

**Reliability of Muscle Fibre Conduction Velocity in the Tibialis Anterior**

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**A thesis submitted in partial completion of the requirements for the degree of  
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(Kinesiology)**

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**ABSTRACT**

**RELIABILITY OF MUSCLE FIBRE CONDUCTION VELOCITY**

**IN THE TIBIALIS ANTERIOR**

**AUGUST 2010**

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Refinement of surface electromyographic (sEMG) techniques for recording voluntary muscle activity offers further opportunity for use as both a research and clinical tool. Recent efforts directed at using muscle fibre conduction velocity (MFCV) to assess neuromuscular disorders have had difficulty in achieving high test-retest reliability across multiple sessions.

Three days of testing were conducted on 21 males and 19 females with at least 48 hours between each session. Subjects performed three isometric contractions of the dorsiflexors at 100 percent maximal voluntary contraction. Maximum force, root-mean-square sEMG amplitude, the frequency of mean power (MPF), and MFCV were obtained via single- (SD) and double differential (DD) recordings and then evaluated using the intraclass correlational analysis of variance technique.

All measures exhibited high reliability coefficients ( $R=0.83 - 0.98$ ), except for MFCV measured by DD recordings ( $R=0.65$ ). It was thus concluded that the methodological procedures put in place were only effective utilizing single differentiation.

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# CHAPTER I

## DEVELOPMENT OF THE PROBLEM

### Introduction

Surface electromyography (sEMG) is the measurement of muscle electrical activity from the skin surface. A number of techniques have been employed to explore the relationship between sEMG and muscle physiology. The thesis focused on the reliability of an EMG method used to measure muscle fibre conduction velocity (MFCV). Muscle fibre conduction velocity refers to the propagation of action potentials traveling from a neuromuscular junction or innervation zone along the muscle fibres, towards the tendon (Mase et al., 2006). The range of values is between 2 and 6 m/s and is associated with the range of muscle fibre diameters for the different motor unit types (Lange et al., 2002; Merletti et al., 1995; Nishihara, 2005; Zwarts, 1989).

Muscle fibre conduction velocity is altered with the gradation of muscle force, local muscular fatigue, and neuromuscular disorders. It is important to have confidence that changes in muscle fibre conduction velocity are associated with real physiological events and not measurement error. High reliability is particularly important when monitoring changes across multiple sessions (day-to-day) to assess the effects of an intervention or the clinical progression of neuromuscular disorders. While there appears to be similarity between reproducibility and reliability, there is a distinct difference between the two in the statistical domain. Reproducibility refers to the ability of a test or experiment to be repeated, regardless of the testing apparatus that is used, experimenter

or location, whereas reliability refers to the ability of a test or measurement to remain consistent over repeated testing sessions of the same subject while following an identical protocol (Farina et al., 2004). Studies on the reliability of MFCV across multiple test sessions have thus far produced mixed results (Farina et al., 2004; Merletti et al., 1995).

Merletti and colleagues (1995) showed that only 83 of 150 electrically evoked contractions of the tibialis anterior (10 subjects) resulted in MFCV values within acceptable physiological limits (2-8 m/s). Muscle fibre conduction velocity was determined using double differential electrodes and the cross-correlation function to measure the signal delay between detection surfaces. Even with the non-physiological values omitted from the analysis, the intraclass correlation coefficient was still “poor” ( $R=0.11$ ). In contrast, Farina et al. (2002) studied MFCV during submaximal (50%) isometric contractions of the biceps brachii and reported an intraclass correlation coefficient that was “good” ( $R=0.75$ ). The increase in reliability may be attributed to a more sophisticated electrode detection system (matrix) combined with a signal processing technique (maximum likelihood) that allowed for more stable MFCV estimates.

Despite the improvement in reliability associated with more sophisticated methodology, the reliability of MFCV is still insufficient ( $R>0.90$  required) to monitor changes across multiple sessions (day-to-day) to assess the effects of an intervention or the clinical progression of neuromuscular disorders (Kimura et al., 2001; Merletti et al., 1995). Both the study by Merletti and by Farina’s group indicated the main source of error was a result of the day-to-day placement and “replacement” of surface electrodes in the same location so that they aligned with respect to the longitudinal axis of the muscle fibres. Off-axis alignment of the surface electrodes can result in the over estimation of

MFCV and is most likely the main source of non-physiological values (Farina et al., 2004; Sadoyama et al., 1985; Sollie et al., 1985).

There are two methods for determining electrode alignment with respect to the muscle fibres. The most prevalent method involves the use of an electrode array or matrix to visually inspect individual action potentials (Farina et al., 2004; Merletti et al., 1995; Kleine et al., 2001; Gazzoni et al., 2005). Visual inspection involves maximizing one or more of the following criteria: (1) signal amplitude; (2) the signal delay between detection surfaces; and (3) the similarity in action potential shape across channels. The second method is to evoke a small twitch in the muscle fibres and mark their orientation on the skin surface on the basis of the contraction (Zwarts & Arendt-Nielsen, 1988).

Using a twitch contraction to identify the muscle fibre orientation is appealing because it utilizes simple instrumentation that is readily available at the clinical level.

Unfortunately, the reliability of MFCV obtained by twitch identification of muscle fibre orientation has not been studied. Technological advances in electrode detection systems and signal processing techniques do offer distinct advantages, but may not be necessary if a simple approach can be demonstrated to offer the same reliability of measurement. The main advantage would be widespread use of MFCV as a physiological measure to investigate and/or assess the neuromuscular system.

### Statement of the Problem

The purpose of this study was to examine the reliability of MFCV in the tibialis anterior (TA) while using the twitch contraction to orient the surface electrodes with respect to the longitudinal axis of the muscle fibres. Muscle fibre conduction velocity

was calculated while participants performed isometric actions of the tibialis anterior at 100 percent of maximal voluntary contraction (MVC) on three non-consecutive test sessions. A linear array of four electrodes spaced 5 mm apart was placed distally from electrically identified innervation zones towards the distal tendon. Muscle fibre conduction velocity was then calculated using the cross-correlation technique to identify the time delay between detection surfaces.

### Problem Hypotheses

The following hypothesis was tested in this study: with sufficient methodological controls, MFCV obtained using a simple electrode array can exhibit good reliability as demonstrated by the intraclass correlational analysis of variance technique ( $R \geq 0.75$ ).

### Significance of the Study

There has been increased attention in sEMG research towards the development of noninvasive diagnostic tools that can provide insight into the health status of the neuromuscular system (De Luca, 2006). The gold standard in EMG has been the use of needle electrodes as they circumvent the limitations associated with subcutaneous tissue, possess the ability to measure small muscles, and provide greater information about individual muscle activity. However, in situations where a patient may be sensitive to needle electrodes, a non-invasive technique may facilitate electrodiagnostic characterization of the neuromuscular system (van der Hoeven, 1995; Zwarts, 1989; Türker, 1993). The proposed study will contribute more generally to the refinement sEMG methodology so that it can be used as a clinical tool. More specifically, an

evaluation of the measurement properties of MFCV in a statistical sense will help promote its use in research related to the assessment of neuromuscular disorders.

For example, Blijham et al. (2004; 2006) examined MFCV and its diagnostic capabilities. In particular, they have noted that patients with inflammatory myopathies have a significantly lower MFCV than a healthy population (Blijham et al., 2004). The authors hypothesized that the lower MFCV was due to an increase in the variability of both muscle fibre diameter and excitability of the sarcolemma, which are early signs of myopathic disease. In a second study, Blijham et al. (2006) performed a muscle biopsy investigation of the relationship between MFCV and fibre diameter in patients suffering from different neuromuscular disorders. Blijham et al. (2006) confirmed that muscle fibre diameter was the most important factor in determining MFCV in their patient population. The two studies by Blijham et al. (2004; 2006) highlight the potential for a sEMG methodology to contribute to assessment of the neuromuscular system in a clinically meaningful way.

#### 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. The cross-correlation method used to calculate MFCV provides an accurate representation of the signal delay between the successive waveforms as recorded by the electrode detection surfaces.

3. Dorsiflexion force is dominated by the tibialis anterior without substantial contribution from the toe extensor muscles (extensor digitorum longus and extensor hallucis longus).
4. The resulting maximal voluntary contraction generated by participants truly represents 100 percent of the maximal voluntary effort.
5. Coactivation from the triceps surae (soleus and gastrocnemius) during isometric dorsiflexion contractions is minimal.
6. The experimental protocol does not result in significant muscular fatigue.

#### Delimitations

The following delimitations were placed upon the study:

1. Only right leg dominant individuals between the ages of 19 and 35 years were studied.
2. Only one type of muscle contraction was investigated which was isometric actions of the dorsiflexors.
3. Only one contraction intensity, 100 percent of maximal voluntary contraction, was investigated.
4. Only a one joint action was investigated and it consisted of dorsiflexion.
5. The tibialis anterior was the only muscle studied.



## Limitations

The delimitations outlined above will impose the following limitations to the interpretation of the results and impact the generalizability of this study:

1. Since only right leg dominant individuals between the ages of 19 and 35 years were tested in this study, the results may not apply to the non-preferred limbs, or to humans of a different age group.
2. Since other types of muscle contractions involving complex joint actions were not investigated, the results may not apply to other contraction types (i.e., isotonic or isokinetic), at other contraction intensities, wherein more than one body segment is involved.
3. Since the tibialis anterior was the only muscle that was studied, the results may not apply to other muscle groups.

## CHAPTER II

### REVIEW OF THE LITERATURE

#### Tibialis Anterior

The tibialis anterior is illustrated in Figure 1 below. Its primary function is dorsiflexion and plantar flexion of the foot. The dorsiflexors contract concentrically during the swing phase of gait to decrease the angle at the ankle and eccentrically at the beginning of the stance phase to aid in control of the plantar flexion of the foot (Holmbäck et al., 2003). The tibialis anterior is therefore of great biomechanical importance in activities of daily living.

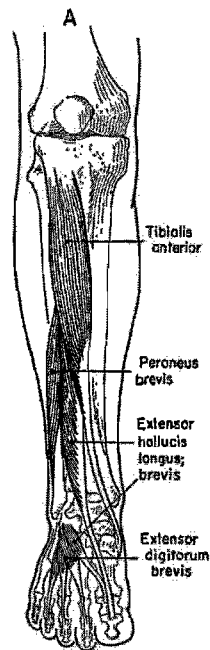


Figure 1. The tibialis anterior and other dorsiflexors. Luttgens, K. and Wells K.F. (1982). *Kinesiology: scientific basis of human motion* (7<sup>th</sup> Ed.). Philadelphia, PA: Saunders College Publishing, Figure 7-21, page 196.

A review of the literature on MFCV reveals that a disproportionate number of papers study the biceps brachii (Table 1) with relatively few examining the tibialis anterior (Table 2). The biceps brachii is an ideal candidate for MFCV research because it possesses long parallel (fusiform) fibres while the bipennate structure of the tibialis anterior discourages its use (Rababy et al., 1989). Mesin et al (2007) conducted a modeling and simulation study of MFCV and found that, when subcutaneous tissue is limited, the difference in estimation accuracy between unipennate and bipennate muscles is negligible. The study also confirmed the basic assumption that, as long as the electrode is placed in parallel with the muscle fibres, the tibialis anterior presents an excellent model for MFCV estimates (Mesin et al., 2007).

Another key characteristic associated with the tibialis anterior is the number of motor points associated with its relative surface area. Roy et al. (1986) studied the effect of electrode location in relation to MFCV and median frequency (MDF) in the tibialis anterior. It was found that MDF was greatest at the innervation zone and muscle-tendon junction and decreased in proportion with the distance away from these areas. The same was true for the stability (repeatability) of measurement; the two measures were most variable at the innervations zone and muscle-tendon junction. The study concluded that the ideal and most stable/repeatable region for measurement was the area between the distal tendon and the adjacent innervation zone. The study also found that the tibialis anterior had a range of 1-5 motor points with an average of 2.44 motor points. Staying away from these motor points is an important consideration in obtaining reliable measures. Masuda and Sadoyama (1987) studied a number of skeletal muscles, including the tibialis anterior, to determine which muscles could be used to detect the propagation

of MUAPs with a surface electrode array. The study demonstrated that the propagation of the action potential was best observed in the distal half to third of the muscle. The electrodes must therefore be placed between the most distal motor point and the distal tendon.

Holmbäck and colleagues (2003) examined the structure and function of the dorsiflexors in males and females, 21 to 30 years old. The anatomic (aCSA) and contractile (cCSA) cross-sectional areas, the percentage of type I and II fibres, and their relative areas were measured in 15 male and 15 female participants. Magnetic resonance imaging and muscle biopsies revealed no significant difference in the proportion of type I and II fibres (77.8% in males and 76.9% in females). However, type I fibres occupied a greater relative percentage of the tibialis anterior cross-sectional area in females, suggesting that the type II fibres in males were greater in diameter. This observation was further supported by the cross-sectional area data as measured by both aCSA and cCSA. The CSA for males was 20% greater than that for females, which also corresponded with a proportionally greater MVC.

The effect of the relationship between muscle fibre type and size on EMG variables was studied in isolated muscle preparations from the rat (Kupa et al., 1995). Kupa et al. (1995) were able to demonstrate a clear relationship between muscle fibre type, the size and shape of evoked potential, and resulting spectral parameters. The study also found a positive correlation ( $r = 0.78$ ) between MFCV and the percentage of type II fibres. Because muscle fibre diameter is the basis of this relationship, in muscles wherein the distribution of fibre types is different between males and females, it may be expected that males would have greater MFCV values than females. This is not the case

with the tibialis anterior as the male and female population, as shown by Holmbäck et al. (2003).

Table 1 – *Muscle fibre conduction velocity (MFCV) studies conducted on the biceps brachii. The type of contraction (voluntary or evoked), the percentage of maximal voluntary contraction (% MVC) that was used, the sex of the participants e.g. male (m) and female (f), and the MFCV values (mean ± standard deviation).*

<b>Authors</b>	<b>Contraction</b>	<b>% MVC</b>	<b>Gender</b>	<b>MFCV (m/s)</b>
Farina et al., 2000	Voluntary	50	1 m	3.62–4.05
Hunter et al., 1987	Voluntary	30	1 m	5
Kereshi et al., 1983	Evoked	–	29 m, 13 f	2.8–5.5
Krogh-Lund & Jorgensen, 1993	Voluntary	30	10 m	3.9 ± 0.69 – 5.1 ± 0.41
Lange et al., 2002	Voluntary	20	15 m	3.53–4.81
Lange et al., 2002	Voluntary	>50	15 m	3.11–5.59
Li & Sakamoto, 1996a	Voluntary	20,40,60	12 m	4.13 ± 0.23 – 10.23 ± 1.30
Li & Sakamoto, 1996b	Voluntary	20,40,60	12 m	4.12 ± 0.17 – 4.92 ± 0.22
Lowery et al., 2000	Voluntary	100	5 m, 3 f	3.08–2.78
Masuda & Sadoyama, 1986	Voluntary	10-40	8 m	3.4–3.5
Naeije & Zorn, 1982	Voluntary	50	8 m, 3 f	4.4 ± 0.4 – 3.7 ± 0.7
Nishihara et al., 2005	Voluntary	50	6 m, 4 f	3.93 ± 0.43
Nishizono et al., 1979	Voluntary	<30	4 m	4.2 ± 0.5 – 5.5 ± 0.5
Sadoyama et al., 1983	Voluntary	4, 8, 12 kg	2 m	4.05–4.28

Table 1 Continued

<b>Authors</b>	<b>Contraction</b>	<b>% MVC</b>	<b>Sex</b>	<b>MFCV (m/s)</b>
Sadoyama et al., 1985	Voluntary	30	1 m	3.78
Sadoyama & Masuda, 1987	Voluntary	0–100	2 m	3.28–3.77
Sakamoto & Li, 1997	Voluntary	30	10 m	3.74 ± 0.19 – 10.14 ± 1.13
Sollie et al., 1985	Voluntary	<30	unknown	3.98–4.23
Troni et al., 1991	Evoked	–	7 m, 8 f	1.25 ± 0.28 – 3.91 ± 0.76
van der Hoeven & Lange, 1994	Voluntary	50	10 m	4.20 ± 0.09 – 4.31 ± 0.11
van der Hoeven & Lange, 1994	Voluntary	100	10 m	4.30 ± 0.09 – 4.40 ± 0.12
van der Hoeven et al., 1993	Voluntary	20, 50	12 m	4.08 ± 0.14 – 4.16 ± 0.11
van der Hoeven et al., 1993	Voluntary	100	12 m	2.94 ± 0.12 – 4.50 ± 0.13
Yaar & Niles, 1991	Evoked	–	16 (unknown)	4.60 ± 1.5
Yaar & Niles, 1992	Evoked	–	130 m, 99 f	4.0 ± 1.2 – 4.2 ± 1.5
Zwarts, 1989	Evoked	–	7 m, 7 f	4.2 ± 0.5; 4.0 ± 0.3

Table 2 – *Muscle fibre conduction velocity (MFCV) studies conducted on the tibialis anterior. The type of contraction (voluntary or evoked), the percentage of maximal voluntary contraction (% MVC) that was used, the sex of the participants e.g. male (m) and female (f), and the MFCV values (mean ± standard deviation).*

<b>Authors</b>	<b>Contraction</b>	<b>% MVC</b>	<b>Sex</b>	<b>MFCV Values</b>
Andreassen & Arendt-Nielsen, 1987	Evoked	–	24 (unknown)	2.6–5.3
Broman et al., 1985a	Voluntary	50	8 m	3.37–4.87
Farina et al., 2002	Voluntary	25	8 m, 3 f	4.11 ± 0.3 – 4.32 ± 1.01
Merletti et al., 1995	Evoked	–	10 m	2–8 m/s (poor repeatability)
Roy et al., 1986	Voluntary	20	10 m	~3–5



## Muscle Fibre Conduction Velocity

The examination of muscle fibre conduction velocity (MFCV) is appealing because it is one of the few electrophysiological measurements that do not involve interpretation of its meaning, and it can provide insight into a number of characteristics associated with muscle activity. Muscle fibre conduction velocity refers to the velocity of propagation of action potentials along the muscle fibres (Broman et al., 1985a). This variable is measured by detecting the same myoelectric signal at two different locations along the muscle fibres, and estimating the delay in time between the two detection surfaces (Broman et al., 1985a; Naeije & Zorn, 1982). The velocity of propagation is simply the interelectrode distance (m) divided by the delay time (s).

### Estimation Techniques

Dips Analysis Technique. The first frequency domain technique used to determine MFCV was proposed by Lindstrom and Magnusson (1970) and is referred to as dips analysis. Dips analysis is based on calculating the power spectrum of the voluntary sEMG signal. It is important to remember that the voluntary sEMG is an interference pattern. The time-delay between signals reaching the two pairs of electrode detection surfaces result in a low energy “dip” in the power spectra the location of which corresponds to the MFCV (Figure 2). The bipolar electrode configuration behaves as a spatial filter. The transfer function is a “comb filter” that alternately passes some frequencies while canceling others (Figure 3). The spatial filtering properties of bipolar electrodes introduce dips into the power spectrum (Lindstrom & Magnusson, 1970; Sollie et al., 1985; Farina & Merletti, 2004).

The frequencies of the sEMG power spectrum at which low (or, close to zero) energy dips will occur is therefore given by the filtering transfer function for bipolar electrodes:

$$\sin(\omega d/v) \approx 0$$

where  $\omega$  is the angular frequency ( $\omega = 2\pi f$ ), and  $f$  is frequency in Hertz,  $d$  is half the distance between electrodes, and  $v$  is muscle fibre conduction velocity. The argument of the sine function results in a zero value when:

$$\omega_k d/v = k\pi, \quad k = 0, 1, 2, \dots$$

where  $k$  is the integer representing the dip number. The calculation of MFCV depends on the location of the first dip where the order  $k = 1$ . To find the dip frequency:

$$f_{dip} = \omega_1/2\pi$$

Substituting for  $\omega_1 = \pi v/d$  and rearranging:

$$v = f_{dip} 2d$$

Additional dips may occur for higher orders ( $k > 1$ ) at frequencies which are simple harmonics of  $f_{dip}$ .

The location of the first dip is dependent on the interelectrode distance and MFCV, and its sharpness is primarily dependent on the dispersion of the MFCV values (Farina & Merletti, 2004; Hunter et al., 1987). The ability to identify a well-defined dip determines the accuracy of this technique. Electrode misalignment and tissue

inhomogeneities can make dip identification even more difficult (Yaar & Niles, 1992; Yaar & Niles, 1991; Arendt-Nielsen & Zwarts, 1989; Sinderby et al., 1996; Farina & Merletti, 2004). Dips analysis is not often used as it is effective primarily at lower force contractions (Yaar & Niles, 1992). However, it has played a large part in introducing another set of techniques for estimation of MFCV.

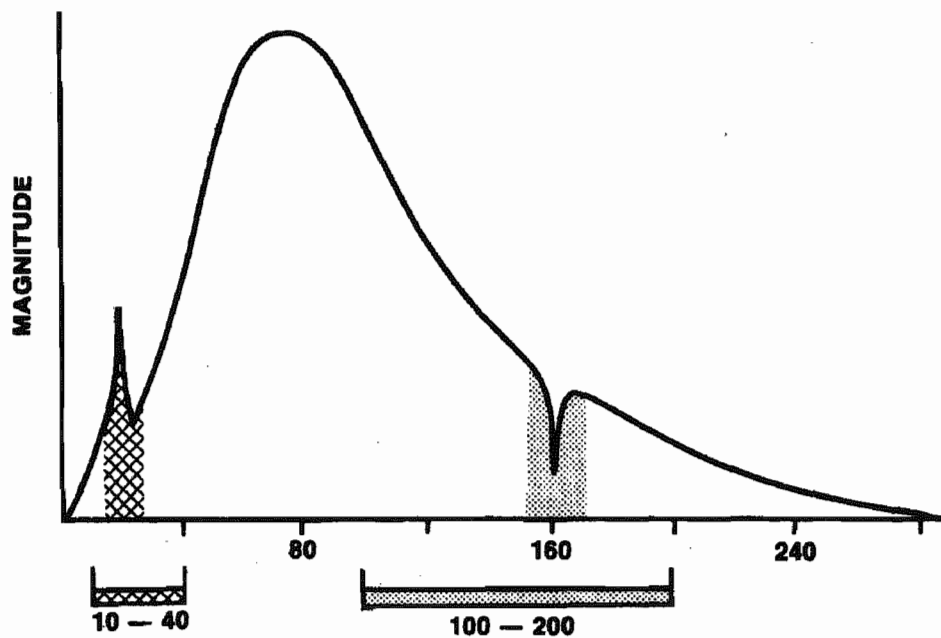


Figure 2. The idealized power spectrum for the surface electromyogram. Dips may occur in the frequency range between 100 and 200 Hz. Basmajian, J.V., & DeLuca, C.J. (1985). *Muscle alive: Their functions revealed by electromyography* (5th Edition). Williams & Wilkins, Baltimore: MD. (Figure 3.13, page 91).

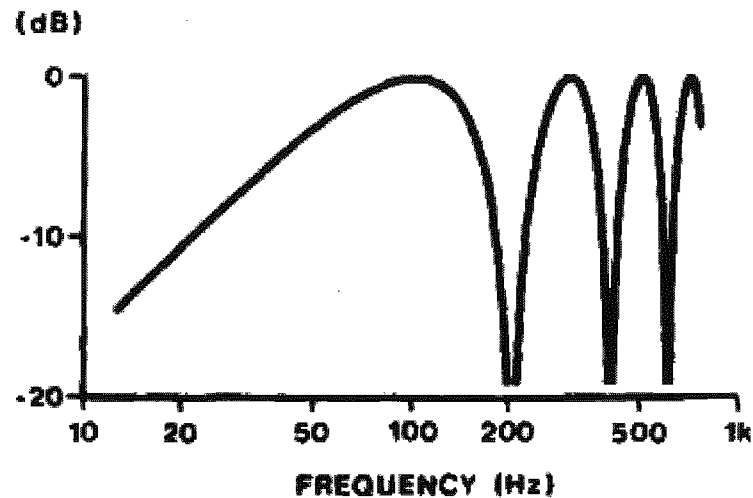


Figure 3. Surface electrode filter function for an inter-electrode distance of 2 cm and muscle fiber conduction velocity of 4 m/s. Lindström, L.H., & Magnusson, R.I. (1977). Interpretation of myoelectric power spectra: a model and its applications. Proceedings of the IEEE, 65: 653-662. (Figure 3, page 655).

The Zero Crossing Technique. The zero-crossing technique was first introduced by Lynn (1979) to overcome the limitation of dips analysis technique (Arendt-Nielsen & Zwarts, 1989). Lynn (1979) used the time delay between zero-crossings in the interference pattern and employed a digital filter to eliminate erroneous signal fluctuations (noise) that would otherwise create difficulty in matching two sEMG signals (Arendt-Nielsen & Zwarts, 1989). This technique was refined by Masuda and colleagues (1982) who introduced a threshold criterion for a zero-crossing to be considered significant. The first derivative of the waveform was high-pass filtered, followed by the estimation of peaks and the determination of time lags (Arendt-Nielsen & Zwarts, 1989). However, this technique has still been found to be sensitive to noise and electrode orientation as shown by a paper by Sadoyama and colleagues (1985).

A recently introduced technique based on detecting the latency between the “same” peaks in the interference pattern recorded across electrodes pairs, is similar in practice to the latency between zero-crossings. Lange (2002) took the derivative of both signals sampled at 4000 Hz. Polarity was then marked at all the zero-crossings. The following criteria were required: 15 samples before the zero-crossing had to be positive and the 15 samples following the zero-crossing had to be negative, otherwise the peak would be excluded from analysis. The same peak is then identified within a window across the two signals. A determination of the delays between all the peaks across both signals provides a range or spread of muscle fibre conduction velocities. The obtained values compared well with cross-correlation technique. There are other variants of the peak latency technique which differ in the threshold criteria for peak identification (Beck et al., 2004; Nishihara 2005). The basic problem is to identify the “true” peak rather than the noise embedded within interference pattern. The obtained values compared well with the cross-correlation technique. Variants of the peak latency technique differ in the threshold criteria for peak identification (Beck et al., 2004; Nishihara, 2005). The basic problem is to identify the “true” peak rather than the noise embedded within the interference pattern.

Cross Correlation Technique. The majority of papers on MFCV have used the cross-correlation technique. Naeije and Zorn (1982) first demonstrated the technique with success in 1982. This technique involves two sEMG recordings that are similar but not identical, due to changes in the action potential shape as it propagates between the two electrodes (Figure 4, left panel). The cross-correlation method compares the two signals on a point-by-point basis. The lag time ( $\tau$ ) associated with the maximum value

corresponds to the amount of time that the signal in the first channel would have to be shifted-back so that the peaks between the two channels would be aligned. This time-shift maximizes the cross-correlation between the two signals. Where the  $\otimes$  denotes the correlation, the cross-correlation  $R_{xy}(t)$  between the two signals  $x(t)$  and  $y(t)$  is:

$$R_{xy}(t) = x(t) \otimes y(t) = \int_{-\infty}^{\infty} x(\tau)y(t + \tau) dt$$

Consider the two sEMG signals in adjacent columns in an Excel spreadsheet. The function above is analogous to calculating the correlation between  $x(t)$  and  $y(t)$  as one of the two signals shifted down a row, one data point at a time. Since each sampled datum represents a point in time, the correlation between the two signals for that particular time-shift ( $\tau$ ) is then determined. This process is repeated for each shifted data-point, for the length of the two signals. Plotting the resulting correlation for each time-shift reveals the cross-correlation as a function of time-shift, termed the cross correlation function. The peak correlation then gives an estimate of the time-delay between the two signals (Figure 4, right panel). This time delay is divided by the inter-electrode distance to obtain the MFCV.

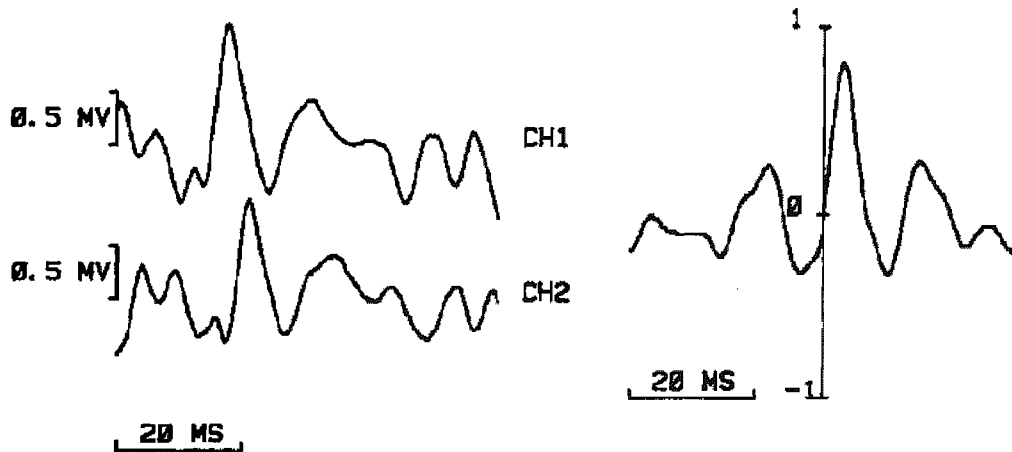


Figure 4. The same muscle fiber action potential recorded at two different electrodes on channel 1 (CH1) and channel 2 (CH2), respectively (left). Channel 1 is closer to the motor point. The cross-correlation function has a maximum value at a positive lag time ( $\tau$ ), indicating that channel 2 is delayed with respect to channel 1 (right). Arendt-Nielsen, L., & Mills, K.R. (1985). The relationship between mean power frequency of the EMG spectrum and muscle fiber conduction velocity. *Electroencephalography and Clinical Neurophysiology*, 60, 130-134. (Figure 1, page 131).

There are two ways to improve the precision of measurement that is limited by sampling rate. Farina and colleagues (2000) demonstrated improvement by employing a higher sampling rate, and by applying a simple interpolation function around the peak value. Roy and colleagues (1986) suggested a similar method wherein the cross-correlation peak is interpolated by multiplying it with the function by  $\sin(x)/x$  to preserve the frequency content of the frequency spectrum of the cross-correlation function itself (Roy et al., 1986). Yaar and Niles (1992) demonstrated the cross-correlation technique

was superior to the frequency domain based technique as it was less susceptible to experimental noise and misalignment of the electrodes.

### Electrode Considerations

Typically, at least three silver or silver-chloride recording surfaces placed along the direction of motor unit potential propagation are used to detect the latencies of action potentials. The three detection surfaces are used to create at least two differentially recorded signals (Lowery et al., 2002; Broman et al., 1985a). An additional recording surface may be included to yield two double-differential signals (Roy et al., 1986; Farina et al., 2002). The double differentiation minimizes the presence of non-delayed activity which can result in non-physiological MFCV values (Broman et al., 1985b; Farina et al., 2002; Hogrel & Duchêne, 2002). Stimulus artifact and far field potentials (muscle-tendon end effects) are also greatly reduced by double differential electrodes (Fiorito et al., 1994; Broman et al., 1985b; Farina et al., 2002). However, there is a trade-off between competing factors that must be considered. An inter-electrode distance of 5 mm minimizes changes in action potential shape between detection surfaces (Sadoyama et al., 1985; Sollie et al., 1985; Farina et al., 2004), and is minimally sufficient to prevent salt-bridge formation with the use of electrolyte gel (Rababy et al., 1989).

### Electrode Placement and Orientation

The motor point is an electrically defined region of the muscle. This dense collection of motor endplates may be identified electrically as a point on the skin surface where the lowest possible current will produce a muscle twitch (Roy et al., 1986). If a dense collection of motor endplates lies close to the surface of the skin, there should be a



lower threshold for muscle activation. Because of branching of the motor nerves, it is not unusual for the motor point to reside within a more broadly defined “innervation zone”. The innervation zone is defined through electrode mapping. Motor unit action potentials are generated, then propagate simultaneously, bidirectionally away from the innervation zone (Masuda & Sadoyama, 1987).

Sollie et al. (1985) and Sadoyama et al. (1985) reported difficulty in detecting the latency between signals if the electrodes were placed on the motor point. Both research groups also noted that the cross-correlation values were significantly lower when the electrode was placed on, or in close proximity to, the motor point (Sollie et al., 1985; Sadoyama et al., 1985). In contrast, a distinct region between the motor point and tendon was observed wherein the cross-correlation values were highest and the MFCV values were relatively constant (Sollie et al., 1985; Sadoyama et al., 1985). Sollie (1985) and colleagues were more specific and laid out a set of conditions that should be followed when examining conduction velocity via the cross-correlation method. As well, they found results similar to those of Sadoyama and colleagues (1985), where they found a distinct region where the highest correlation values were found and the conduction velocity calculated was near constant.

Roy et al. (1986) specifically addressed the issue of electrode placement relative to the motor point. They measured MFCV in the tibialis anterior and found that the highest cross-correlation values were obtained when the electrode was placed within a 3 – 4 cm region between the motor point and distal the tendon. It was suggested that the high cross-correlations were due to the more symmetrical propagation of action potentials as the muscle tapered towards a common tendon below the most distal motor point (Roy

et al., 1986). A number of investigators have confirmed the findings of Roy et al. (1986) to show that muscles have a distinct region where physiologically correct, repeatable/stable MFCV values may be obtained (Li & Sakamoto, 1996a; 1996b; Sollie et al., 1985; Sadoyama et al., 1985; Rababy et al., 1989; Andreassen & Arendt-Nielsen, 1987). Moreover there is a U-shaped function for MFCV values that extends from the motor point towards the tendon, with the highest, most unstable values at either end (Li & Sakamoto, 1996a; 1996b; Roy et al., 1986; Sadoyama et al., 1985).

A simple misalignment of an electrode can also result in values that are nearly double the accepted physiological values. Sollie et al. (1985) examined misalignment of the electrode relative to MFCV values. They found that misalignments of 5, 10 and 15 degrees in electrode placement relative to the muscle fibres resulted in overestimations of velocity of 1.7, 1.9 and 6.5%, respectively. This is due to the fact that the interelectrode distance is now multiplied by the cosine of the angle of misdirection, which is inherently a shorter distance (Sollie et al., 1985; Sadoyama et al., 1985; Farina et al., 2004) (Figure 5 & 6).

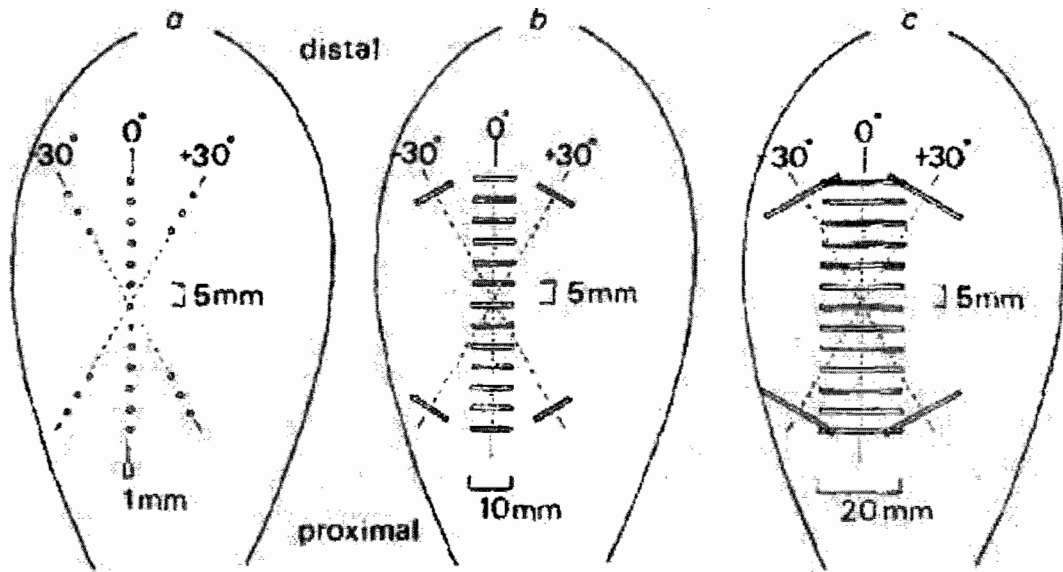


Figure 5. Schematic diagram of electrode placement. Myoelectric potentials were led off bipolarly from the adjacent pairs of the contacts in the electrode array. The contacts were spaced at 5 mm intervals. Contact widths were (a) 1 mm, (b) 10 mm and (c) 20 mm.

Sadoyama, T., Masuda, T., Miyata, H., & Katsuta, S. (1988). Fibre conduction velocity and fibre composition in human vastus lateralis. *European Journal of Applied Physiology*, 57, 767-771. (Figure 1, page 340).

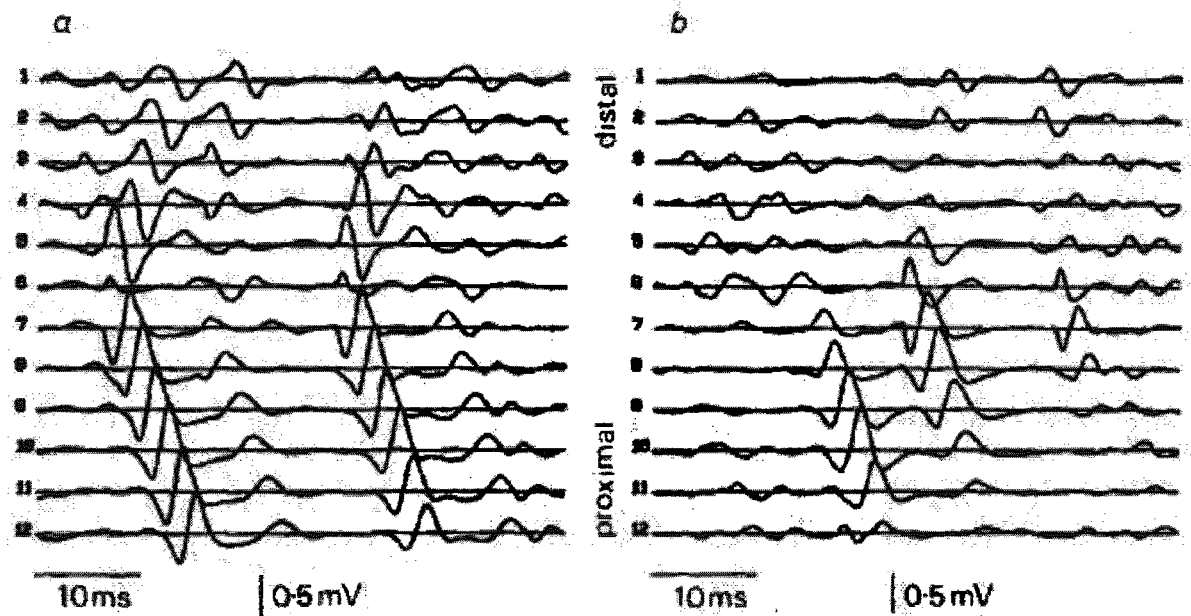


Figure 6. Raw tracings of 12 myoelectric signals obtained by the 10 mm width electrode (a) placed along the datum reference line and (b) rotated at + 30° with respect to the datum reference line. Contraction was 30 per cent of the maximum voluntary contraction. Sadoyama, T., Masuda, T., Miyata, H., & Katsuta, S. (1988). Fibre conduction velocity and fibre composition in human vastus lateralis. *European Journal of Applied Physiology*, 57, 767-771. (Figure 2, page 340).

### Muscle Fibre Characteristics

The relationship between MFCV and fibre diameter has been the subject of debate (Sadoyama et al., 1988; Kupa et al., 1995; Blijham et al., 2006). Originally, Sadoyama et al. (1988) failed to show a relationship between the two variables. The investigators studied distance runners and sprinters for their obvious differences in muscle fibre type on the assumption that the low threshold type I fibres have smaller diameters than the higher threshold type II fibres. As might be expected, the two groups displayed

differences in fibre type composition. However, there was an increase in fibre size relative to the athletic endeavor which blurred the differences between groups. The sprinter's type II fibres were 89.7  $\mu\text{m}$  in diameter and the distance runner's type I fibres were 85.3  $\mu\text{m}$ . A clear relationship between fibre diameter and MFCV was demonstrated more recently by Blijham and colleagues (2006) who showed a linear relationship between the two variables. It was also concluded that fibre diameter played the most important role in the determination of MFCV.

There are both physical and metabolic reasons why muscle fibre diameter is important in determining MFCV. The muscle fibre action potential (MFAP) can be represented by a traveling tripole (+ - +) (Loeb & Gans, 1986; Dumitru, 2000). The electrochemical events involved in generating the MFAP are illustrated in Figure 7. The electric potential associated with each electrochemical event is depicted immediately below the muscle fibre, as it would be recorded by an extracellular electrode. All these events are occurring at the same time as the action potential propagates along the muscle fibre, from left-to-right towards the electrode. However, imagine that we can freeze these events in time to explore the electrode recordings further.

Focusing first on the point of depolarization, when  $\text{Na}^+$  ions rush into the muscle fibre, they leave behind a relatively strong negativity in the extracellular space. This strong negativity is called a current sink because positive charges are drawn to it. If an electrode is placed directly over the depolarization event, a negative potential is recorded with respect to the extracellular space (position #1). However, the current sink is so strong that it attracts positive ions from the membrane area in front of the depolarization event. This forward membrane area is called a weak current source area because it

provides the positive ions that are drawn to the current sink. An electrode placed in front of the depolarization event would record a slight positivity (position #2). As positive ions leave the forward membrane area, the charge difference across the membrane decreases, which leads to passive depolarization of the muscle fibre. The ion channel-mediated rush of positive ions ( $K^+$ ) outside the muscle fibre gives rise to the repolarization event and is a strong current source. An electrode placed directly over the repolarization event would record a large positivity (position #3).

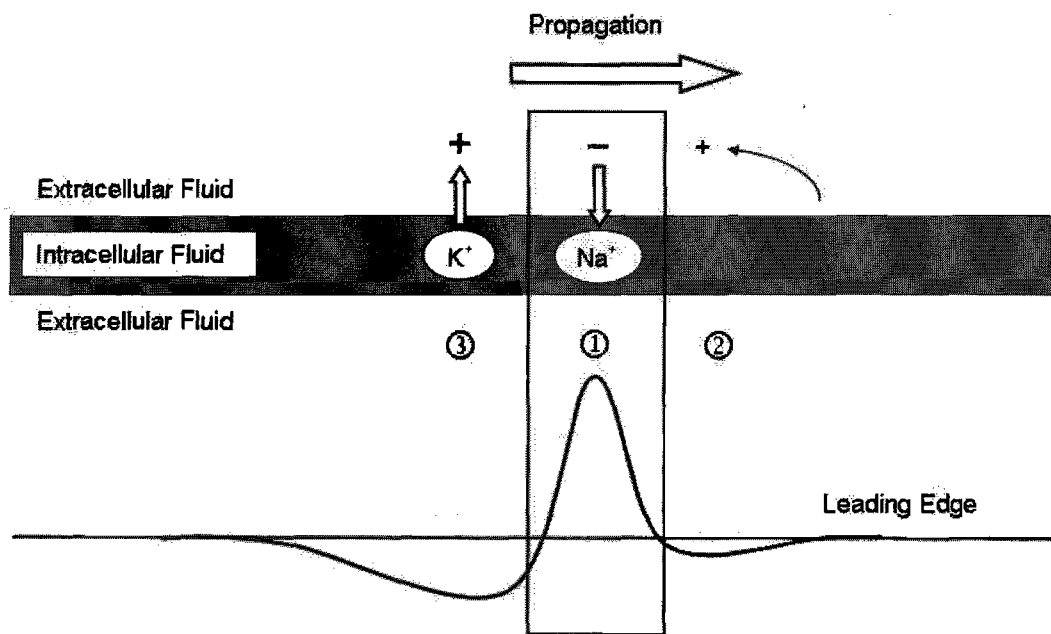


Figure 7. The electrochemical events associated with generating the muscle fiber action potential. Kamen, G., & Gabriel, D.A. (2009). Essential of electromyography. Human Kinetics, Champaign: IL. (Figure 2.5, page 26).

As the MFAP propagates along the muscle fibre from left-to-right towards the electrode, the leading edge (weak current source) is detected first, followed by the depolarization phase (current sink), and then the repolarization phase (strong current

source). The motor unit action potential (MUAP) is also triphasic because it is the linear sum of all its associated muscle fibres.

Electrical stimulation of the peripheral nerve results in the activation of a large number of motor units simultaneously. The triphasic wave form is still apparent in the simultaneous depolarization and repolarization of the recruited motor units. The evoked potential is logically termed, the massed action potential (or, M-wave). It is also called the compound muscle action potential (CMAP) due to the linear summation of all the constituent MUAPs. The large number of muscle fibres involved in the evoked response results in an electric potential that is several millivolts in magnitude.

A fundamental quantity in the electrophysiology of muscle and nerve fibres is the length constant ( $\lambda$ ) (Nicholls et al., 2001). It is the segment length at which the axial and radial resistances are equal. At distances greater than  $\lambda$  the axial resistance is greater than the radial resistance and the majority of the current leaks through the membrane. The formula for the length constant is:

$$\lambda = \sqrt{\frac{R_m a}{2\rho_m}}$$

where ( $\rho_m$ ) is the resistivity of the myoplasm (or, axoplasm), ( $R_m$ ) is membrane resistance, ( $a$ ) is muscle fibre diameter. The conduction velocity of action potentials depends on how quickly the membrane can be brought to threshold. If the area of the membrane in front of the active region is brought closer to threshold through the passive spread of positive charges, depolarization will have an earlier onset. The length of the

leading edge of the depolarizing current is dependent upon the length constant ( $\lambda$ ), which is a function of the square root of the diameter ( $a^{1/2}$ ).

Larger muscle fibres have a greater length constant, which means that the depolarizing current will travel farther forward, passively. Ultimately, this occurs because there is a decrease in axial resistance associated with an increase in fibre diameter. The action potential has a greater forward extent (or leading edge) that brings the membrane area farther ahead of the traveling dipole, closer to threshold. The membrane area will depolarize more rapidly. The result is an overall increase in conduction velocity with an increase in muscle fibre size. There is also a metabolic reason that an increase in muscle fibre diameter is associated with an increase in MFCV. Type II muscle fibres have larger diameters, but they also have a higher resting membrane potential and a greater action potential amplitude that is shorter in duration than type I fibres. Hanson (1974) suggested that the increased membrane excitability is attributed to higher intracellular  $[K^+]$  and lower intracellular  $[Na^+]$  for type II fibres. The mechanism appears to reside in a difference in the  $Na^+-K^+$  leak/pump ratio for type II versus type I fibres (Clausen et al., 2004).

### Force of the Voluntary Contraction

The relationship between force and MFCV remains a subject of debate. The main experimental paradigm used to explore the relationship between the two variables is the use of ramp or step isometric contractions at force levels varying from 0 to 100% of MVC. The assumption is that the gradation of muscle force will follow Henneman's (1957) size principle wherein the central nervous system recruits progressively larger



motoneurons. Both muscle and nerve fibre diameter for low threshold (type I) motor units are smaller than for higher threshold (type II) motor units (Andreassen & Arendt-Nielsen, 1987). Thus, the expectation is that there would be a progressive increase in MFCV as higher threshold motor units are recruited to increase the force of the muscle contraction (Okajima et al., 1998; Sadoyama & Masuda, 1987).

Studies that have tested this hypothesis on the tibialis anterior (Broman et al., 1985a; Andreassen & Arendt-Nielsen, 1987) and biceps brachii (Zwarts & Arendt-Nielsen, 1988; Lange et al., 2002) have reported a progressive increase in MFCV with force as predicted. The increase generally plateaued towards the end of the recruitment range of the muscle, providing additional evidence that the increase in MFCV was indeed associated with the recruitment of higher threshold, larger diameter fibres.

Unfortunately, counterexamples are found in the literature that suggest either no relationship or a weak one at best. Masuda et al. (1996) reported a lack of relationship between MFCV and force in the biceps brachii while Rababy et al. (1989) observed a very mild increase. Interestingly, Masuda et al. (1996) studied two other muscles: MFCV in the vastus lateralis demonstrated a strong dependence on contraction force while the tibialis anterior exhibited a moderate increase.

It is tempting to blame the discrepant findings on differences in the electrode detection systems, signal processing techniques, or type of contraction studied. However, the studies cited above followed well-accepted practices in the measurement of MFCV. There is, however, evidence that the relationship between force and MFCV depends on the rate of increase in force. Sbriccoli et al. (2003) investigated MFCV in the biceps brachii during ramp isometric contractions of the elbow flexors at 5, 10, and 20  $\text{MVC} \cdot \text{s}^{-1}$ , from 0

to 100% of MVC. The MFCV values increased rapidly until the first two and one-half seconds of the contraction at 5 and 10  $\text{MVC}\cdot\text{s}^{-1}$ ; it then plateaued for the remainder of the trial until 100% of MVC was reached. In contrast, there was a steady increase in MFVC from 0 to 100% of MVC during the 20  $\text{MVC}\cdot\text{s}^{-1}$  condition. If the relationship between force and MFCV depends on the rate of increase in force, it is reasonable to argue that the main difference between studies was “how fast” participants achieved the target force.

The focus of the study was on the reliability of MFCV using a simple electrode alignment technique. The MFCV of the tibialis anterior was assessed at the maximal voluntary force level. The logic behind the use of this contraction was that it would place our signal analysis at an optimal level of signal to noise ratio.

## Reliability

### General Principles

Intraclass correlational analysis of variance involves a consideration of both the “consistency” and “stability” of the criterion measures (Lindquist, 1956; Feldt & McKee, 1958; Hetherington, 1973). This section reviews these two basic concepts while the Statistical Analysis details their calculation. Consider the simple case in which reliability is assessed with the interclass correlation coefficient (Pearson’s “r”). The example is based on measuring the root-mean-square (RMS) sEMG amplitude for six participants during an isometric contraction of the tibialis anterior at 100% MVC, on two separate test sessions (Figure 8).

The interclass correlation coefficient ( $r = 0.93$ ) would indicate that the measure is highly reliable when inspection of the graph would indicate otherwise. In this case, the high interclass correlation coefficient indicates that participants maintained their relative ranking within the distribution of scores. The measure lacks consistency “within subjects” because participants were unable to reproduce their own score; it was different across the two sessions. The measure also lacks stability because the means “between subjects” on session one versus session two exhibited an increase ( $\Delta 0.31$  mV). To clarify further, the term “between subjects” is taken from an analysis of variance (ANOVA) which would be used to examine the difference in means “between” two or more groups of subjects. Although the same group of subjects is tested twice to yield two means, or a “repeated measure” on the group, the terminology is the same. Sole reliance upon the interclass correlation coefficient to determine reliability can therefore be very misleading.

The second case illustrated in Figure 9 is equally troublesome. The means obtained on session one versus session two exhibit excellent stability because they are identical. However, these identical means were obtained because the differences in scores produced by one subject were compensated for by differences in scores produced by another subject. The “net result” is that there is no difference between means across the two sessions. Testing for significant differences to demonstrate that the criterion measure is stable and therefore reliable is necessary, but not insufficient.

The ideal situation is depicted in Figure 10. That is, every subject can reproduce their own score perfectly. Perfect consistency results in an interclass correlation coefficient of  $r = 1.0$ . The following are critical by-products of the fact that subjects can

reproduce their own score: (1) one subject is distinctly different from the next, and (2) each subject maintains their relative ranking within the distribution scores across the two sessions. In this way the correlation coefficient is used to assess only part of reliability, “consistency”. The second half of reliability is the “stability” of means across test sessions. In the absence of any experimental manipulation, the group means should fluctuate very little across test sessions.

Referring back to Figure 8, the high (0.93) interclass correlation coefficient would ordinarily indicate highly reliable scores. However, each subject exhibited an average increase of 0.31 mV while maintaining their relative rank within the distribution of scores. It should be evident from the graph that the increase in means across test sessions actually decreased the interclass correlation coefficient down from  $r = 1.0$ . Thus, a test of significant differences (i.e., *t*-test or ANOVA) in means across sessions is a very important diagnostic component in the assessment of reliability. The ANOVA model can be extended to any level of measurement that is desired. The study had two levels of measurement: days and trials.

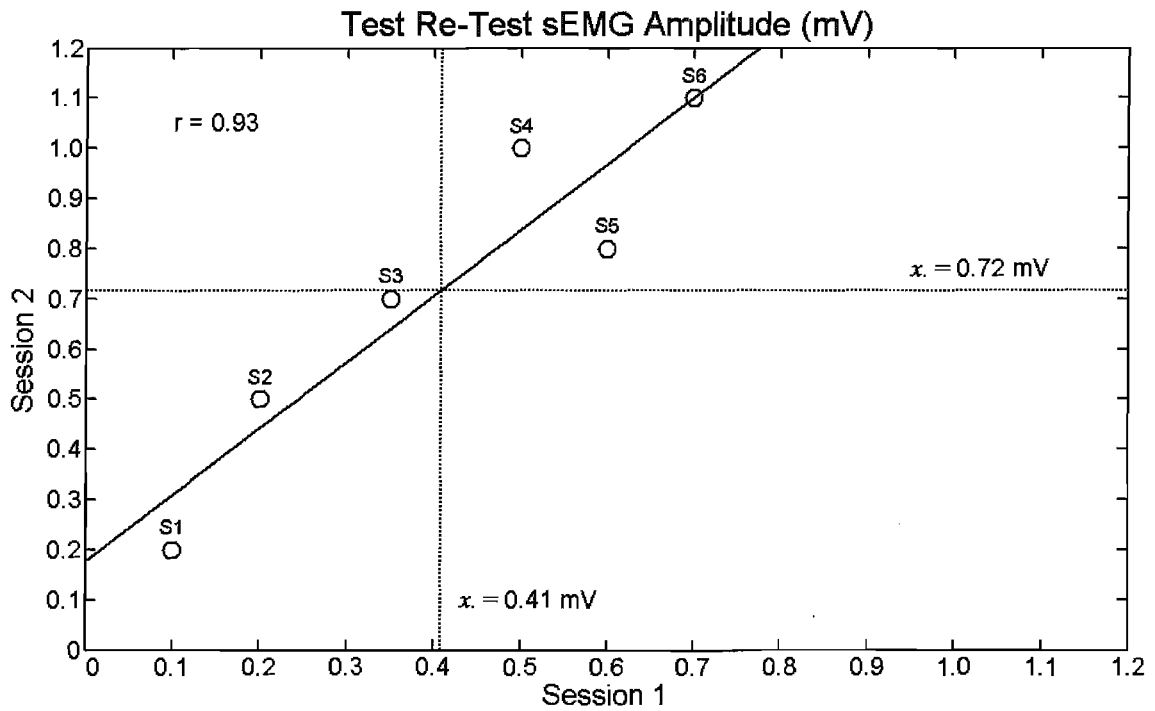


Figure 8. Test-retest reliability for root-mean-square amplitude of surface electromyographic (sEMG) activity. There is a high interclass correlation coefficient ( $r = 0.93$ ) but with a difference in means between the first and second test sessions.

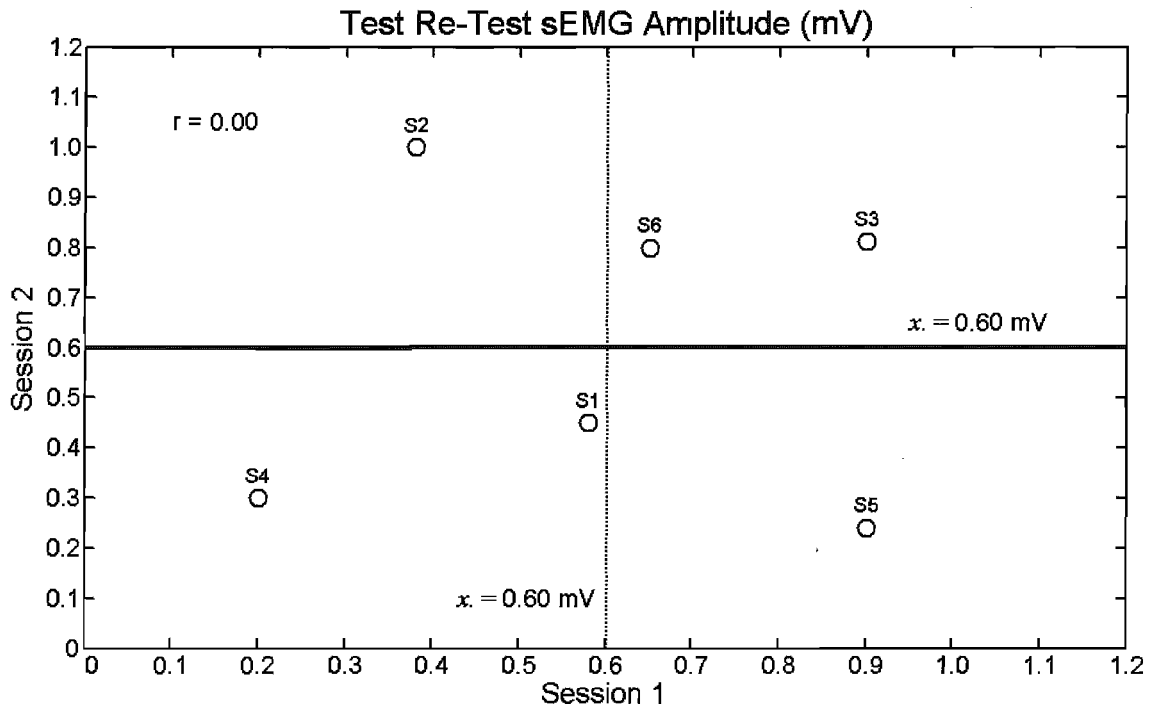


Figure 9. Test-retest reliability for root-mean-square amplitude of surface electromyographic (sEMG) activity. The interclass correlation is  $r = 0.00$ , but there is no difference in means between the first and second test sessions. The regression line is perfectly horizontal.

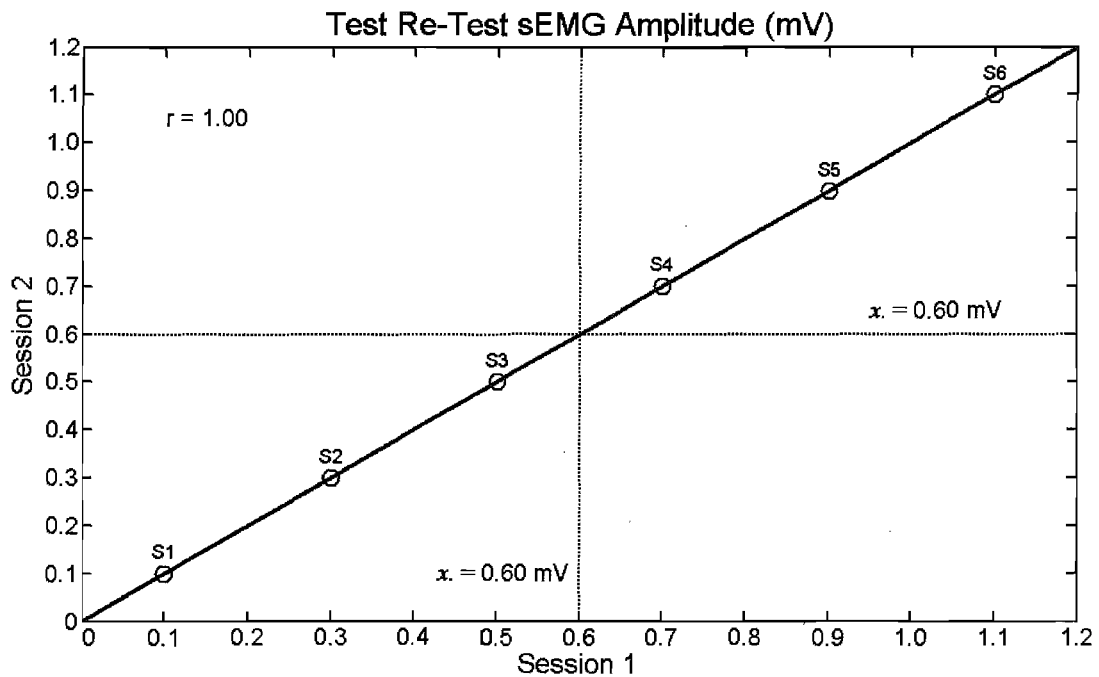


Figure 10. Test-retest reliability for root-mean-square amplitude of surface electromyographic (sEMG) activity. Perfect reliability ( $r = 1.00$ ) is the ideal situation. In this case, each subject reproduces their own magnitude of sEMG activity across both test sessions.

### The Intraclass Correlation Coefficient

The intraclass correlation coefficient is actually a ratio of mean squares extracted from the results of an ANOVA (Portney & Watkins, 1993; Weir, 2005). However, the ANOVA model used to calculate the intraclass correlation coefficient is different from the one generally used to assess stability. Starting from first principles, the observed score ( $x$ ) can be viewed as the sum of the true score ( $T$ ) and associated error ( $e$ ):

$$x = T + e$$

Since the true score for any individual is constant, any fluctuations in the observed score may be attributed to error. The error may be associated with either biological variability or measurement error, or some combination of the two.

Restated in terms of an ANOVA framework, the total variance in a set of observed scores ( $\sigma_x^2$ ) is the sum of the true score variance ( $\sigma_t^2$ ) and error variance ( $\sigma_e^2$ ). The error variance may be partitioned into various sources of measurement error (i.e., days and trials) and explored using the ANOVA model:

$$\sigma_x^2 = \sigma_t^2 + \sigma_e^2$$

The true score variance is not known, but it must be a fixed value because the true score for each individual is constant. Consider the case wherein the scores are perfectly reliable and there is no error variance (Figure 9). The differences between subjects are attributed only to the differences between their true scores, i.e., the true score variance. Similarly, if the true score is a fixed value, then any fluctuation would be due to error variance. The introduction of error to each true score would then obscure the differences between subjects, and decrease the reliability of the measure. Reliability is conceived as a ratio between the true score variance ( $\sigma_t^2$ ) and the total variance ( $\sigma_t^2 + \sigma_e^2$ ). The smaller the error variance, the closer the total variance approaches the true score variance and the ratio ( $R$ ) tends toward unity:

$$R = \frac{\sigma_t^2}{\sigma_t^2 + \sigma_e^2}$$

In this theoretical formula,  $R$  is proportional to the Between Mean Squares where it is inversely proportional to the Error Mean Squares. It should also be noted that in



typical cases,  $0 \leq R \leq 1.0$ ; in some cases where sample size is insufficient,  $R$  may fall below 0, however the value is usually rounded to 0. It is important to remember that the differences between subjects is linked with the true score variance. The true score variance is the between-subjects mean squares (BMS) in the repeated measures ANOVA. Consider a data matrix wherein each row represents a series of scores for a given subject (Table 3). The between-subjects mean squares is based on the differences between rows. If the series of scores for each subject are very “consistent”, the rows will be distinctly different from one another and the between-subjects mean squares will be high. The differences between subjects can then be attributable to the differences between each subjects’ true score. Since the fluctuation of scores within each subject is due to error, differences between scores within rows results in the error variance. The error variance is the within-subjects mean squares (EMS) in the repeated measures ANOVA.

The theoretical formula is translated to a practical method for calculating the intraclass correlation coefficient by using the mean squares from the most basic one-way repeated measures ANOVA model:

$$R \uparrow = \frac{BMS \uparrow}{BMS \uparrow + EMS \downarrow}$$

As subjects become better at reproducing their own score, they become distinctly different from each other, resulting in an increase in the between subjects mean squares. At the same time, the scores within subjects fluctuate less, decreasing the error mean squares. The result is an overall increase in the intraclass correlation coefficient. A criterion measure that is highly reliable has a true score variance that accounts for the greatest percentage of the total variance.

Table 3 – Data matrix for a one-way repeated measure analysis of variance.

Subjects (SS)	Session 1	Session 2	Session 3	Session 4	Session 5
S1	$x_{1,1}$	$x_{1,2}$	$x_{1,3}$	$x_{1,4}$	$x_{1,5}$
S2	$x_{2,1}$	$x_{2,2}$	$x_{2,3}$	$x_{2,4}$	$x_{2,5}$
S3	$x_{3,1}$	$x_{3,2}$	$x_{3,3}$	$x_{3,4}$	$x_{3,5}$
S4	$x_{4,1}$	$x_{4,2}$	$x_{4,3}$	$x_{4,4}$	$x_{4,5}$
S5	$x_{5,1}$	$x_{5,2}$	$x_{5,3}$	$x_{5,4}$	$x_{5,5}$
S6	$x_{6,1}$	$x_{6,2}$	$x_{6,3}$	$x_{6,4}$	$x_{6,5}$

### Reliability of Muscle Fibre Conduction Velocity

Several investigators have struggled with obtaining reliable MFCV measurement. Despite sophisticated signal processing techniques and state of the art instrumentation, the studies conducted thus far have attributed electrode replacement as the single dominant factor in obtaining low reliability (Farina et al., 2004; Merletti et al., 1995; Merletti, et al., 1998). Obtaining highly reliable sEMG measures is certainly possible, even when the electrodes placement is required over multiple sessions. The Electromyographic Kinesiology Laboratory has repeatedly demonstrated high intraclass correlations for sEMG (H-Reflex, M-Wave) produced by electrical stimulation of the peripheral nerve ( $R > 0.90$ ) and voluntary contractions ( $R > 0.80$ ) (Christie et al., 2004; 2005; Calder et al., 2005; 2007).

Merletti et al. (1998) reported a low intraclass coefficient ( $R = 0.36$ ) for MFCV in the tibialis anterior. There were two potential sources of error. One possibility was the use of a large (10 mm) inter-electrode distance. Farina et al. (2005) noted that a 10 mm

inter-electrode distance can decrease the standard deviation of the measurement, but it also can increase the dissimilarity between action potential shapes recorded at electrode pairs. Dissimilar action potential shapes would increase the measurement error in the MFCV estimate. Merletti et al. (1995; 1998) studied the MFCV associated with evoked potentials. It was suggested that error in repositioning the stimulation probes across test sessions could have also contributed to lowering the reliability.

## CHAPTER III

### METHODS

#### Subjects

Forty college-age individuals (21 males and 19 females) participated in this study. Only right leg dominant individuals with a body mass index under 25 were used. Participants signed an informed consent document (REB #02-283) that complied with Brock University Research Ethics Board (Appendix A). All testing occurred within a Faraday cage housed in the Electromyographic Kinesiology Laboratory in Welch Hall, room 18.

#### Preliminary Test Procedures

Each participant was introduced to the lab prior to the first test session. They were verbally acquainted with the testing device, associated instrumentation, and the demands of the task at that time. Each participant was then asked to complete an informed consent document and the PAR-Q (Appendix B). The informed consent document also included typed written instructions that reinforce the requirements of testing. The following anthropometric measurements were also obtained: height, weight, length and girth of the right lower leg (Appendix C).

#### Electromyography

The participant then lied down supine on a gurney for motor point determination. A constant-current (150 mA) source was used to find the motor point using the lowest

possible voltage. The cathode and anode electrodes were connected in series with an isolation unit (Grass Telefactor SUI8, Astro-Med, Inc., West Warwick, RI) and a stimulator (Grass Telefactor S88, Astro-Med Inc., West Warwick, RI) to deliver a square-wave pulse, 1 msec in duration at a rate of 10 pps (Calder et al., 2005). A self-adhesive anode (Pals Plus, 5.0 cm, Axelgaard, Fallbrook, CA) was then secured on the gastrocnemius while a stainless-steel cathode probe (3 mm) was used to systematically explore the skin surface of the tibialis anterior until a barely perceptible muscle twitch was observed beneath the skin (Roy et al., 1986). Once the motor point had been determined, the stimulus intensity was then increased to produce a clearly visible, localized muscle twitch at 1 pps. The orientation of the muscle fibres was extrapolated along a “best fit line” and marked with indelible ink. The procedure was repeated before the recording session on each test day.

The lower leg was shaved, abraded (NuPrep®, Weaver and Company, Aurora, CO), and cleansed with alcohol to reduce the impedance at the skin-electrode interface. The electrode was then prepared with two-sided tape and electrolyte gel (Signa Gel®, Parker Laboratories, Inc., Fairfield, NJ). The recording electrodes consisted of four stainless-steel tubular surfaces, each 1 mm in diameter and 10 mm long, mounted within a rigid plastic structure with a centre-to-centre interelectrode distance of 5 mm. The electrodes were configured to yield three sets of bipolar signals via single differentiation. Initial placement was in line with the muscle fibres, 5 mm below the most distal motor point towards the distal tendon. The ground electrode (CF5000, Axelgaard, Fallbrook, CA) was located on the lateral malleolus.

Once the electrodes were secured to the skin surface, the impedance was assessed (Grass EZM Electrode Impedance Meter, Astro-Med Inc., Warwick, RI) to ensure that it was lower than 10 k $\Omega$  and no further preparation was necessary. Temperature was also recorded pre-recording and post-recording to ensure that the value did not significantly differ from the previous visit. The location and orientation of the electrode was then traced with an indelible ink marker. Participants were also sent home with an indelible ink marker to maintain the electrode outline between testing sessions. Since the sessions were only 48 hours apart, it was not anticipated that fading would occur to the degree that the electrode cannot be placed in its previous location (Calder et al., 2005). The procedure for identification of the motor point was repeated before each testing session. And the electrode was replaced in the outline trace from the previous session. This placement was merely a guideline that provided a means of minimizing the procedure associated with optimizing electrode placement.

### Force

Participants next sat in a testing chair that was incorporated into a jig designed to isolate the action of the dorsiflexors in an isometric contraction (Figure 11). The position of the testing chair within the jig could be adjusted so that the hip and knee joints were at 90 degrees. A load-cell (JR3 Inc., Woodland, CA) was secured beneath a foot-plate so that the ankle joint was placed at 110° (slight plantar flexion). Dorsiflexion forces were applied perpendicular to the load cell through an adjustable mount. A metal bar was placed over the top of the bare foot at the 5th metatarsal to secure the foot to the foot-plate/load-cell assembly. Participants performed dorsiflexion against the metal bar to transmit forces to the load-cell. A foam cushion between the metal bar and the top of the

foot ensured that the participant was comfortable and the foot was secure. Belts were used to help stabilize the participant within the chair.



Figure 11. Subject test position and experimental set-up. The load cell was at the bottom of the foot-plate. There were separate belts for the thighs and waist to secure the legs and hips, respectively.

### Measurement Schedule

There were a total of 3 days of testing with at least 48 hours between each session. The maximum time between testing sessions was 7 days. The following procedures were performed on each test day. Participants sat in the testing chair located

within the Faraday cage, and they once again read type-written instructions of the demands of the testing while skin temperature (Electrotherm TM99A - Cooper Instrument Corp., Middlefield, CT) was taken. Prior to the first recordings, electrically evoked potentials were elicited to ensure that electrode placement maximized action potential shape similarity between channels, as well as delay between signals (Farina et al., 2004). At this point, there was a check to ensure that the muscle fiber conduction velocity values for the subsequent voluntary contractions fell within physiological range between 2 and 13 m/s (Hunter et al., 1987). If the value did not fall within an acceptable range, or any of the conditions previously established were unsatisfactory, repositioning of the electrode continued until a satisfactory position was obtained.

Participants then performed three maximal isometric contractions of the dorsiflexors. The procedures for determining the dorsiflexion MVC were based on Baratta et al. (1998). The contractions were 5-seconds in duration with a 3-minute rest interval. A target was presented as a horizontal line on an oscilloscope (Hitachi, VC-6525) placed in front of the subject. The target was a voltage that represented 110% of the mean peak value of the previous three contractions. If participants were able to reach the target line during the fourth trial, the MVC value was updated. Electrode impedance and skin temperature were once again assessed following completion of the last trial.

### Data Recording

The three single differentiated sEMG signals and 2 double differentiated were band-pass between 10 and 1000 Hz 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). The force and sEMG signals were sent to the connector block (BNC-2110, National Instruments, Austin, TX) associated with the analogue-to-digital conversion board. All signals were sampled at 5 kHz using a computer-based data acquisition system (DASYLab, DASYTEC National Instruments, Amherst, NH). The datum was stored on a Celeron PC for off-line processing (Dell, Round Rock, TX).

### Data Reduction

The detection surfaces and the associated sEMG signals from a representative subject are show in Figure 10. The sEMG signals were normalized by taking the z-score value of each datum to ensure each channel had zero mean and unit variance (Figure 13). Normalization is designed to minimize the impact of amplitude differences on the peak magnitude of the cross-correlation function (Hunter et al., 1987). The force and sEMG signals were analyzed from a 500 msec window, immediately prior to the middle of the contraction, where the force-time curve was most stable (Farina et al., 2004). The sEMG signals were up-sampled to 25 kHz prior to calculating the cross-correlation function to increase the precision of measurement (Farina & Merletti, 2004). All cross-correlation functions had a coefficient of 0.80 or higher as recommended by Arendt-Nielsen and Zwarts (1989) (Figure 14). Muscle fibre conduction velocity was calculated based on the delay-time identified by the cross correlation function and the known interelectrode distance of 0.5 cm. This was done for the first two single-differential recordings (SD1 and SD2) and for the double-differential signals (DD1 and DD2), which were created off-line (Lowery et al., 2002).

The root-mean-square (RMS) amplitude and the frequency of the mean power (MPF) were also calculated for SD2 from the same data window according to the formulae outlined by Kamen and Gabriel (2009). These measures are well established in the literature and offer a basis of comparison for the general reliability of the sEMG methodology.

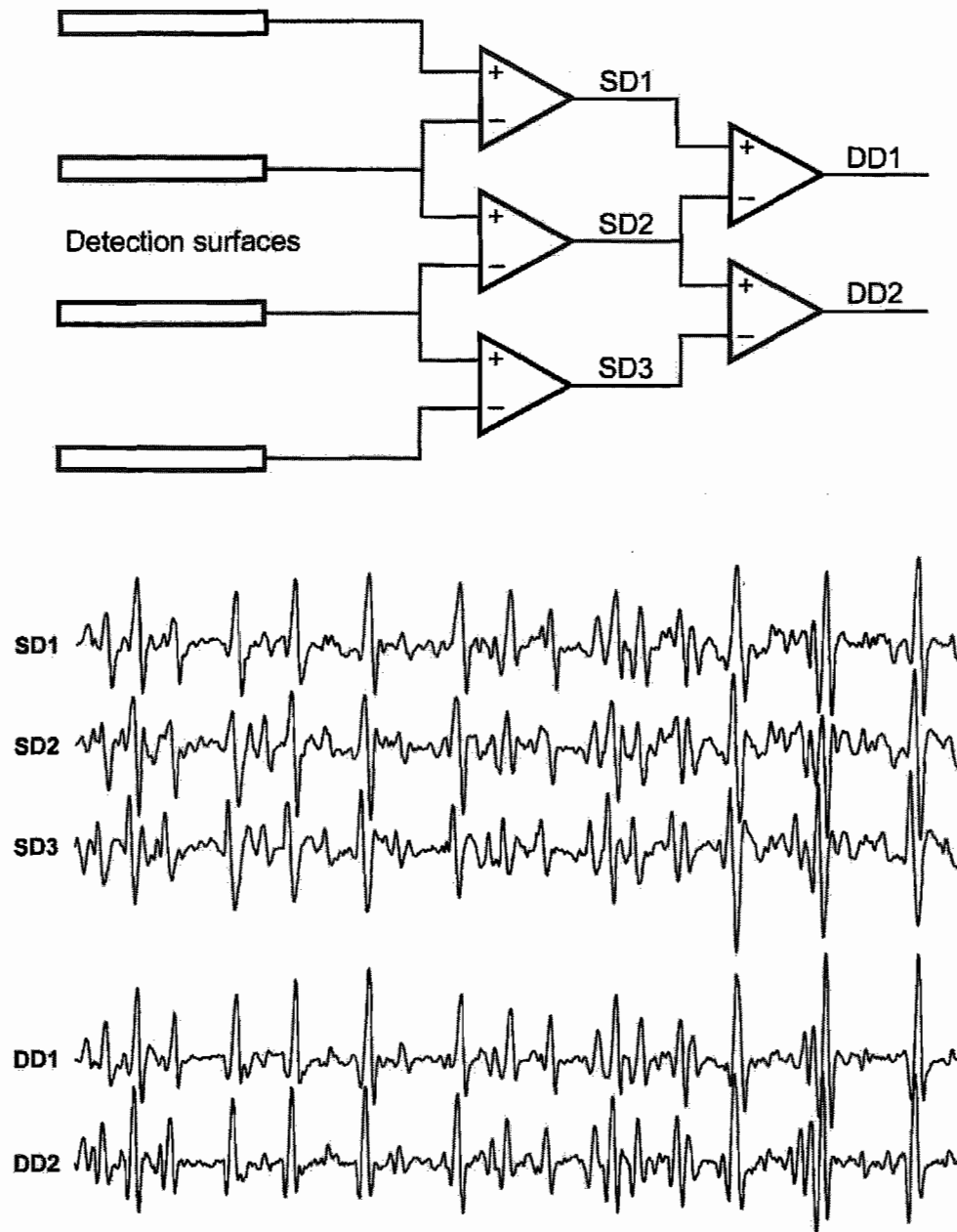


Figure 12. The surface electromyographic activity associated with a 30 percent maximal voluntary contraction condition as recorded by both single and double differential detection surface, SD and DD, respectively. The double differential signals were created by software, off-line. Notice that the additional spatial filtering of the DD signals enhances the identification of individual motor unit action potentials.

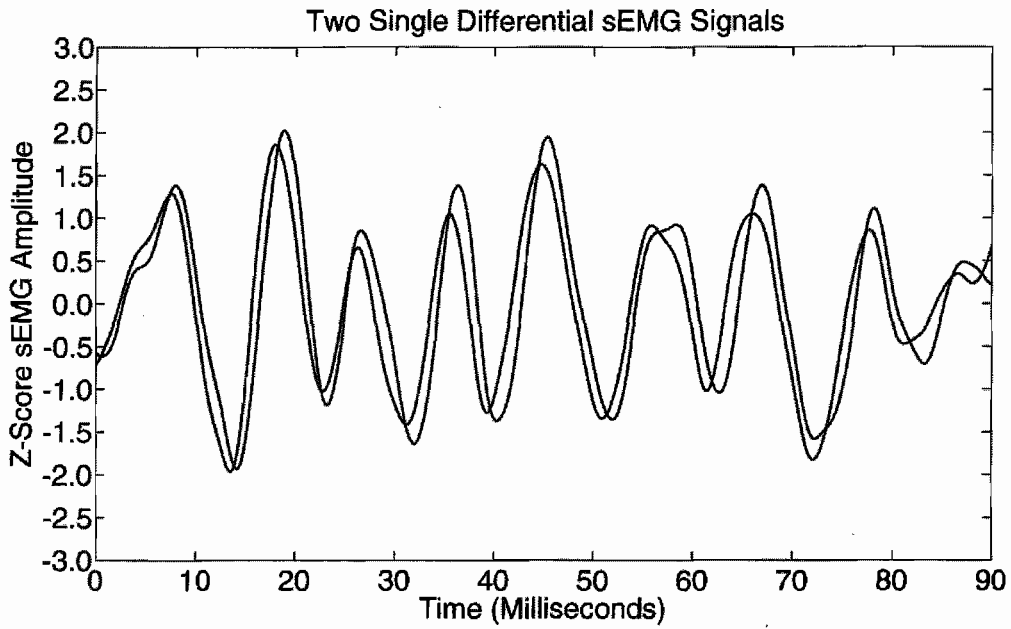


Figure 13. The z-score amplitude for two single differential surface electromyographic (sEMG) signals from a representative subject.

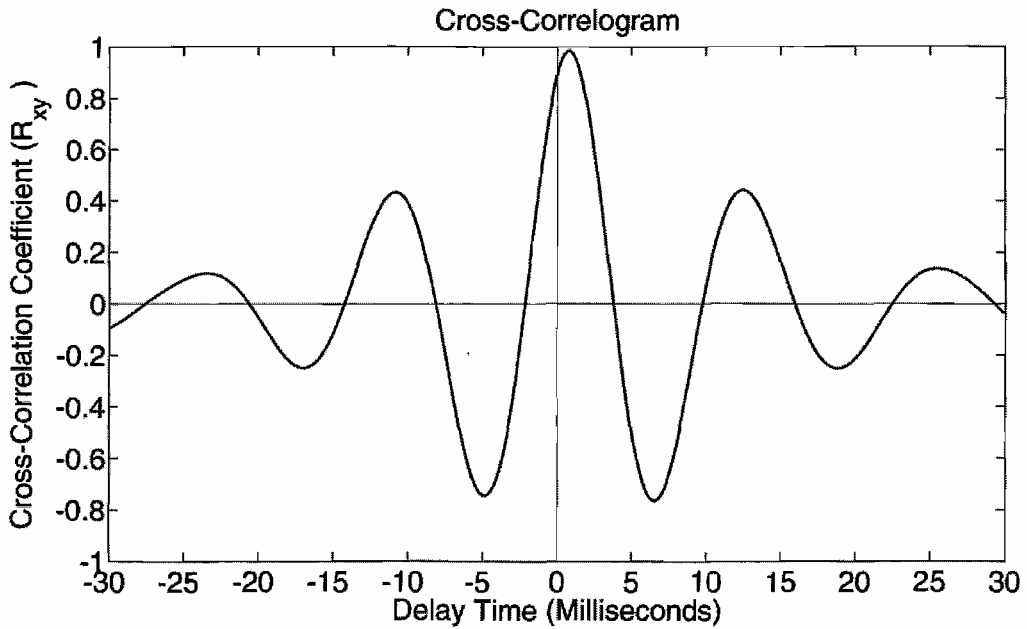


Figure 14. The cross-correlation function for the two single differential surface electromyographic signals depicted in Figure 11.

## Statistical Analysis

Intraclass correlational analysis of variance for reliability estimation, as originally conceived by Feldt and McKee (1958), involved the use of a fully nested ANOVA model with two dimensions. In this scheme, “Test Days” represents the first dimension while “Subjects” is the second dimension. The repeated measurements (trials) on each subject within each test day constitute a “Within-Cells” replication of measures. The error term in this model is the Within-Cells error variance. Feldt and McKee (1958) considered the trial-to-trial variance Within-Cells as random measurement error in the experimental design.

The model used for the analysis reported in Table 3 was used by Kroll (1962; 1963a; 1963b) for test-retest reliability estimation. The mean squares from the ANOVA model presented in Table 4 were then used to calculate the sample estimates for the variance components given in Table 5. The intraclass correlation coefficient described by Feldt and McKee (1958) is defined as the ratio of the true score variance ( $\sigma_{true}^2$ ) to the sum of the variance due to true scores and error. The error variance is further partitioned into error attributed to both day-to-day ( $\sigma_{e_2}^2$ ) and trial-to-trial ( $\sigma_{e_1}^2$ ) sources. The values derived from the equations in Table 4 permit the reliability of a mean from any combination of days ( $a$ ) and trials ( $n$ ) within a testing period to be estimated using the following formula for the intraclass correlation coefficient:

$$R = \frac{\sigma_{true}^2}{\sigma_{true}^2 + \frac{\sigma_{e_2}^2}{a} + \frac{\sigma_{e_1}^2}{a \cdot n}}$$

Table 4 – *Intraclass correlational analysis of variance (nested model). A representative analysis of variance (ANOVA) table for force generated during the maximal isometric actions of the dorsiflexors.*

Source	<i>df</i>	E (MS) Model II	MS
Subjects	$N - 1 = 39$	$\sigma_{e_1}^2 + n\sigma_{e_2}^2 + a\sigma_t^2$	34212.082 ( $MS_S$ )
Days(Subjects)	$N(a - 1) = 80$	$\sigma_{e_1}^2 + n\sigma_{e_2}^2$	573.631 ( $MS_{D:S}$ )
Within-Cells	$aN(n - 1) = 240$	$\sigma_{e_1}^2$	76.719 ( $MS_{WC}$ )
Total	$aNn - 1 = 359$		

N = subjects (40); n = trials on each test day (3); a = test days (3)

Table 5 – *Components of variance necessary to calculate the intraclass correlation coefficient.*

Component	Equation
$\sigma_{e_1}^2 - \text{Trials}$	$MS_{WC}$
$\sigma_{e_2}^2 - \text{Days}$	$\frac{MS_{D:S} - MS_{WC}}{a}$
$\sigma_{true}^2 - \text{Subjects}$	$\frac{MS_S - MS_{D:S}}{a \cdot n}$

N = number of subjects; n = number of trials each day; a = number of test days

A second (companion) ANOVA model was used to examine the “stability” of the means across test days. This ANOVA model still has two factors (Days × Subjects). The repeated measurements (trials) on each subject in each day constitute a “within-cells” replication of measures (Lindquist, 1956; Feldt & McKee, 1958; Hetherington, 1973). The mean squares for the second ANOVA model for force during the maximal isometric actions of the dorsiflexors are presented in Table 6 to see the relationship between the two ANOVA models used in this thesis. These two specific ANOVA models are therefore supposed to be used together to allow a diagnostic approach to the sources of error and their impact upon reliability (Feldt & McKee, 1958; Kroll, 1962; 1963a; 1963b). A measure must therefore exhibit both “consistency” and “stability” to be considered reliable.

Table 6 – *Analysis of variance model used to evaluate the stability of means across test days. A representative analysis of variance (ANOVA) table for force generated during the maximal isometric actions of the dorsiflexors.*

Source	df	E (MS) Model II	MS
Days	$a - 1 = 2$	$\sigma_e^2 + n\sigma_{aN}^2 + nN\sigma_a^2$	2132.466 ( $MS_D$ )
Subjects	$N - 1 = 39$	$\sigma_e^2 + n\sigma_{aN}^2 + na\sigma_N^2$	34212.082 ( $MS_S$ )
Day × Subjects	$(N - 1)(a - 1) = 78$	$\sigma_e^2 + n\sigma_{aN}^2$	533.661 ( $MS_{DS}$ )
Within Cell	$aN(n - 1) = 240$	$\sigma_e^2$	76.719 ( $MS_{WC}$ )
Total	$aNn - 1 = 359$		

N = number of subjects; n = number of trials each day; a = number of test days

For the purposes of this thesis, the scale used by Bartko (1966) was used to evaluate the quality of the intraclass correlation coefficients in terms of reliability. A measure with an intraclass coefficient in the range of 0.80-1.00 was considered to have "excellent reliability" while a range of 0.60 to 0.80 was deemed to be "good reliability". It is possible however, to have very stable means across test days and the measure have an intraclass correlation coefficient below 0.60 due to a "range of talent" effect (Kroll, 1967). That is, the differences between subjects are difficult to detect in a naturally homogeneous group. The standard error of measurement (SEM) served as an additional diagnostic tool should a limited range of scores artificially deflate the intraclass correlation coefficient (Calder et al., 2008). The SEM has the benefit of providing an assessment of reliability "within subjects" in terms of the actual units of measurement (Weir, 2005). The standard error of measurement was calculated as follows:

$$SEM = SD\sqrt{1 - R}$$

where the SD is the standard deviation of the scores as determined from the nested ANOVA model. The standard deviation of the scores will be derived from the total sum of squares error ( $SS_{total}$ ):

$$SD = \sqrt{\frac{SS_{total}}{N - 1}}$$



## CHAPTER IV

### RESULTS

The primary objective of this study was to examine the reliability of muscle fibre conduction velocity (MFCV) in the tibialis anterior during isometric voluntary contractions. This was done at 100 percent of maximal voluntary contraction. The variable of MFCV was assessed by single- and double differential electrode configurations involving the cross-correlation technique.

#### Subject Characteristics

The means and standard deviations for the physical characteristics of the participants (21 males and 19 females) are presented in Table 7.

#### Data Screening

The data were pre-screened for outliers during the data collection phase. Verification of electrode placement by stimulating the peripheral nerve and evaluating the separation of peaks in the compound muscle actions potentials associated with the different detection surfaces ensured that range of values was between 2.5 and 10.5 m/s. One subject had a maximum muscle fiber conduction velocity value (13.8 m/s) that was just outside the upper limit of 13 m/s (Hunter et al., 1987). This value was retained in the analysis as it occurred once on two separate test sessions and the ultimate goal of the thesis is to document the error of measurement.

Table 7 – Means (*M*) and standard deviations (*SD*) for the physical characteristics of the participants.

Physical Characteristic	M ± SD
Age (years)	23.55 ± 2.85
Height (m)	1.72 ± 0.12
Weight (kg)	68.47 ± 13.95
Body Mass Index (kg·m <sup>-2</sup> )	22.90 ± 2.26
Foot Length (cm)	26.26 ± 2.76
Leg Length (cm)	43.46 ± 4.65
Leg Girth (cm)	38.86 ± 2.80

### Statistical Assumptions

Application of the intraclass correlation coefficient requires the data exhibit congruency with the mathematical model upon which it is based. The data must therefore conform to the standard univariate assumptions that underlie the use of repeated measures analysis of variance. Violation of these assumptions results in a lower estimate of reliability (Kroll, 1962). There are two sets of statistical assumptions: those associated with the *F*-test itself and those associated with the experimental design model. Since the two sets of assumptions basically overlap (Kirk, 1995), they will be reviewed and evaluated together for the sake of simplicity.

A nested ANOVA was used to generate the mean squares necessary for the calculation of the intraclass correlation coefficient to evaluate the **consistency** of scores within subjects:

$$X_{k(ij)} = \mu + \alpha_i + \beta_{j(i)} + e_{k(ij)}$$

The first assumption is that an observation can be thought of as the simple sum of four components where  $X_{k(ij)}$  is the value of the dependent variable,  $\mu$  is the grand mean of the dependent variable,  $\alpha_i$  is the differences between subjects,  $\beta_{j(i)}$  is the differences between test days nested within subjects, and  $e_{k(ij)}$  is the differences between trials nested within each test day, nested within subjects. The error term in this model is the within-cells error variance.

The second assumption is that  $e_{k(ij)}$  is (a) independent of all other  $e_{k(ij)}$ 's and (b) normally distributed within the population, with (c) a mean equal to zero, and (d) variance equal to  $\sigma_e^2$ . Because  $e_{k(ij)}$  is the only source of variation in the linear model, the normality assumption can be tested by evaluating the raw scores (Kirk, 1995). All measures examined in this study exhibited skewness values less than 2 and kurtosis values less than 3 and were therefore deemed to be normally distributed. Glass, Peckham, and Sanders (1972) have demonstrated that the analysis of variance is robust to mild departures from normality for balanced designs with moderate sample sizes. Robust refers to the fact that the probability of type I and type II errors for the  $F$ -test remain relatively unchanged.

The assumption of independence of errors is no longer applicable in repeated measures ANOVA because the same participants produce multiple scores; the errors are by necessity correlated. In practice, the assumption of the independence of errors is replaced by the assumption of sphericity (Tabachnick & Fidell, 2007). Sphericity is

sometimes called the “homogeneity-of-variances-of-differences” assumption, because the variance of the difference scores between any two levels of a within-subjects factor is supposed to have the same magnitude regardless of which two levels are chosen. There are three levels of the independent variable in this thesis, test days one through three. The difference scores are calculated between test days one and two, test days two and three, and test days one and three. The sphericity assumption assumes that the variance of these difference scores is not significantly different. The SAS® program allows the Mauchly's sphericity test to be conducted prior to the within-subjects tests. If the Chi-square approximation associated with the Mauchly's test has an associated probability value less than the selected alpha level, the sphericity assumption has been violated. Sphericity was however upheld for each of the criterion measures evaluated in this study. Correction of the degrees of freedom for the *F*-tests based on either the Greenhouse-Geisser Epsilon (G-G) or the Huynh-Feldt Epsilon (H-F) was deemed unnecessary.

There is an additional assumption, the stricter assumption of compound symmetry. Compound symmetry involves two conditions. The first condition is homogeneity of variances as in the case of non-repeated measures ANOVA. This means that there is equal variability of scores across treatment levels (i.e., Days). Given there are the same number of observations at each treatment level and the data come from the same subjects, the homogeneity of variance assumption is not likely to be violated (Cohen, 2008). The second condition concerns the degree to which subjects maintain their relative standing within the various conditions as defined by the repeated factor (i.e., Days). The consistency observed for any pair of treatment conditions is the same for all

possible pairs of treatment conditions, so that there is homogeneity of the covariance between pairs of treatment conditions (Keppel, 1973). This has particular implications for the additivity assumption described below. Homogeneity of the covariance is usually violated with repeated measures ANOVA, and it is not a "necessary" condition. A violation of the stricter condition of compound symmetry does not necessarily mean that the sphericity assumption has been violated, which is a necessary condition (Hayes, 1994).

A second repeated measures ANOVA was used to evaluate **stability** of the means across test days:

$$X_{ijk} = \mu + \alpha_i + \beta_j + \alpha\beta_{ij} + e_{ijk}$$

This second model is by necessity complimentary to the nested ANOVA present above; it still has two factors but incorporates the Subjects  $\times$  Days interaction term ( $\alpha\beta_{ij}$ ). The repeated measurements (trials) on each subject in each day constitute a "within-cells" replication of measures and is the error term ( $e_{ijk}$ ).

All of the assumptions outlined above hold true for this second ANOVA model. There is one additional assumption that applies to the interaction term, additivity. It is assumed that any variation in the difference between test days is due to error variation. It is possible that the effect of test days is different between subjects, thus there is truly an interaction between subjects and days and not error variation. Some of what should be considered random error when calculating the interaction term is "systematic error" due to subjects responding differently across test days. The additivity assumption is that there

is no interaction between the subjects and test days that is not error (interactions are multiplicative or "nonadditive"). For a random effects ANOVA model as employed in this study, non-additivity results in a less powerful *F*-test for the main effects. Non-additivity also implies that the homogeneity of the covariance is violated. However, this was not the case for the present study. Glass, Peckham and Sanders (1972) further indicate that non-additivity should be of no concern.

### Stability and Consistency

Force. Maximal isometric dorsiflexion force is being presented first because it represents a nearly ideal example of what constitutes a highly reliable measure. There was a significant ( $p < 0.01$ ) increase from  $173.06 \pm 60.75$  N to  $179.40 \pm 64.42$  N across the three test sessions (Table 8). This increase was small (3.9%) and the differences between means across test days accounted for only 4.16% of the total variance (Table 9). Figure 15 depicts the mean and standard deviation of the scores for each subject. The spread of force scores for each subject was generally grouped tightly around its own mean. Equally important, the low spread of scores within each subject resulted in little overlap of the forces scores between different subjects so that the between subjects variance (true score variance) was high ( $\approx 94\%$ ). If the true score variance accounts for the greatest proportion of the variance in the ANOVA model, the intraclass correlation coefficient is going to be high. The overall result was an intraclass correlation coefficient of 0.98.

Root-Mean-Square Amplitude. Table 8 shows that the magnitude of sEMG activity increased from  $197.11 \pm 92.68$   $\mu$ V to  $205.83 \pm 93.10$   $\mu$ V. While statistically significant ( $p < 0.01$ ), the magnitude of the increase was small (4.4%). The error variance

due to trials (8.02%) and days (19.30%) together accounted for a total of 27.32%, so that the remaining true score variance was 78.68% (Table 9). Figure 16 illustrates the mean and standard deviation of scores for each subject. The grand mean was 203.42  $\mu$ V with standard error of measurement (SEM) of 90.17  $\mu$ V. In general, the spread of scores for each subject appears to be quite moderate. However, relative to the overall range of mean scores for the sample, the differences between subjects was not as obvious as it was for maximal isometric dorsiflexion force. The intraclass reliability correlation coefficient was 0.91; while still considered to excellent, it is lower than that for maximal isometric dorsiflexion force.

Frequency of Mean Power. Table 8 shows that the frequency of sEMG activity increased from  $118.78 \pm 28.00$  Hz to  $123.33 \pm 29.23$  Hz. The small (3.8%) increase was statistically significant ( $p < 0.01$ ). The error variance due to trials (6.03%) and days (10.76%) together accounted for a total of 16.79% while the true score variance was 83.21% (Table 9). Figure 17 illustrates the mean and standard deviation of scores for each subject. The grand mean was 121.10 Hz with standard error of measurement (SEM) of 19.64. In general, the spread of scores for each subject appears to be quite low relative to the overall range of mean scores for the sample. Thus, similar to maximal isometric dorsiflexion force, the differences between subjects is more obvious than it was for root-mean-square amplitude. As a result, the intraclass reliability correlation coefficient was 0.95.

Muscle Fibre Conduction Velocity. Table 8 shows that muscle fibre conduction velocity for single differential electrode detection decreased from  $5.19 \pm 1.31$  m/s to 5.05

$\pm 1.39$  m/s. The small (2.7%) decrease in means was statistically significant ( $p < 0.01$ ). The error variance due to trials (5.37%) and days (34.96%) together accounted for a total of 40.33% while the true score variance was 59.67% (Table 9). Figure 18 illustrates the mean and standard deviation of scores for each subject. The grand mean was 5.14 m/s with standard error of measurement (SEM) of 1.65 m/s. In general, the spread of scores for each subject appears to be high relative to the overall range of mean scores for the sample, so that it is more difficult to visually discern the differences between subjects. In this case, the standard error of measurement is not high but the range of mean scores for the majority of subjects is low. Thus, the sample is relatively more homogeneous with respect to this particular measure. That is, the mean scores for subjects were close to the group mean, resulting in a kurtosis of 1.51 for this distribution. However, the true score variance was still the largest proportion of the total variance; the intraclass correlation coefficient was 0.83, which is still considered excellent.

There was an adverse interaction between the standard error of measurement and a homogeneous sample of scores for double differential electrode detection. Table 8 shows that the means for muscle fibre conduction velocity detected by the double differential electrode fluctuated between  $4.93 \pm 1.36$  m/s and  $5.26 \pm 2.05$  m/s for a significant main effect for test days ( $p < 0.01$ ). The error variance due to trials (6.02%) and days (57.32%) accounted for the large majority of the total variance (63.34%) while the true score variance was only 36.66% (Table 9). The grand mean was 5.06 m/s and the standard error of measurement was 3.04 m/s. Inspection of Figure 19 reveals that the mean scores for subjects were very close to the group mean, resulting in a kurtosis value of 2.89. The between subjects variance (true score variance) is low because the limited



range of mean scores coupled with a large standard error of measurement, made it very difficult to differentiate between individual subjects. The intraclass correlation coefficient was 0.65.

Table 8 – Analysis of variance for 100% maximal voluntary contractions. The units for Force are in Newtons (N), the units for Conduction Velocity (CV) are in metres per second ( $m \cdot s^{-1}$ ), the units for Mean Power Frequency (MPF) are in Hertz (Hz) and the units for Root Mean Square (RMS) are in microvolts ( $\mu V$ ).

		<b>Force</b> (N)	<b>RMS</b> ( $\mu V$ )	<b>MPF</b> (Hz)	<b>Single CV</b> ( $m \cdot sec^{-1}$ )	<b>Double CV</b> ( $m \cdot sec^{-1}$ )
<b>Test Day</b>		M $\pm$ SD	M $\pm$ SD	M $\pm$ SD	M $\pm$ SD	M $\pm$ SD
<b>1</b>		173.06 $\pm$ 60.75	197.11 $\pm$ 92.68	118.78 $\pm$ 28.00	5.19 $\pm$ 1.31	4.93 $\pm$ 1.36
<b>2</b>		180.56 $\pm$ 62.28	207.30 $\pm$ 108.58	121.03 $\pm$ 90.80	5.18 $\pm$ 1.26	5.26 $\pm$ 2.05
<b>3</b>		179.40 $\pm$ 64.42	205.83 $\pm$ 93.10	123.33 $\pm$ 29.23	5.05 $\pm$ 1.39	4.99 $\pm$ 1.60
Percent Change		6.34 (3.9%)	8.71 (4.4%)	4.55 (3.8%)	0.13 (2.7%)	0.055 (1.2%)
Difference of Means						
Day 1-2		-7.50	-10.18	-2.25	0.00049	-0.32
Day 2-3		1.16	1.47	-2.29	0.13	0.27
Day 1-3		-6.34	-8.71	-4.55	0.13	-0.055
Days $\times$ Subjects	78	533.66*	0.0065*	329.00*	1.98*	5.21*
Within Cells	240					

\* Significant at the 0.01 probability level

Table 9 – Intraclass correlation analysis of variance for 100% maximal voluntary contractions. Below are the mean squares (MS), variance components, and the mean, standard error of measurement (SEM), and the resultant intraclass correlation coefficients (R) for Force, Conduction Velocity (CV), Mean Power Frequency (MPF) and Root Mean Square (RMS).

Source	df	Force	RMS	MPF	Single CV	Double CV
Subjects	39	34212.00	70697.63	6908.00	11.44	14.74
Day(Subjects)	80	573.63	6.472.56	336.30	1.95	5.16
Within Cell	240					
$(\sigma_{e1}^2 - \text{Trials})$		76.72 (1.93%)	787.03 (8.02%)	52.95 (6.03%)	0.09 (5.37%)	0.17 (6.02%)
$\sigma_{e2}^2 - \text{Days}$		165.64 (4.16%)	1895.17 (19.30%)	94.45 (10.76%)	0.62 (34.96%)	1.66 (57.32%)
$\sigma_t^2 - \text{True}$		3737.61 (93.91%)	7136.12 (72.68%)	730.19 (83.21%)	1.05 (59.67%)	1.06 (36.66%)
Grand Mean		177.67	203.42	121.05	5.14	5.06
SEM		24.52	90.17	19.64	1.65	3.04
R		0.98	0.91	0.95	0.83	0.65

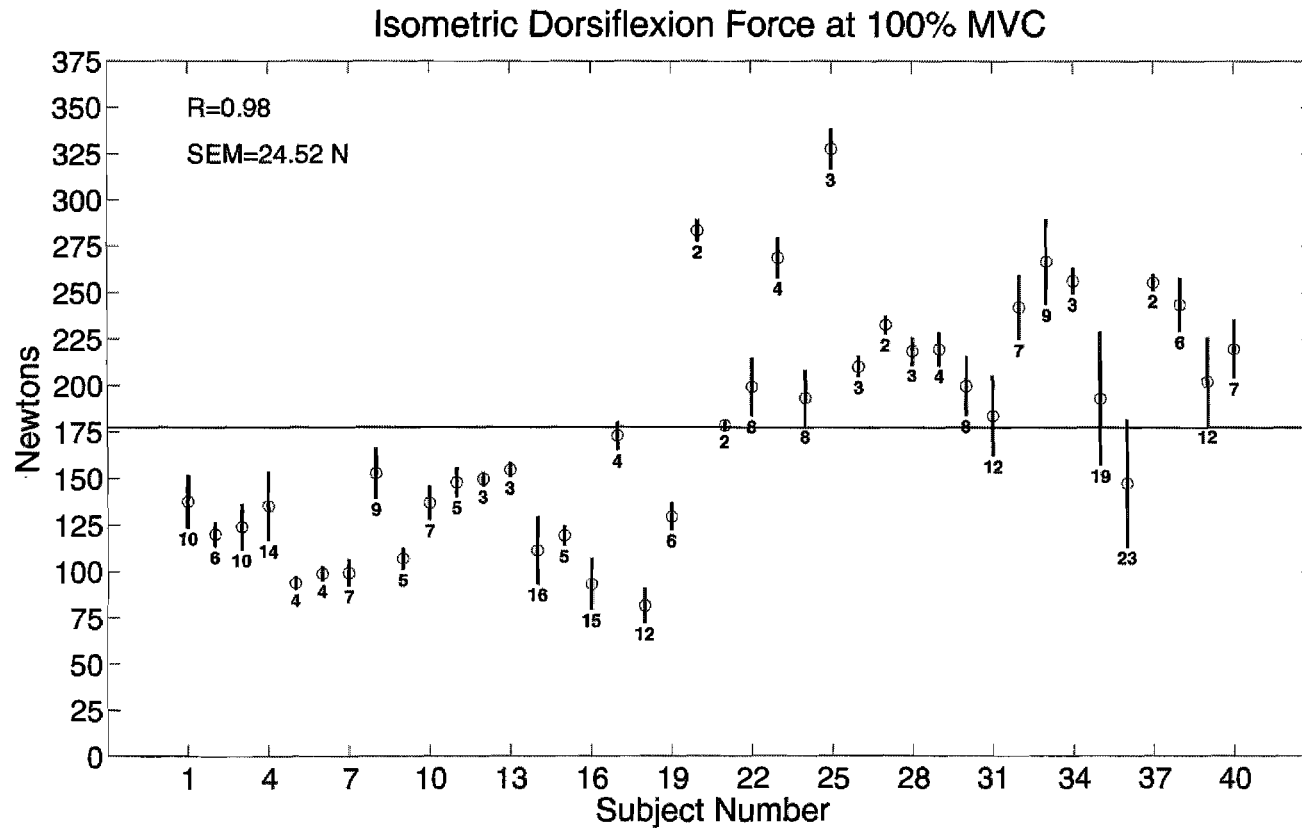


Figure 15. The means (circles) and standard deviations (vertical bars) for maximal isometric dorsiflexion force for each subject. The number below the vertical bar is the coefficient of variation for the individual subject. Each subject has three respective values accounted for.

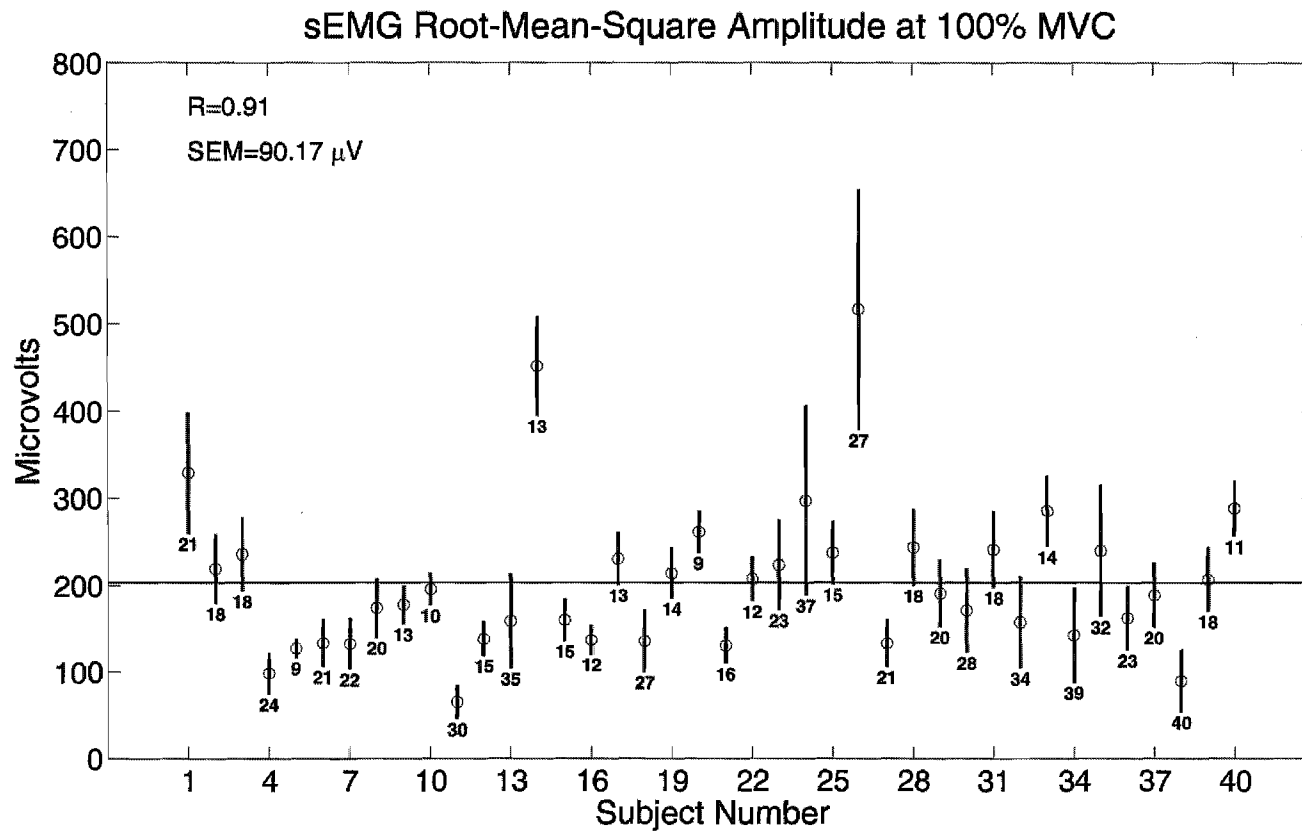


Figure 16. The means (circles) and standard deviations (vertical bars) for root-mean-square amplitude for surface electromyographic (sEMG) activity for each subject. The number below the vertical bar is the coefficient of variation for the individual subject. Each subject has three respective values accounted for.

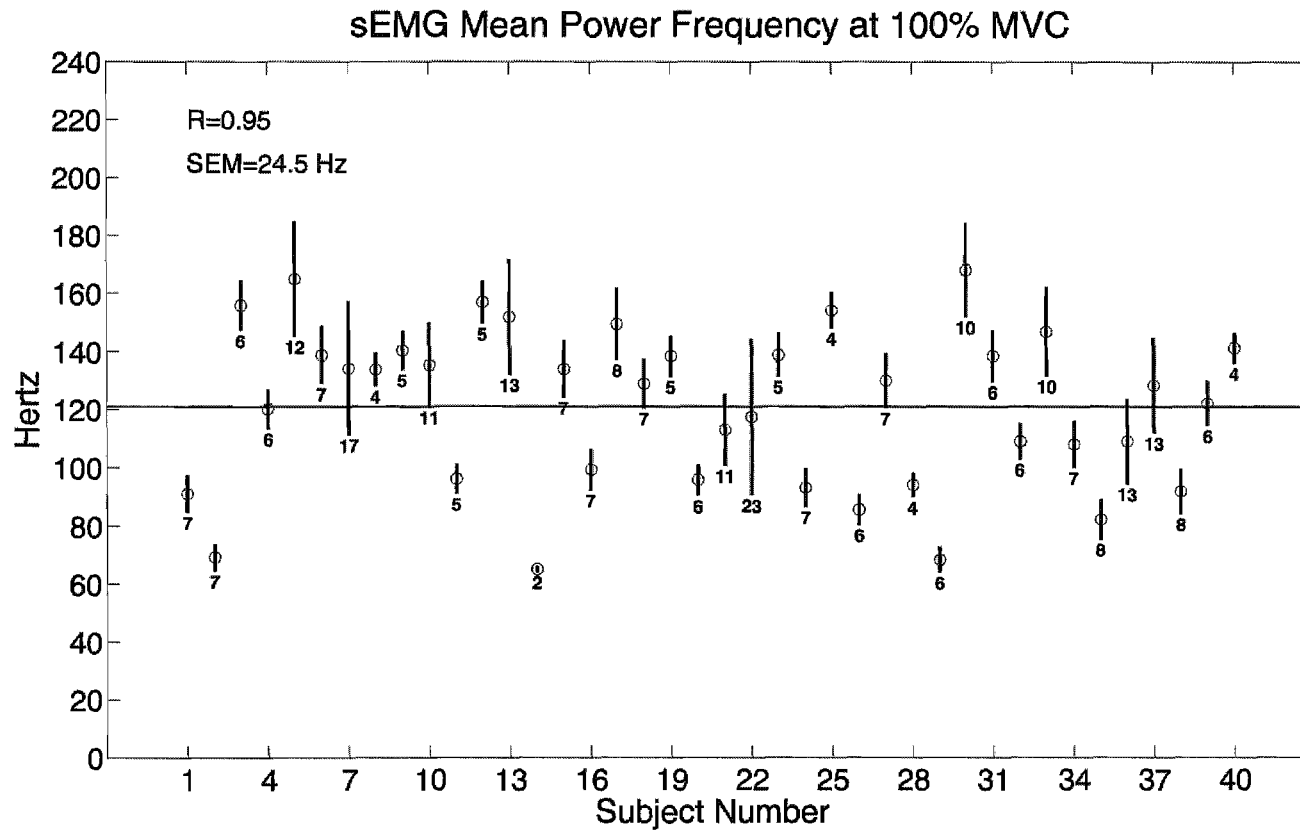


Figure 17. The means (circles) and standard deviations (vertical bars) for the mean power frequency for surface electromyographic (sEMG) activity for each subject. The number below the vertical bar is the coefficient of variation for the individual subject. Each subject has three respective values accounted for.

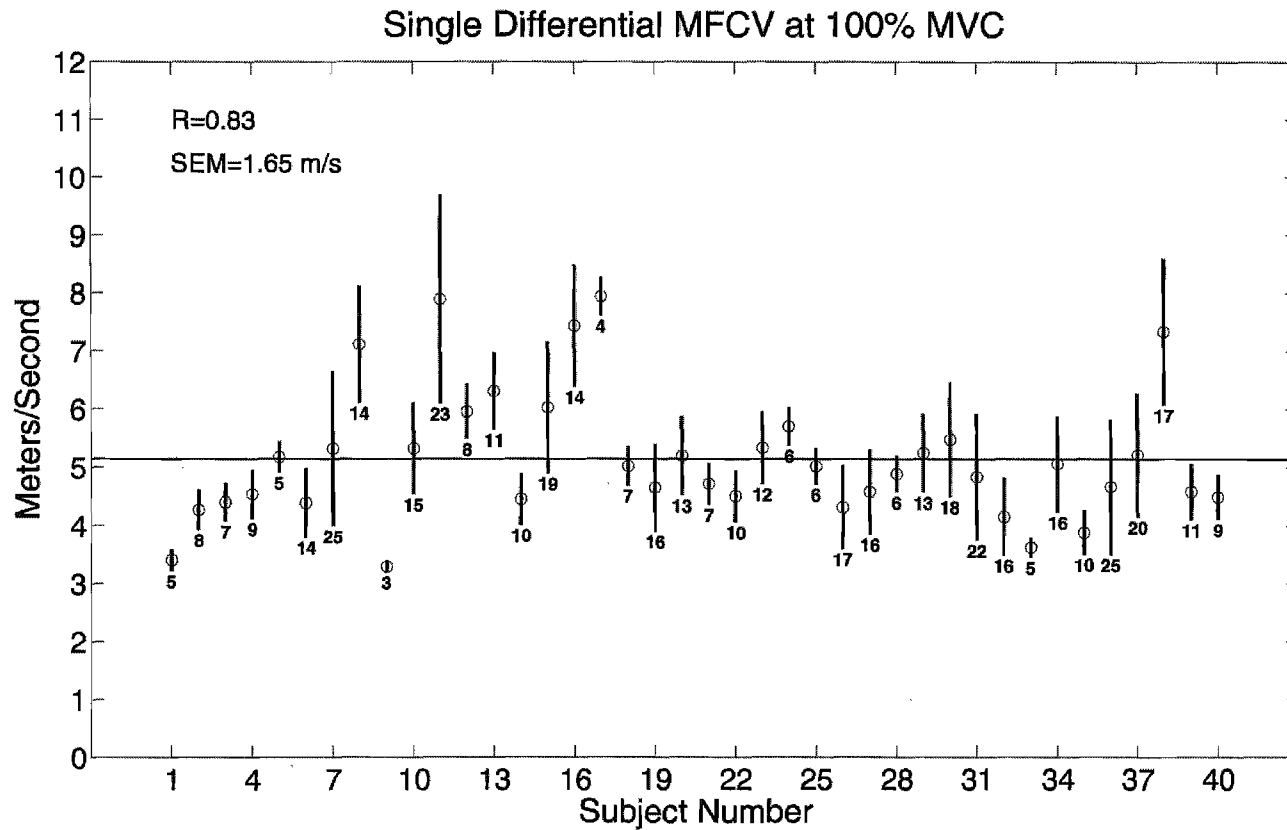


Figure 18. The means (circles) and standard deviations (vertical bars) for muscle fibre conduction velocity (MFCV) determined from single differential electrodes for each subject. The number below the vertical bar is the coefficient of variation for the individual subject. Each subject has three respective values accounted for.

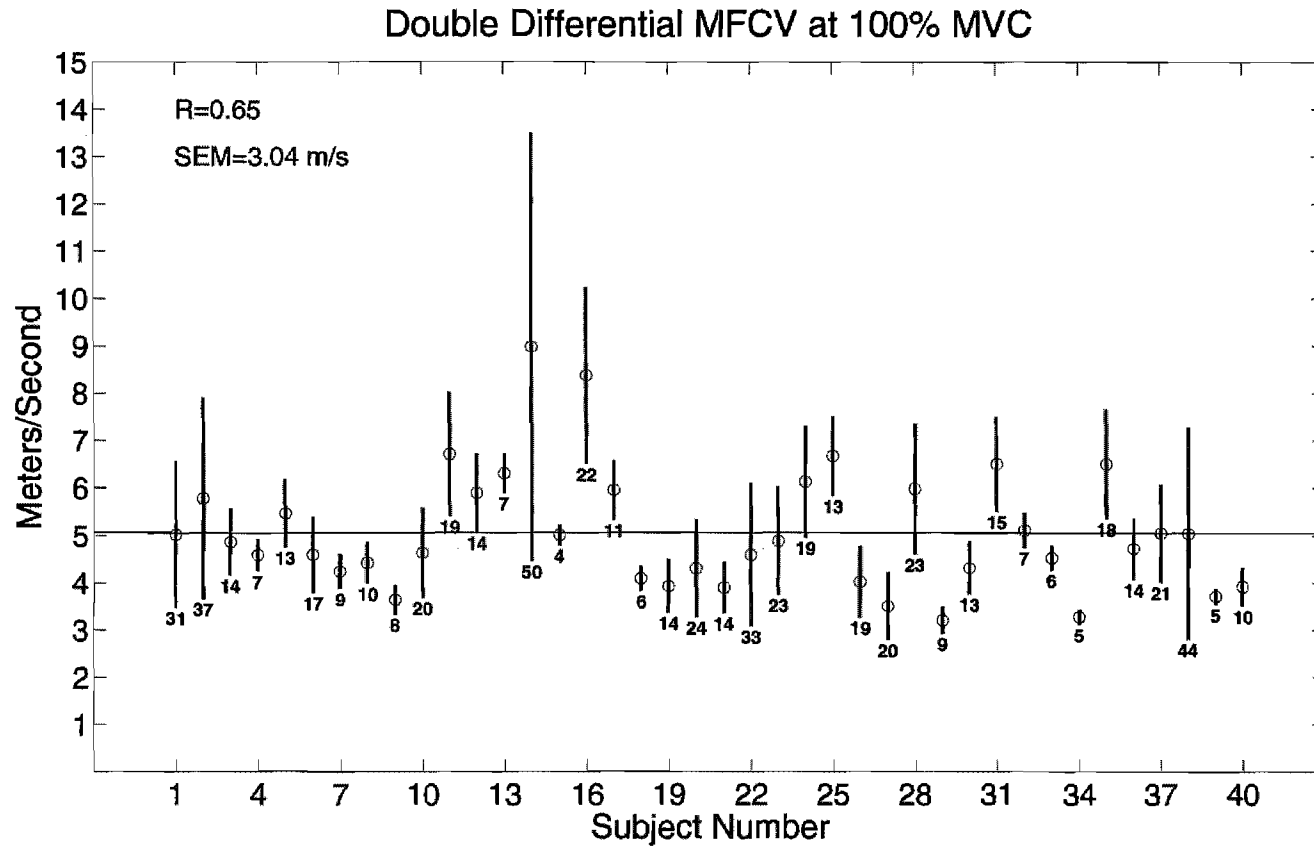


Figure 19. The means (circles) and standard deviations (vertical bars) for muscle fibre conduction velocity (MFCV) determined from double differential electrodes for each subject. The number below the vertical bar is the coefficient of variation for the individual subject. Each subject has three respective values accounted for.



## CHAPTER V

### DISCUSSION

More recent reliability studies on force and sEMG variables have moved away from the traditional use of intraclass correlational analysis of variance. There is an increasing trend to only report the proportion of variance accounted for by each level of measurement, without the actual coefficient. The basic idea is that the most important aspect of the criterion measure is that the between subjects variance accounts for a much greater proportion of total variance than the error components. The Pearson's interclass correlation coefficient, Bland-Altman plots, and the standard error of measurement then form the primary basis upon which reliability analysis is performed (Ng & Richardson, 1996; Kollmitzer et al., 1999; Merletti et al., 1995; 1998; Rainoldi et al., 1999; 2001; Falla et al., 2002). Complicating any potential comparisons further, several reliability studies have employed evoked contractions (Merletti et al., 1995; 1998), submaximal contractions (Bilodeau et al., 1990; Rainoldi et al., 1999; 2001), and/or analyzed the sEMG data normalized to the maximal voluntary contraction (Yang & Winter, 1983). In the following paragraphs, the theoretical implications and practical application of the present findings will be discussed in relation to the literature that most closely matches both the experimental design and measures used in the current work.

#### Force

A grand mean of  $177.7 \pm 62.177$  N for maximal isometric dorsiflexion force was observed in the current study. The group means for males and females will be considered to facilitate a comparison with the literature. Males had a maximal isometric dorsiflexion

force of  $225.88 \pm 43.54$  N while it was  $124.39 \pm 26.29$  N for females. These values are similar in magnitude to those reported by Lenhardt et al. (2009) from this laboratory. Males and females in that study had maximal isometric dorsiflexion force values of  $269 \pm 61$  N and  $191 \pm 51$  N, respectively. Kent-Braun and Ng (1999) observed a maximal isometric dorsiflexion force of  $262 \pm 19$  N for males and  $136 \pm 15$  N for females while Patten and Kamen (2000) reported  $251 \pm 8$  N for males and  $150.9 \pm 4.2$  N for females. Two studies by Broman et al. (1985; 1995) highlight the role of joint position in explaining the differences between studies with respect to maximal force values. The apparatus in these studies also included inversion of the ankle at  $15^\circ$ , which most likely increased force output at the joint to values ranging from 480 to 743 N.

Concern with cross-talk motivated a critical methodological control in the study that may have resulted in "slightly" lower force values (Solomonow et al., 1994; De Luca & Merletti, 1988). Electrical stimulation of the peripheral nerve was conducted to align the electrode on the basis of maximum temporal separation between identical compound muscle action potentials across electrodes detection surfaces. However, subsequent testing using voluntary contractions resulted in excessively high ( $>15$  m/s) muscle fibre conduction velocity values. Extensive pilot testing revealed that subjects were using the toe extensors to augment dorsiflexion force (Marsh et al., 1981; Belanger et al., 1981). It is hypothesized that volume conducted sEMG activity from the long extensors of the toes contaminated the muscle fibre conduction velocity estimates. As evidence, once great care was taken to minimize toe extension during the task, estimates immediately fell to within normal values and were indeed highly stable and consistent within subjects. Nevertheless, controlling for toe extension also reduced the overall dorsiflexion force.

Maximal dorsiflexion force was presented first because it has all of the theoretical characteristics of a highly reliable measure as defined by intraclass correlational analysis of variance. It has been known since the seminal work of Kroll in 1962 that maximal isometric strength scores exhibit remarkable consistency with intraclass correlation coefficients that are typically greater than 0.90. This has been demonstrated for the wrist flexors (Kroll, 1962; 1963), elbow flexors (Carlson & Kroll, 1970; Gabriel & Kroll, 1991), plantar flexors (Kamen, 1983), and knee extensors (Kroll, 1973; Warshal, 1979; Zech et al., 2008). The current work extends these findings with an observed intraclass correlation coefficient of 0.98 for the ankle dorsiflexors. A review of the related literature revealed no other reliability study for maximal isometric actions of the dorsiflexors to offer a basis of comparison.

Reliability assessment using the intraclass correlational analysis of variance is based on the assumption that errors are random and uncorrelated to the magnitude of the score. The more consistent someone is at reproducing their own score, the tighter the spread of scores will be around their own mean (true score). As a result, the within subject variance (error variance) will be low. Further, the spread of scores around each subjects' own mean must be sufficiently small so that there are distinct differences between subjects. That is, the scores for one subject do not overlap with the scores of another subject. It is an important part of measurement theory that a test reveals individual differences between subjects (Henry, 1959). If the mean of one subject is distinctly separated from that of another because the within subjects variance is low, the differences between subjects will be high, unless the range of mean scores is limited. Recall, the between subjects variance is the true score variance. If the between subjects

variance (true score variance) is high relative to the within subjects variance (error variance), the intraclass correlation coefficient will indicate a high level of consistency as demonstrated in this thesis. The true score variance for maximum isometric dorsiflexion strength was approximately 94% of the total variance with the remaining 6% distributed across days and trials.

While consistency is a necessary characteristic for a reliable measure, it is not necessarily sufficient. A highly reliable measure must also exhibit stability in the group mean across repeated measurements. Stability can be problematic without taking the appropriate precautions with respect to the measurement schedule. Kroll (1962; 1963) observed that repeated maximal isometric strength testing is typically associated with a significant increase in strength (up to 15%) due to the measurement schedule alone, without any specific training between sessions. It was suggested that the measurement schedule elicited a motor learning effect because these strength gains occurred in the absence of any overt hypertrophic changes. A number of potential neural mechanisms have therefore been investigated: (1) increased agonist activation; (2) reduced antagonist coactivation; (3) increased recruitment of synergists; and/or (4) alterations in motor unit activity patterns (i.e., rate coding and recruitment) (Calder & Gabriel, 2007; Rutherford & Jones, 1986; Carolan & Cafarelli, 1992; Kamen & Knight, 2004).

Further reliability analysis led to the general practice of including at least one pre-test session to "subtract-out" the initial strength gains associated with motor learning (Calder & Gabriel, 2007). Because the intraclass correlation coefficient is sensitive to the mean square error associated with days, stabilizing the means by subtracting-out the learning effect resulted in a substantial increase in reliability (Kroll, 1963). The

measurement schedule included multiple test days with the expectation that there would be a significant increase in strength. There was a small (3%) but statistically significant increase across the three test sessions that is much lower in magnitude than the expected change (8 – 15%) reported in the literature (Kamen & Knight, 2004; Calder & Gabriel, 2007). The blunted response goes to one of the proposed mechanism involved in the neuromotor aspects of strength gains. That is, learning how to coactivate synergist muscles involved in the task (Rutherford & Jones, 1986). Participants were constantly monitored and discouraged from recruiting the long extensors of the toes during dorsiflexion. They were therefore trained not to recruit an important synergist involved in the task. The stable means for maximal dorsiflexion force observed in this study contributed to extremely high (0.98) intraclass correlation coefficient.

#### Root-Mean-Square Amplitude

It is very difficult to compare actual sEMG values from the same muscle across studies as there a myriad of factors that affect the observed amplitude. There are technical factors associated with the electrode detection system that include the geometry, recording surface area, interelectrode distance, pre-amplification and the resistance associated with the length of the electrode leads, and movement of the muscle underneath the electrode. Methodological issues such as skin preparation and the degree of electrode-skin input resistance, electrode orientation in relation to the muscle fibres, and placement of the electrodes relative to the motor point, and temperature also affect the magnitude of sEMG. The amount of subcutaneous tissue and the training status of the sample are biological factors that can have a significant impact on the magnitude of sEMG (Winkel & Jørgensen, 1991; Knutson et al., 1994; Mathiassen et al., 1995;

Lehman & McGill, 1999; Burden & Bartlett, 1999; Barr et al., 2001; Nordander et al., 2003; Beck et al., 2009; Vera-Garcia et al., 2010).

For example, even in the same laboratory, the observed root-mean-square amplitude sEMG differed markedly from the study by Lenhardt et al. (2009). Exactly the same testing apparatus and sEMG methodology were applied to an identical convenience sample. Yet, the mean value obtained in this study was  $203.42 \pm 95.78 \mu\text{V}$  compared to  $516 \pm 32.20 \mu\text{V}$  observed by Lenhardt et al. (2009). The main differences were that the current study used bar versus the disk electrodes and the inter-electrode distance (0.5 cm) was half that used in the Lenhardt et al. (2009) study. While electrode geometry and surface certainly contribute to this difference, interelectrode distance is a dominating factor. The detection volume of tissue below the electrodes may be conceived as a semi-sphere centered between the two electrodes whose radius is equal to the interelectrode distance (Lynn, 1978). Thus, a larger inter-electrode distance will have a larger detection volume that includes more electrically active tissue. The electrode spatial filtering effects associated with a larger interelectrode distance can also result in an increase in sEMG amplitude (Kamen & Gabriel, 2009).

The debate over ipsative versus normative scaling in sEMG methodology centers on biological factors that can vary between subjects and methodological factors that can vary within subjects across test days (Yang & Winter, 1983; Burden & Bartlett, 1999; Lehman & McGill, 1999). These factors not only affect the magnitude of the observed sEMG activity but also the variability of any measure extracted from the signal. In the case of normative scaling, the sEMG signal may be divided by the net force exerted during a maximal voluntary contraction of the muscle or the peak-to-peak amplitude of

the electrically evoked compound muscle action potential, just to name two possibilities (Yang & Winter, 1983; Pucci et al., 2005). Ipsative scaling for sEMG activity using the raw units of microvolts was used in the current work.

This study demonstrated that the variability of the sEMG signal was sufficiently low so that the spread of scores around each subjects' own mean was small enough to reveal differences between subjects. The error variance due to days and trials (27.32%) was a much smaller percentage of total than the true score variance (72.68%). There is an inherent relationship between force and the magnitude of sEMG activity (Staudenmann et al., 2010). As a result, the sEMG means across test days generally mirrors any changes in force (Häkkinen & Komi, 1983; Calder & Gabriel, 2007). The force values in this study were quite stable across the three test days. The same was true for root-mean-square sEMG amplitude which ranged from  $197.11 \pm 92.68 \mu\text{V}$  to  $207.30 \pm 108.58 \mu\text{V}$  from the first to third test days, respectively. Similar to maximal isometric dorsiflexion force, there was a small (3.8%) but statistically significant increase in root-mean-square sEMG amplitude. This magnitude of change is less than expected (<25%) when repeated testing results in strength gains (Gabriel et al., 2001; Calder & Gabriel, 2007). Thus, the consistency of scores within subjects and the stability of the means across test sessions resulted in a high intraclass correlation coefficient (0.91).

A review of the related literature revealed no reliability studies for root-mean-square sEMG amplitude in tibialis anterior during maximal isometric dorsiflexion. However, excellent reliability ( $R > 0.80$ ) of raw, non-normalised root-mean-square sEMG amplitude is not surprising as the current study replicates the same findings for other muscle groups during maximal effort contractions: these muscles include the elbow

extensors (Gabriel et al., 2001) and flexors (Gabriel & Kroll, 1991; Ollivier et al., 2005), knee extensors (Zakaria, Kramer, & Harburn, 1996; Larsson et al., 2003), and back extensors (Pitcher, Behm, & MacKinnon, 2008). The intraclass reliability coefficients for the average rectified value (ARV) of elbow (Rainoldi et al., 1999) and neck (Falla et al., 2002) flexors have been demonstrated to be slightly lower (0.60 – 0.80) than what has been reported above for the root-mean-square sEMG amplitude. The average rectified value is related to the root-mean-square amplitude, but it is more affected by wave cancellation during the generation of the sEMG interference pattern (De Luca & Van Dyk, 1975; De Luca, 1979; Lowery & O'Malley, 2003).

Thus, while the various potential sources of error in recording the sEMG signal are cause for legitimate concern, this study and others demonstrated that extremely careful methodological controls can result in highly reliable raw sEMG measures. Normalization is not only an unwarranted data transformation, it is inappropriate because factoring the data by a different value for each subject changes the rank order of the distribution and narrows the distribution of scores (Tanner, 1949; Lindquist, 1953). The exception to this rule would be if there were a perfect correlation ( $r=1.0$ ) between the factor and the dependent variable. This would be the same as factoring all the data by the same constant, which is highly unlikely for force and sEMG data as anthropometric and other quantities vary between subjects (Tanner, 1949). In support, intraclass correlational analysis of variance has been conducted on force-normalized sEMG amplitude and lower reliability coefficients were often observed compared to the non-normalized data (Zakaria et al., 1996; Finucane et al., 1998; Arnall et al., 2002; Mathur et al., 2005).



## Frequency of Mean Power

The same technical, methodological and biological factors that affect the root-mean square sEMG amplitude also affect the mean power frequency of signal (Winkel & Jørgensen, 1991; Knutson et al., 1994; Mathiassen et al., 1995; Lehman & McGill, 1999; Burden & Bartlett, 1999; Barr et al., 2001; Nordander et al., 2003; Beck et al., 2009; Vera-Garcia et al., 2010). Thus, a direct comparison of absolute values across studies can prove difficult. A grand mean of  $121.05 \pm 28.81$  Hz was observed in the current work, which is higher than the  $108.00 \pm 28.00$  Hz reported by Lenhardt et al. (2009). However, the difference in absolute magnitude may be predicted by the 0.5 cm smaller interelectrode distance used in this study. Larger interelectrode distances are associated with a broader frequency spectrum that results in a lower mean power (Lynn, 1978; Zipp, 1978; Sinderby et al., 1996; Beck et al., 2005).

The relationship between force and the mean power frequency of the sEMG signal is somewhat more ambiguous than it is for root-mean-square sEMG amplitude. Mean power frequency has been observed to increase up to between 60 and 80% of maximal voluntary contraction (Hagberg & Ericson, 1982; Gerdle et al., 1990; Bilodeau et al., 1992; Sanchez et al., 1993; Sbriccoli et al., 2003; Beck et al., 2005; Lenhardt et al., 2009). It has been suggested that the increase is due to Henneman's Size Principle as higher threshold motor units with faster conduction velocities are progressively recruited (Henneman, 1981; Moritani & Muro, 1987; Solomonow et al., 1990; Kupa et al., 1995). Higher threshold motor units have larger amplitude, shorter duration action potentials (Boe et al., 2005). Mean power frequency therefore increases because the spectrum of sEMG signal is dominated by the shape of the motor unit action potentials (Mills, 1982)

while motor unit firing rate statistics impact the band between 10 and 40 Hz (Lago & Jones, 1977; 1981; Van Boxtel & Schomaker, 1984).

The break point in the force versus sEMG mean power frequency curve tends to be between 60 and 80% of maximal voluntary contraction, close to the limit of the recruitment range for this particular muscle (Christie et al., 2009). After that point, mean power frequency tends to either plateau or exhibit a slight decrease (Gabriel & Kamen, 2009). Thus, the mean power frequency of the sEMG signal can be relatively resistant to changes in maximum isometric strength as would occur during repeated maximum force testing (Gabriel et al., 2001). In contrast, the mean power frequency is sensitive to changes in neuromuscular status (Muro et al., 1982; Latash, 1988a; 1988b) and in the detection of muscular fatigue, where metabolic alterations decrease muscle fibre conduction velocity (Mills, 1982; Kranz et al., 1985; Merletti et al., 1990; Luttmann et al., 2000).

The spread of scores around each subjects' own mean was sufficiently small so there were large, distinct differences between each subject. As a result the true score variance (83.21%) was much greater than the sum of the error variances (16.79%). No similar reliability studies exist for mean power frequency of the sEMG signal from the tibialis anterior during maximal voluntary isometric dorsiflexion contractions. There are, however, several studies that have assessed reliability using the intraclass correlational analysis of variance approach for other muscles and have reported similar results. High intraclass reliability coefficients have been reported for the elbow extensors ( $R=0.93$ ; Gabriel et al., 2001), elbow flexors ( $R=0.99$ ; Daanen et al., 1990), and knee extensors

( $R^2 > 0.80$ ; Larsson et al., 2003) during maximal voluntary contractions of the muscle group.

The reliability observed in this study is higher than what has generally been observed ( $R^2 < 0.90$ ) for median power frequency of the sEMG signal from the knee extensors (Pincivero et al., 2000; Mathur et al., 2005) and back extensors (Ng & Richardson, 1996; Elfving et al., 1999; Arnall et al., 2002; Peach et al., 1998). The mean and median power frequencies are mathematically related to each other (Farina & Merletti, 2000). Some investigators prefer the median frequency for specific applications in assessment of low-back musculature as it is less affected by noise in the sEMG signal and is more sensitive to muscle fatigue (Merletti et al., 1990; Stulen & De Luca, 1981; Hof, 1991). The sensitivity to fatigue is due to the fact that the sEMG power spectrum is positively skewed and becomes even more so due the spectral compression associated with a reduction in muscle fibre conduction velocity (Farina & Merletti, 2000).

### Muscle Fibre Conduction Velocity

Early sEMG studies of muscle fibre conduction velocity involved a minimum array of three electrode detection surfaces to examine the delay time between two single-differentiated sEMG signals (Naeije & Zorn, 1982; Sollie et al., 1985). A fourth detection surface then allowed double-differential recording of sEMG signals. The additional spatial filtering helped to minimize environmental noise and non-propagating activity, which reduced the overall error in the muscle fibre conduction velocity estimates (Broman et al., 1985b). This additional spatial filtering has led to the linear electrode array giving way to matrix-grid electrodes, combined with sophisticated filtering and

signal decomposition software (Rau & Disselhorst-Klug, 1997; Farina et al., 2004). However, both the instrumentation and software are not routine to the average electromyographic kinesiology or clinical electrophysiology laboratory. The main purpose of this thesis was to determine if methodological procedures could be developed for a smaller electrode array, to minimize errors in the cross-correlation method for delay determination and conduction velocity calculation.

It is common practice to reject data that fall outside a specific range because the values would easily be detected and rejected in the clinical situation as non-physiological (Lange et al., 2002; Beck et al., 2004). The rejection criteria are based the range of normal values extracted from the frequency distribution histogram of conduction velocities of individual motor unit action potentials obtained during evoked and voluntary contractions recorded by both surface and indwelling recordings (Troni et al., 1983; Kereshi et al., 1983; Andreassen & Arendt-Nielsen, 1987; Martínez, 1989; Nishizono et al., 1990; Vogt & Fritz, 2006), which are known to yield different estimates (Zwarts, 1989).

Focusing most closely on the study related to the present work, Merletti et al. (1995) eliminated values outside the range of 2 to 8 m/s for evoked contractions of the tibialis anterior. In contrast, the current work included higher muscle fibre conduction velocity values up to 13 m/s based on the probability density function reported by Hunter, Kearney and Jones (1987). The investigators employed the impulse-response method to calculate the probability density function of the muscle fibre conduction velocities contributing to the sEMG signal during a voluntary contraction. The impulse-response

approach avoids the errors associated with the more recent peak identification methods (Lange et al., 2002; Beck et al., 2004).

The decision to increase the upper limit of acceptable muscle fibre conduction velocity was based on both theoretical and practical considerations. Practically, when the voluntary muscle contraction is greater than 60 percent of maximum and the electrodes are close to either the tendon or innervation zone, values up to 12 m/s have been observed (Li & Sakamoto, 1996; Hogrel et al., 1998). Every effort was made to place the electrodes away from electrically identified motor points. In subjects with a smaller tibialis anterior surface territory, the electrode “might” have been placed close to another unidentified innervation zone or tendon. This leads to the more theoretical motivation that it is important to include all data in a realistic assessment of errors and their impact upon reliability (Henry, 1950).

The grand mean value of single-differential conduction velocity across the three days of testing was  $5.14 \pm 1.30$  m/s. There are no other studies in the literature that report means and standard deviations for tibialis muscle fibre conduction velocity during maximal isometric dorsiflexion contractions. Andreassen and Arendt-Nielsen (1987) used microstimulation of the motor units in the tibialis anterior to record evoked responses with single differential electrodes and observed  $3.7 \pm 0.7$  m/s. Broman et al. (1985a) used double differential electrodes to monitor voluntary tibialis muscle activity during isometric dorsiflexion at 50% of maximal voluntary contraction and observed  $4.14 \pm 0.52$  m/s. Double differential recording in the current work resulted in a grand mean of  $5.06 \pm 1.66$  m/s.

The magnitudes of the differences between studies are quite reasonable based on the experimental designs employed. It is well known that voluntary contractions yield higher muscle fibre contraction velocity values than evoked (non-voluntary) contractions. The motor unit firing rates associated with voluntary contractions alter the absolute refractory period of the muscle fibre membrane resulting in faster conducting action potentials (Zwarts, 1989). Muscle fibre conduction velocity is also higher for maximal versus submaximal contractions because higher threshold motor units with faster conduction velocities are recruited (Masuda & De Luca, 1991; Sbriccoli et al., 2003), in addition to potential changes with increased motor unit firing rates (Arendt-Nielsen & Zwarts, 1989). In support, the mean maximal value reported by Zwarts and Arendt-Nielsen (1988) for vastus lateralis muscle fibre conduction velocity at 100% of maximal voluntary contraction was 5.11 m/s, which is almost identical to what was observed in the current work. In fact, mean values slightly less than or greater than 5.0 m/s are not uncommon, depending on the experimental conditions (Lange et al., 2002).

The difference between single- and double-differential values is consistent with the hypothesised benefit of additional spatial filtering. Non-propagating sEMG activity associated with muscle-tendon end effects (also known as far-field potentials or standing waves) is detected by all electrodes simultaneously, increasing the muscle fibre conduction velocity estimates (Broman et al., 1985b). Reduction of the non-propagating waves resulted in lower values for the double-differential recordings.

The coefficient of variation (calculated by dividing the standard error of measurement by the mean and the result multiplied by 100) will be used to provide a basis of comparison. Using maximum isometric force as the gold-standard measure, a

coefficient of variation of 13.8% was observed indicating remarkable consistency within subjects. The low coefficient of variation translated to a very high intraclass correlation coefficient (0.95). The coefficient of variation for muscle fibre conduction velocity obtained by single-differential recordings was 32.0%. Visual inspection of the spread of scores around each subjects' own mean indicates that this was still quite reasonable (Figure 16). There is much more overlap in the scores between different subjects than observed for force, but the differences between subjects can still be observed. Because the true scores (59.67%) dominated the total variance, the intraclass correlation coefficient is still very good (0.83).

The means across test sessions changed less than 3% from  $5.05 \pm 1.39$  m/s to  $5.19 \pm 1.31$  m/s, indicating excellent stability. In this case, the lower intraclass reliability coefficient (0.83) was most likely due to the restricted range effect (Kroll, 1967). This makes teleological sense because “normal values” for a physiological measure such as conduction velocity tend to have a narrow range upon which a clinical diagnosis is made (Preston & Shapiro, 2005). The standard error of measurement was 8.9% of the total range of force scores versus 18.9% for muscle fibre conduction velocity. Nevertheless, the methodology utilized in the current work resulted in the highest intraclass reliability coefficient reported to date. This compares to 0.75 and 0.80 for the biceps brachii using a grid matrix of electrodes (Farina et al., 2004) and recording configuration of two bipolar electrodes (Ollivier et al., 2005), respectively.

The study by Merletti et al. (1995) was the closest in nature to the current work. The investigators reported an intraclass correlation coefficient of 0.11 for tibialis muscle fibre conduction velocity during evoked contractions. Given the careful methodological

controls used in the study, the low intraclass correlation coefficient is rather difficult to explain. One possibility is that data from only six subjects was used to determine muscle fibre conduction velocity. Carlson and Kroll (1970) showed unequivocally that the intraclass correlational analysis of variance technique is sensitive to the number of subjects and not the degrees of freedom involved the *F*-tests. In support of this hypothesis, Merletti et al. (1995) used the data from ten to calculate dorsiflexion torque and all other sEMG measures and they reported intraclass correlation coefficients between 0.78 and 0.88.

A common finding for both force and sEMG measurement is that the day-to-day error variance is the second largest portion of the total variance (Merletti et al., 1998; Rainoldi et al., 1999; 2001; Gabriel et al., 2001; Farina et al., 2004; Calder et al., 2005; Christie et al., 2005). The largest potential sources of day-to-day error are changes in subject test position, muscle temperature, skin preparation, and electrode placement, all of which remain constant across trials within each day (Daanen et al., 1990). All the measures in this study followed this same distribution of variance and re-affirms the notion that, if strict methodological controls are followed, the day-to-day error variance can be much smaller in magnitude than the variance between subjects (true score variance), affording a high reliability coefficient.

There was a 5.4% change in muscle fibre conduction velocity group means across test days for double-differential recordings. The values ranged from  $4.99 \pm 1.60$  m/s to  $5.26 \pm 2.05$  m/s across the three test sessions, which is fairly stable. The day-to-day error variance was however the greatest proportion of the total variance (57.32%). The “relatively” stable group means across test sessions but high day-to-day error variance



within subjects suggest that changes in rank across days by one subject were compensated for by changes in rank by another subject. The day-to-day error also contributed to a much larger standard error of measurement (3.04 m/s) that resulted in a coefficient of variation of 60% of the grand mean, which is quite high. If the one extreme value (13.8 m/s) is eliminated from the double-differential recording data, the “true” range of values was from 2.71 to 8.33 m/s. The standard error of measurement was then 54.1% of the total range of scores, making it very difficult to observe differences between subjects (Figure 17). The low true score variance resulted in an intraclass reliability coefficient (0.65) that would be deemed adequate at best.

The low intraclass reliability coefficient is perplexing given that the day-to-day error variance, which is the largest potential source of error variance, “should have” been the same between single- and double-differential recordings. One possible explanation is that the increased selectivity of double-differential recordings is a “double-edged sword”. Olliver et al. (2005) reported a similar finding when comparing muscle fibre conduction velocity estimates for the biceps brachii at 100% of maximal voluntary contraction obtained from bipolar and Laplacian electrode configurations. The bipolar electrode configuration was highly reliable ( $R=0.80$ ) while the Laplacian electrode configuration was not ( $R=0.11$ ). The investigators argued that the increased selectivity of the Laplacian electrode configuration increases the sensitivity to slight displacements of the muscle underneath the electrodes or small alterations in motor unit activity patterns that would modify the sEMG signal (Ollivier et al., 2005). Heightened sensitivity to alterations in motor unit activity patterns would increase the day-to-day error variance with slight differences in test position and mechanical advantage of the lower leg. However, a

higher trial-to-trial error variance would also be expected as subjects would recruit the tibialis slightly differently from one trial to next (Rummel, 1974; Gabriel, 2000; Merletti et al., 1995). This was not the case as the trial-to-trial error variances for muscle fibre conduction velocity were similar between the single- and double-differential recordings. Koh and Grabiner (1993) reported similar findings for the variability in the amplitude of sEMG activity.

The interelectrode distance used in the present study was 0.5 cm which translates to the muscle fibre action potentials traveling 2 cm from the first to last detection surface. Every effort was made to select an electrode placement that maximized shape similarity and delay across the three single-differential signals during the evoked contractions. While the conditions may have been optimal for the first two single-differential signals, it may not have been the case for the third differential signal. That is, the first two or three detection surfaces were “more” in line with the muscle fibres than the last one. This might be expected due to the curvature of the lower leg. Hägg (1993) observed a similar problem when trying to align an electrode array of four detection surfaces spaced 0.5 cm apart on the trapezius. This type of error could differ markedly from day-to-day. One possibility is to use smaller (i.e., 0.25 cm) interelectrode distances, with wider detection surfaces as they appear to be less sensitive to misalignment of the electrode in respect to the muscle fibre orientation (Sadoyama et al., 1985; Sollie et al., 1985; Farina et al., 2004).

The methodological controls placed upon this study may place the ecological validity of this study under criticism. These controls includes participant exclusion criteria, the use of a Faraday Cage, as well as the complexity of the technique utilized for

ensuring maximum delay between waveforms. The use of voluntary contractions as opposed to evoked potentials is another consideration. These claims are to a certain extent justified, but are secondary to demonstrating that the reliability measurement of MFCV can be obtained.

### Summary and Conclusions

The purpose of this thesis was to develop procedures using simple instrumentation to measure muscle fibre conduction velocity during maximal effort contractions. The efficacy of the procedures was assessed using intraclass correlational analysis of variance technique. It was assumed that a high intraclass correlation coefficient would indicate that the procedures were indeed reliable and therefore useful. The reliability of maximal force and other sEMG variables were evaluated as benchmark measures for the experimental techniques used in this thesis, to offer a basis of comparison.

The mean squares used to calculate the intraclass correlation coefficient are sensitive to the day-to-day error variances. The group means across test days for all measures exhibited very slight changes (<5%, at most) and were therefore considered very stable. All measures also exhibited remarkable consistency within subjects as indicated by high reliability coefficients (0.83 – 0.98), except for muscle fibre conduction velocity measured by double-differential recordings ( $R=0.65$ ). The reliability of muscle fibre conduction velocity measured by single-differential electrodes was higher than what has been reported using more sophisticated instrumentation and software ( $R=0.83$ ).

The methods section of this thesis outlines all the traditional sEMG methodological controls that were followed. It may be concluded that the following additional procedures can result in highly reliable muscle fibre conduction velocity using only single-differential sEMG signals: (1) electrical identification of motor points prior to electrode placement, (2) twitch identification of muscle fibre orientation to guide initial electrode placement, (3) placement verification to maximize the similarity and delay of evoked potentials across all detection surfaces, and (4) minimization of synergistic activity during voluntary contractions.

APPENDIX A  
INFORMED CONSENT DOCUMENT

Consent Form

**INFORMED CONSENT DOCUMENT**

***Title of Project:*** Reliability of Muscle Fibre Conduction Velocity in the Tibialis Anterior

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*This study has been reviewed and approved by the Brock Research Ethics Board (#02-283). 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 Kyle McIntosh and supervised by David Gabriel Ph.D. The electrical signal of skeletal muscle is measured from the skin surface, similar to electrocardiography (ECG) which measures the electrical activity of cardiac (heart) muscle. The electrical signal representative of skeletal muscle is termed, electromyography (EMG). The electromyogram is recorded via electrodes placed on the skin and can be analyzed in a number of different ways to evaluate muscle activation.

You will come to the Electromyographic Kinesiology Laboratory (WH21), where a series of five testing sessions will be conducted, each separated by a minimum of 48 hours. Each of these sessions will be approximately an hour and a half in duration. You are requested to refrain from any strenuous activity for a period of 24 hours prior to testing.

## **Plan and Procedures**

All testing will be conducted by Kyle McIntosh who will implement the following protocol. You will be asked to complete the PAR-Q assessment of physical health status to ensure that you are not at health risk during the following experiment.

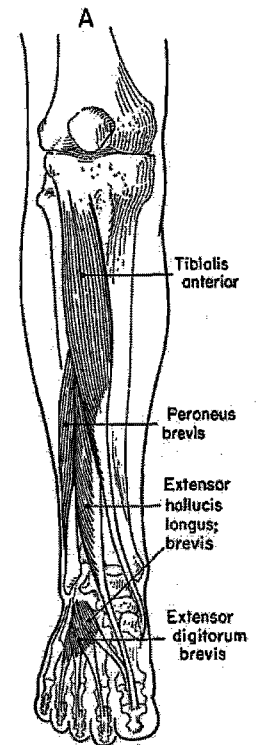
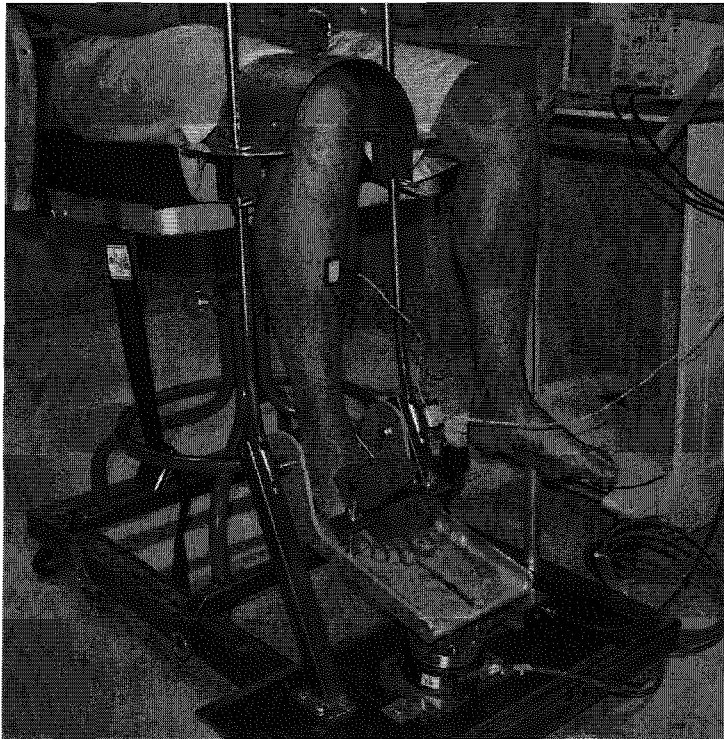
Upon completion of the assessment, the right leg will be prepared for testing. Small superficial areas of the tibialis anterior and soleus will be shaved, lightly abraded and cleansed with alcohol. These areas correspond to the location of the electrodes that will be taped to the skin surface. The electrodes will measure the electrical activity of tibialis anterior and soleus.

We will locate the motor point (neuromuscular junction) with a mild electrical stimulation of the tibialis anterior. A linear 4-bar electrode array will be placed 10 millimeters distal to the most distal motor point.

The protocol will begin with five maximal voluntary contractions (MVCs) of the dorsiflexors that will be separated by three minutes of rest between each contraction. This will be followed by three MVCs of the plantar-flexors once again separated by three minutes rest. Following the final rest period another seven voluntary contractions will take place at 30% of MVC. Three minutes of rest will again separate each of these contractions. Finally the testing session will conclude with another set of three MVCs.

## **Recording Voluntary Muscle Activity**

Before electrode placement, the skin surface will be shaved, lightly abraded, and cleansed with alcohol to reduce the skin-electrode impedance. A bipolar surface electrode array will record voluntary muscle activity at the skin surface of the tibialis anterior while 2 Ag Ag-Cl electrodes will be placed on the soleus. The positions of the electrodes will be marked with indelible ink to ensure the consistency of the placement.



### Risks and Discomforts

1. 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 investigator anticipates no major risks or discomforts will occur in the present project.
2. 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.
3. There may be discomfort related to the delayed onset of muscle soreness associated with isometric contractions of the leg muscles. If muscle soreness does occur, it is usually very mild and should dissipate within 72 hours.
4. 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, Kyle McIntosh, 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, Kyle McIntosh or supervising faculty 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 do 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. The 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 investigators, Kyle McIntosh by telephone 905-688-5550 ext. 3965 or by e-mail at km03ki@brocku.ca. If questions arise about your rights as a research subject, you may contact the Office of Research Services at 905-688-5550 ext. 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)                      Kyle McIntosh (M.Sc. candidate)  
   *Department of Physical Education and  
   Kinesiology*

APPENDIX B

PAR - Q

## **Electromyographic Kinesiology Laboratory**

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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.*

<b>YES</b>	<b>NO</b>	
<input type="checkbox"/>	<input type="checkbox"/>	1. Has your doctor ever said that you have a heart condition <u>and</u> that you should only do physical activity recommended by a doctor?
<input type="checkbox"/>	<input type="checkbox"/>	2. Do you feel pain in your chest when you do physical activity?
<input type="checkbox"/>	<input type="checkbox"/>	3. In the past month, have you had chest pain when you were not doing physical activity?
<input type="checkbox"/>	<input type="checkbox"/>	4. Do you lose your balance because of dizziness or do you ever lose consciousness?
<input type="checkbox"/>	<input type="checkbox"/>	5. Do you have a bone or joint problem that could be made worse by a change in your physical activity?
<input type="checkbox"/>	<input type="checkbox"/>	6. Is your doctor currently prescribing drugs (e.g. water pills) for your blood pressure or heart condition?

7. Do you know of any other reason why you should not do physical activity?

**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): \_\_\_\_\_

\_\_\_\_\_

Signature of Participant

Date (day/month/year)

Witness:

Name of Witness (Please Print): \_\_\_\_\_

\_\_\_\_\_

Signature of Witness

Date (day/month/year)

**APPENDIX C**

**SUBJECT DATA COLLECTION SHEET**

## Subject Data Collection Sheet

### **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:** \_\_\_\_\_ **Sex:** \_\_\_\_\_

**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 1 yr)  
\_\_\_\_\_
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? \_\_\_\_\_

### **Anthropometric Measurements**

1. Height \_\_\_\_\_
2. Weight \_\_\_\_\_
3. BMI (Weight in kg / Height<sup>2</sup> in metres) \_\_\_\_\_
4. Lower Leg Circumference (widest point) \_\_\_\_\_
5. Lower Leg Length (lateral head of the fibula – lateral malleolus) \_\_\_\_\_
6. Foot Length \_\_\_\_\_



Date: \_\_\_\_\_

Pre- $\Omega$ : \_\_\_\_\_

Post- $\Omega$ : \_\_\_\_\_

Subject #: \_\_\_\_\_

Pre-temp: \_\_\_\_\_

Post-temp: \_\_\_\_\_

	<b>Trial 1</b>	<b>Trial 2</b>	<b>Trial 3</b>
<b>FORCE (mV) 100%</b>			
<b>Amplification</b>			
<b>MPF</b>			
<b>RMS</b>			
<b>CV – Single</b>			
<b>x-corr</b>			
<b>CV – Double</b>			
<b>x-corr</b>			

<b>FORCE (mV) 30%</b>	<b>Trial 1</b>	<b>Trial 2</b>	<b>Trial 3</b>	<b>Trial 4</b>	<b>Trail 5</b>
<b>Amplification</b>					
<b>MPF</b>					
<b>RMS</b>					
<b>CV – Single</b>					
<b>x-corr</b>					
<b>CV – Double</b>					
<b>x-corr</b>					

	<b>Trial 1</b>	<b>Trial 2</b>	<b>Trial 3</b>
<b>FORCE (mV) 100%</b>			
<b>Amplification</b>			
<b>MPF</b>			
<b>RMS</b>			
<b>CV – Single</b>			
<b>x-corr</b>			
<b>CV – Double</b>			
<b>x-corr</b>			

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