St. Catharines, Ontario, Canada, L2S 3A1

PAR – Q

(A Questionnaire for People Aged 15 to 69)

Regular physical activity is fun and healthy, and increasingly more people are starting to become more active every day. Being more active is very safe for most people. However, some people should check with their doctor before they start becoming much more physically active. If you are planning to become much more physically active than you are now, start by answering the seven questions in the box below. If you are between the ages of 15 and 69, the PAR – Q will tell you if you should check with your doctor before you start. If you are over 69 years of age, and you are not used to being very active, check with your doctor.

Common sense is your best guide when you answer these questions. Please read the questions carefully and answer each one honestly: check YES or NO.

<table>
<thead>
<tr>
<th>Yes</th>
<th>No</th>
</tr>
</thead>
<tbody>
<tr>
<td>☐ ☐ 1. Has your doctor ever said that you have a heart condition and that you should only do physical activity recommended by a doctor?</td>
<td></td>
</tr>
<tr>
<td>☐ ☐ 2. Do you feel pain in your chest when you do physical activity?</td>
<td></td>
</tr>
<tr>
<td>☐ ☐ 3. In the past month, have you had chest pain when you were not doing physical activity?</td>
<td></td>
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<tr>
<td>☐ ☐ 4. Do you lose your balance because of dizziness or do you ever lose consciousness?</td>
<td></td>
</tr>
<tr>
<td>☐ ☐ 5. Do you have a bone or joint problem that could be made worse by a change in your physical activity?</td>
<td></td>
</tr>
<tr>
<td>6.</td>
<td>Is your doctor currently prescribing drugs (e.g. water pills) for your blood pressure or heart condition?</td>
</tr>
<tr>
<td>7.</td>
<td>Do you know of any other reason why you should not do physical activity?</td>
</tr>
</tbody>
</table>

If you answered yes to one or more questions:

Talk with your doctor by phone or in person BEFORE you start becoming much more physically active or BEFORE you have a fitness appraisal. Tell your doctor about the PAR - Q and which questions you answered YES.

You may be able to do any activity you want - as long as you start and build up gradually. Or, you may need to restrict your activities to those which are safe for you. Talk with your doctor about the kinds of activities you wish to participate in and follow his/her advice.

Find out which community programs are safe and helpful for you.

If you answered no to all questions:

If you answered NO honestly to all PAR - Q questions, you can be reasonably sure that you can:

1. Start becoming much more physically active - begin slowly and build up gradually. This is the safest and easiest way to go.

2. Take part in a fitness appraisal - this is an excellent way to determine your basic fitness so that you can plan the best way for you to live actively.

Delay becoming much more active:

If you are not feeling well because of a temporary illness such as a cold or a fever - wait until you feel better; or take part in a fitness appraisal. This is an excellent way to determine your basic fitness so that you can plan the best way for you to live actively.

Please Note:

If your health changes so that you then answer YES to any of above questions, tell your fitness or health professional. Ask whether you should change your physical activity plan.
I have read, understood and completed this questionnaire. Any questions I had were answered to my full satisfaction.

NOTE: The responsibility is yours to fill in the PAR – Q and participate within your own limitations, as this is individual, unsupervised activity. If the PAR – Q is being given to a person before he or she participates in a physical activity program or a fitness appraisal, this section may be used for legal or administrative purposes.

Informed Use of the PAR – Q: The Canadian Society for Exercise Physiology, Health Canada, and their agents assume no liability for persons who undertake physical activity, and if in doubt after completing this questionnaire, consult your doctor prior to physical activity.

Name of Participant (Please Print): ________________________________

_________________________________________ Date (day/month/year)
Signature of Participant

Witness:

Name of Witness (Please Print): ________________________________

_________________________________________ Date (day/month/year)
Signature of Witness
Anthropometric Measurements

Subject #: 

Height (cm): 
Weight (kg): 

Fine measurements

Calf circumference (cm): 

Lengths (cm):
1. Fibula head to lateral malleolus 
2. Medial malleolus to 1st metatarsophalangeal joint 
3. Medial malleolus to base of foot 
4. Calcaneus to 1st metatarsophalangeal joint
Dorsiflexion EMG data collection sheet

Date: ____________________

Subject #: ______________

Subject Variables

Pre-Ω: ___________      Post-Ω: ___________

Pre-temp: ___________    Post-temp: __________

Contraction Chart

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</table>
Appendix C

Informed consent

INFORMED CONSENT DOCUMENT

Title of Project: Investigation of muscle electrical activity and force

Principle Investigator: David A. Gabriel, Ph.D., FACSM

Associate Professor Biomechanics
Department of Physical Education and Kinesiology
Brock University
500 Glenridge Avenue
St. Catharines, Ontario, Canada
L2S 3A1
Phone: 905-688-5550 ext. 4362
E-mail: dgabriel@brocku.ca

This study has been reviewed and approved by the Brock Research Ethics Board (#02-284). The Brock Research Ethics Board requires written informed consent from participants prior to participation in a research study so that they can know the nature and risks of participation and can decide to participate or not to participate in a free and informed manner. You are asked to read the following material to ensure that you are informed of the nature of this research study and how you will participate in it if you consent to do so. Signing this form will indicate that you have been so informed and that you give you consent.

Introduction

You are being asked to participate in research being conducted by David A. Gabriel, Ph.D. The on-going research program is focused on the relationship between skeletal muscle force and the electrical activity that it generates. The electrical signal of skeletal
muscle is measured from the skin surface, similar to electrocardiography (EKG) which measures the electrical activity of cardiac (heart) muscle. The skeletal muscle electrical signal is termed, electromyography (EMG). The signal can then be analyzed in a number of different ways, and each method has an impact on the EMG-to-force relationship. The long-term aspect of this project involves a continual search for better signal processing techniques on the computer, not changes in the measurement procedures.

You will come to the Electromyographic Kinesiology Laboratory (WH21). One session will last approximately two hours.

**Plan and Procedures**

David A. Gabriel, Ph.D. and/or his surrogate will conduct all testing. The following procedures will take place during each session. First, you will be asked to complete a physical activity questionnaire. This questionnaire is short in length and is designed to determine the quantity and type of exercise you engage in.

Upon completion of the questionnaire, the dominant leg will be prepared for exercise testing. Small areas of the tibialis anterior, soleus and the patella will be shaved, lightly abraded and cleansed with alcohol. These areas correspond to the location of the electrodes that will be taped to the skin surface. The electrodes will measure the electrical activity of arm muscles, similar to the more familiar electrocardiogram that measures the electrical activity of the cardiac (heart) muscles.

You will perform maximal effort isometric (same length) contractions to assess varying percentages of your maximum. These percentages will consist of 20, 40, 60, 80, and 100% of maximum effort. The contractions will be 5 seconds in duration and 3 minutes rest will be given between every contraction to avoid fatigue. Strength measurements
will be taken while seated in a chair designed to isolate the action of the dorsiflexor muscles. Adjustable supports on the chair and clamps on the chair will ensure stability and minimize extraneous movements. A foot plate will be attached just below the 1st big toe on the metatarsal joint. The leg not being tested rested close to the subject’s side. It is important not to hold your breath while strength testing.

**Risks and Discomforts**

It is not possible to predict all possible risks or discomforts that volunteer participants may experience in any research study. Based upon previous experience, the present investigator anticipates no major risks or discomforts will occur in the present project. Participants sometimes experience mild discomfort when the skin is gently cleaned and rubbed with a mild abrasive in preparation for electrode placement. On occasion, some subjects may experience skin irritation associated with the placement of the electrodes. This is usually very mild and goes away in a few hours, or a day.

There may be discomfort related to the delayed onset of muscle soreness associated with isometric contractions of the arm muscles. If muscle soreness does occur, it is usually very mild and should dissipate within 72 hours.

Maximal effort isometric contractions are associated with an increase in blood pressure. You must make sure that you do NOT hold your breath during maximal exertions. If you have received medical clearance and/or are already physically active, the risks are minimal.
**Voluntary Participation**

Participation in this study is voluntary. Refusal to participate will not result in loss of access to any services or programs at Brock University to which you are entitled. You will inform the investigator, David A. Gabriel, Ph.D., of your intention to withdrawal prior to removing yourself from this study.

**Discontinuation of Participation**

Participation in this research study may be discontinued under the following circumstances. The investigator, David A. Gabriel, Ph.D., may discontinue your involvement in the study at any time if it is felt to be in your best interest, if you not comply with study requirements, or if the study is stopped. You will be informed of any changes in the nature of the study or in the procedures described if they occur. It is important to remember that you are free to terminate your participation at any time, for any reason.

**Potential Benefits**

Participants will receive no direct benefits from participating in this study. However, participants should know that their willingness to serve as a subject for this experiment will help a Brock University researcher and other scientists develop new theories of exercise that will benefit individuals in the future.

**Costs and Compensation**

The cost of the test and procedures are free. You will not receive any form of compensation for your participation in this study.
Confidentiality

Although data from this study will be published, confidentiality of information concerning all participants will be maintained. All data will be coded without personal reference to you. Any personal information related to you will be kept in a locked office, to which only the investigator has access. Four investigators will have access to the data; however, names of participants or material identifying participants will not be released without written permission except as such release is required by law.

Persons to Contact with Questions

The investigator will be available to answer any questions concerning this research, now or in the future. You may contact the investigator, David A. Gabriel, Ph.D., by telephone during office hours at (905) 688-5550 extension 4362, or by email at dgabriel@brocku.ca. Also, if questions arise about your rights as a research subject, you may contact the Office of Research Services at (905) 688-5550 extension 3035. If you wish to speak with someone not involved in the study, please call the Chair of the Department of Physical Education and Kinesiology at (905) 688-5550 ext. 4361.

Consent to Participate

Certify that you have read all the above, asked questions and received answers concerning areas you did not understand, and have received satisfactory answers to these questions. Furthermore, you have completed the PAR-Q questionnaire indicating that you are physically able to participate. You willingly give consent for participation in this study. (A copy of the consent form will be given to you).

Name of Participant (Please Print): _________________________________
Signature of Participant

Date (day/month/year)
In addition to the considerations described in this document, the investigator fully intends to conduct all procedures with the subject’s best interest uppermost in mind, to insure the subject’s safety and comfort. I have fully explained the procedures of this study to the above volunteer. I believe that the person signing this form understands what is involved in this study and voluntarily agrees to participate.

__________________________  __________________________
Date (day/month/year)        David A. Gabriel, Ph.D., FACSM

Associate Professor Biomechanics

Department of Physical Education and Kinesiology