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Title: The relationship between myoelectrical properties and contraction intensity

Name: Christopher Lane

THE RELATIONSHIP BETWEEN MYOELECTRICAL PROPERTIES AND CONTRACTION INTENSITY

CHRISTOPHER LANE

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The Relationship Between Myoelectrical Properties and Contraction Intensity

BY

Christopher Lane

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Author's Declaration

I, Christopher Lane, declare that this thesis and the work presented in it are my own and has been generated by me as the result of my own original research.

The relationship between myoelectrical properties and contraction intensity.

I confirm that:

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3. Where I have cited the published work of others, this is always clearly attributed;

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Abstract

The aim of this study was to investigate the relationship between muscle force and electromyography (EMG) amplitude of muscles around the ankle, elbow, and knee. Frequency analysis was also conducted.

31 healthy volunteers (24.55 ± 4.64 years, 1.72 ± 0.08 m, 72.08 ± 23.67 kg) participated in this study. Volunteers were injury-free and provided informed consent before taking part. Surface EMG (sEMG) electrodes were situated over a total of 15 muscles over three joints. The right ankle, elbow, and knee joints were fixed at their mid-range of motion while participants isometrically contracted antagonistic muscle pairs at 20, 40, 60, 80, and 100 per cent contraction intensity.

Visual inspection and coefficient of determination (R^2) of a linear trendline was used to determine linearity of the EMG-force relationship of each muscle tested. An R^2 value ≥ 0.990 indicated a linear relationship, between 0.980 – 0.989 indicated a curvilinear relationship, and ≤ 0.979 indicated an elbow point relationship;

<u>Linear relationships:</u> triceps brachii – lateral head (R^2 0.995), triceps brachii – long head (R^2 1.000), semitendinosus (R^2 0.997), fibularis brevis (R^2 0.999), fibularis longus (R^2 0.992).

<u>Curvilinear relationships</u>: biceps femoris (R^2 0.987), soleus (R^2 0.985), tibialis anterior (R^2 0.983).

<u>Elbow point relationships:</u> biceps brachii (R^2 0.965), brachioradialis (R^2 0.930), rectus femoris (R^2 0.968), vastus lateralis (R^2 0.971), vastus medialis (R^2 0.967), gastrocnemius lateralis (R^2 0.964), gastrocnemius medialis (R^2 0.968).

Based on these data muscle fibre type and joint angle appear to have the greatest influence on the EMG-force relationship. Future research using a wider range of joint angles would be recommended.

Acknowledgements

The Masters by Research was not an easy journey as this was the longest project I have had to work on to date. This year has provided me with opportunities I never thought I would be experiencing, including taking on the role of an Hourly-Paid Lecturer and presenting at the 15th Staffordshire Conference on Clinical Biomechanics. As a result, I feel I was able to further develop my academic writing skills and boost my self-confidence.

I would like to thank my supervisors, Dr. Dan Robbins and Dr. John McCarthy, for supporting my learning and guiding me throughout the year. I would also like to thank the participants who volunteered their time to take part in the data collection process. Finally, many thanks to my family and friends for encouraging, motivating, and supporting me to keep doing the best I can.

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Glossary

Amplitude – The single maximum positive or negative voltage of a sine wave, with respect to zero.

Elbow point relationship – An EMG-force relationship where two linear lines of differing gradients join at a single point.

Frequency - The number of cycles completed in a single second.

Innervation zones – Regions of a muscle where the neuromuscular junctions are located.

Instantaneous value – A value of varying quantity at a specific time in a cycle.

Motor unit - Consists of a motor neuron and the muscle fibres innervated by it.

Muscle fascicles – Muscle fibres arranged in bundles and surrounded by perimysium.

Natural numbers – Positive integers used for counting (cardinal) and ordering (ordinal).

Real numbers – Includes rational numbers (positive and negative integers, and fractions) and irrational numbers (Pi π, $\sqrt{2}$).

Torque – A force applied around an axis of rotation.

Abbreviations

1RM – One Repetition Maximum

ACSA – Anatomical Cross-Sectional Area

ACSM – American College of Sports Medicine

A/D – Analogue-to-Digital

Ag - Silver

AgCl – Silver Chloride

ALS – Amyotrophic Lateral Sclerosis

ASCII – American Standard Code for Information Interchange

BB – Biceps Brachii

BF – Biceps Femoris

BR – Brachioradialis

DAQ – Data Acquisition

dEMG – Decomposition Electromyography

ECG – Electrocardiography

EMG – Electromyography

FB – Fibularis Brevis

FFT – Fast Fourier Transform

FL – Fibularis Longus

GL – Gastrocnemius Lateralis

GM – Gastrocnemius Medialis

Hz – Hertz

IHR – Institute for Health Research

MVC – Maximal Voluntary Contraction

PA – Pennation Angle

PCSA – Physiological Cross-Sectional Area

PNF – Proprioceptive Neuromuscular Facilitation

PPS – Pulses per second

RF – Rectus Femoris

RMS – Root Mean Square

RoM – Range of Motion

sEMG – Surface Electromyography

ST – Semiteninosus

Sol - Soleus

TB Lateral – Triceps Brachii (Lateral Head)

TB Long – Triceps Brachii (Long head)

TA – Tibialis Anterior

VL – Vastus Lateralis

VM – Vastus Medialis

1. Introduction

The examination of muscle function can be achieved by using various methods and techniques, either by using manual methods or by using equipment such as hand-held dynamometers, weights, and isokinetic dynamometers (Jones and Stratton, 2000). Physical assessments may be used to establish whether factors such as changes in sport, equipment, a rehabilitation program, nutritional supplements, or medication has affected muscle function. Furthermore, factors including illness, injury, or if a change in body position or posture has affected muscle function can be assessed with physical assessments. Most importantly for this study, physical assessments can be used to compare muscle groups when looking at their activity levels. Monitoring muscle activity levels is crucial to the assessment of their function, as it determines whether the muscle groups can be assessed in the same way via establishing if the amplitude of the electromyography (EMG) signals respond in the same way. For instance, assessments of muscle function cannot be processed in the same way if the amplitude of the gastrocnemius and the soleus EMG signal do not respond in the same way as each other.

It has been previously reported that the relationship is a proportional one between the muscle force produced and the EMG signal amplitude observed, where the more force that is produced the greater the EMG signal amplitude is (De Luca, 1997). Muscle force is related to the length of the muscle and the velocity at which it contracts. It has been reported that as a muscle contracts and its fibres become shorter, muscle force decreases (Maganaris, 2001). Consequently, as a muscle contracts and its fibres shorten, less force is produced, thus the EMG amplitude decreases. Contrary to this, the velocity at which a muscle is contracted is inversely proportionate to the force produced (Tipton, 2006).

For many years, it has been accepted that the amount of muscle force produced during a contraction relies on how many motor neurons are recruited and the rate at which action potentials are discharged (Enoka, 2015). Currently, there are two established and widely accepted theories of controlled modulation of force; Henneman's Size Principle (Henneman, Somjen and Carpenter, 1964) and rate coding (Enoka, 2015). However, with the development of technology over the past 60 years, these existing theories are being challenged. In recent years, it has been shown that motor units which have been recruited earlier during a sustained voluntary contraction maintain higher firing rates than those which have been recruited later (De Luca and Contessa, 2015). With this progress, the validity of old methods of assessing muscle function are beginning to be questioned, while new methods are being developed to provide more reliable and valid results which can be used by practitioners.

Surface EMG (sEMG) is a technique used to measure action potentials by placing electrodes on the skin between the myotendinous junction and closest innervation zone (De Luca, 1997). If surface electrodes are placed directly over innervation zones, signal attenuation will occur (Saitou et al., 2000). However, if four or more motor units are around the area of measurement the action potentials recorded may be not be distinguishable from each other due to superimposition of EMG signals. To resolve this issue, a technique called decomposition electromyography (dEMG) could be used to break down raw EMG signals to obtain information about individual motor unit action potentials (De Luca et al., 2006). dEMG may also be used to develop understanding of how the central nervous system controls motor units to generate force (De Luca et al., 2006). However, these would be very difficult to achieve and evaluate as, to the best of the author's knowledge, there are only two dEMG systems located in the United Kingdom. Due to this, the importance of increasing the knowledge of sEMG systems to take advantage of their capabilities becomes greater.

The data collected from sEMG can be used to identify the relationship between muscle force and EMG amplitude, which has been seldom researched over the past 70 years. Data produced in existing research have shown that different muscles may have a linear or a non-linear EMG-force relationship, although it is unclear why.

The data that has previously been compiled from investigating a relatively small number of muscles in different studies have generally produced inconsistent results. An example of this is brachioradialis, which was examined in two studies and was shown to have a linear relationship in one and non-linear in the other (Bigland-Ritchie, 1981; Praagman et al., 2003). This may be due to how technology greatly differed during the times of each study as the latter study was conducted 22 years after the former. Not only this, but as it has been a long time since existing research has been conducted the results may differ with the use of updated methods. Therefore, the purpose of this study was to analyse the sEMG-force relationship of muscles around the elbow, knee and ankle to examine the linearity of the sEMG-force relationship.

2. Literature Review

2.1 Muscle Physiology and Biomechanics

2.1.1 Sarcomere Structure

A *sarcomere* can be described as the basic unit of contraction within a muscle and is measured from one *Z-disk* to the next (Enoka, 2015). Sarcomeres contain thick and thin protein filaments known as *myosin* and *actin*, respectively, giving muscles their striated appearance when sarcomere units are repeated (Epstein and Herzog, 1998). The diameter of myosin is 12 nm with a length of 1.6 µm while actin has a diameter of 7 nm and is 1.27 µm long (Enoka, 2015). The middle of a sarcomere is known as the *M-line*. Sarcomeres aligned in series are known as a *myofibril* which has a diameter of approximately 1 µm (Enoka, 2015). Many myofibrils arranged in parallel form a *muscle fibre*. Muscle fibres are innervated by neurons known as *motor neurons*. Furthermore, the grouping of motor neurons and muscle fibres are known as *motor units* (Enoka, 2015).

Cross-bridges are formed when a myosin filament head attaches to an actin filament to contribute towards the shortening of the sarcomere (Wood, 2016). Cross-bridges located on the thick filament are approximately 14.3 nm apart with a rotation of 60° from each other (Epstein and Herzog, 1998). Therefore, it can be approximated that cross-bridge pairs with identical rotations of 180° are 43 nm apart (Epstein and Herzog, 1998).

2.1.2 Torque

When force is applied to an object a different effect can be achieved depending on how the object is supported, as well as the direction in which the force is applied (Nordin and Frankel, 2012). *Torque*, measured in Newton-meters (N·m) or foot-pounds (ft·lb), is the term given to a force applied around an axis of rotation (Hall, 2015; Nordin and Frankel, 2012). Torque (T) is the product of multiplying force (F) by moment arm:

$$T = Fd_{\perp} \tag{1}$$

where the moment arm is the perpendicular displacement ($d_{\perp}$) from the axis of rotation and line of force (Hall, 2015; Knudson, 2007). Measuring torque during isometric contractions can provide researchers and clinicians with valuable data, including peak torque throughout a contraction and peak torque at various joint angles.

When measuring peak torque during a contraction a single value is recorded, which does not present information on how torque changes over time. Changes in peak torque over time can be measured to offer data on muscular endurance or fatigue, which is typically presented as a percentage of the maximum torque generated (Richards, 2008).

It can also be useful to know how joint angle affects peak torque as one angle may have a mechanical advantage over another due to muscle origins and insertions. A series of peak torque measurements can be recorded at the varying joint angles which can be compared against each other (Richards, 2008).

2.1.3 Muscle Volume and Thickness

Research concerning muscle volume is generally quite limited. Vastus lateralis (VL) and vastus medialis (VM) have been reported to have a volume of 647 cm^3 and 461 cm^3, respectively (Erskine et al., 2009). Furthermore, the volume of rectus femoris (RF) has been shown to be 339 cm^3 (Erskine et al., 2009). Sol has been reported to have a volume of 489 cm^3 (Fukunaga et al., 1992) which is most similar to VM. The volume of the remaining two triceps surae muscles, gastrocnemius lateralis (GL) and gastrocnemius medialis (GM), have been shown to range between 140 – 185.6 cm^3 and 185 – 303.3 cm^3, respectively (Fukunaga et al., 1992; Narici et al., 1998; Thom et al., 2007; Tomlinson et al., 2014).

A characteristic related to volume it the thickness of muscles, which has been shown to be similar across various joints, although slightly decreases from elbow to knee to ankle. For example, Miyatani et al. (2004) reported the thickness of biceps brachii (BB) to be 32 mm and 33 mm using MRI and ultrasound, respectively. Similarly, the thickness of lateral head of triceps brachii (TB Lateral) and long head of triceps brachii (TB Long) has been found to be between 30 mm – 35 mm, although age and sex were indicated to be influencing factors (Ichinose et al., 1998; Kawakami et al., 1995; Kawakami et al., 2006). RF, VL, and VM also have been shown to be of similar thickness to each other, with studies reporting each to be 20 mm, 25 mm – 26.1 mm, and 30 mm, respectively (Ando et al., 2014; Baroni et al., 2013). Likewise, biceps femoris (BF) and semitendinosus (ST) have been shown to have a thickness of 20 mm and 21 mm, respectively (Kellis et al., 2010). The thickness of the individual muscles of the triceps surae group has also been studied. GM has previously been shown to have a thickness between 16.9 mm – 18.6 mm (Maganaris et al., 1998; Narici et al., 1996), GL between 15.6 mm – 16 mm (Maganaris et al., 1998), and soleus (Sol) between 13.9 mm – 16 mm (Maganaris et al., 1998).

2.1.4 Blood Supply and Innervation

The arteries supplying blood to muscles are a characteristic that is recognisable throughout the population. Blood is supplied to BB via the brachial artery and BR via the radial recurrent artery while TB Lateral and TB Long have the deep brachial artery supplying blood to both. GM and GL share their blood supply from the sural artery. Likewise, fibularis brevis (FB) and fibularis longus (FL) are both supplied blood from the peroneal artery. Tibialis anterior (TA) does not share a blood supply with any ankle muscle investigated in the current study, receiving blood from the anterior tibial artery. Additionally, Sol is supplied by popliteal artery branches, the peroneal artery, and the posterior tibial artery. Regarding the knee muscles, VL and VM obtains blood from the lateral circumflex femoral and femoral arteries, respectively, while RF is supplied by the lateral circumflex femoral artery descending branch. BF and ST are both supplied by the perforating arteries as well as having a separate blood supply: BF is also supplied by the deep femoral artery and ST is supplied by the inferior gluteal artery.

Like the arteries, particular nerves are also recognisable throughout the population. Several muscles are known to be innervated by the same nerve. For example, FB and FL are innervated by the superficial peroneal nerve at L4, L5, and S1. Likewise, GM, GL, and Sol are all innervated by the tibial nerve at L5, S1, and S2. With regards to the elbow muscles of interest, brachioradialis (BR), TB Lateral, and TB Long are all innervated by the radial nerve at C6, C7, C8, and T1. However, BB is innervated by the musculocutaneous nerve at C5 and C6 which is not shared with the previously mentioned elbow muscles. Following this, the knee extensors are innervated by the same femoral nerve at L2, L3, and L4. Additionally, BF and ST are innervated by branches of the sciatic nerve at L4, L5, S1, S2, and S3.

Relating to the nerves, muscles contain innervation zones. Innervation zones are regions of a muscle where the neuromuscular junctions are located (Masuda and Sadoyama, 1991). The distribution of innervation zones has previously been investigated (Saito et al., 2000) and presented (Appendix 1.1, 1.2, and 1.3).

2.2 Electromyography

Electromyography (EMG) is a technique used to record muscle activation, where the information can be used in various settings.[1] All living organisms consist of cells that generate electrical impulses to function and is particularly true for muscle cells. During a muscle contraction, EMG records the electrical activity of muscle cells, which is initiated by nerve impulses (Najarian and Splinter, 2012).

[1] The term *electromyogram* is comprised of three terms: *electro*, relating to electrical impulses; *myo*, a prefix of Greek origin meaning muscle; and *gram*, a suffix of Greek origin relating to a drawing or recording (Najarian and Splinter, 2012).

sEMG was used in the current research over intramuscular EMG due to the equipment that was available at the time of data collection. Additionally, sEMG electrodes are relatively easy to position and remove without the use of invasive electrodes that participants may have felt uncomfortable using.

2.2.1 Electrodes

The purpose of an EMG electrode is to enable a process called *signal transduction*; electric potentials produced by muscles are converted into electrical signals that are conducted and relayed to an amplifier via connecting wires (Kamen and Gabriel, 2006).

Several aspects should be taken into careful consideration when determining which electrodes should be used and how they should be applied. De Luca (1997) categorised electrode placement and structure as extrinsic causative factors. Four main points of interest have been identified for these factors. Electrode configuration is one factor; the shape and size of the electrode determines how many motor units will be covered when placed on the skin surface. In addition to this point, the distance between electrodes can affect the amplitude and frequency of the recorded signal (De Luca, 1997; Kamen and Gabriel, 2010; Reaz et al., 2006). Another factor is the location of the electrode. With respect to muscle motor points and myotendinous junctions, electrode location can influence the amplitude and frequency of a signal (De Luca, 1997; Reaz et al., 2006). Furthermore, the amount of crosstalk detected is affected by the position of the electrode with respect to the lateral side of a muscle (De Luca, 1997; Reaz et al., 2006). Finally, the orientation of the electrode with respect to muscle fibres can affect the amplitude and frequency of the EMG signal (De Luca, 1997; Reaz et al., 2006). Electrodes aligned perpendicular to muscle fibres has been shown to decrease measured amplitude and frequency values, as opposed to parallel placement (Camic et al., 2010).

2.2.1.1 Indwelling Electrodes

Indwelling electrodes may be used by clinicians or researchers if individual motor unit activity needs to be assessed. Those who wish to examine individual motor unit activity during an isometric contraction can opt to use a single needle electrode. Similarly, two fine-wire electrodes can be utilised when focussing on individual motor units, although they are generally used during a dynamic contraction, deep within a muscle (Kamen and Gabriel, 2010).

Needle electrodes consist of a cannula with a bevelled edge of 15° – 20° and one or more wires running through it (Kamen and Gabriel, 2010). The wire is usually made of either stainless steel, silver, platinum, or nichrome and generally has a diameter of between 25 µm – 100 µm (Kamen and Gabriel, 2010). The needle diameter is usually between 0.30 mm and 0.65 mm (Richards, 2008). Configurations can be either monopolar or bipolar. A monopolar configuration is comprised of one wire (Kamen and Gabriel, 2010). However, a bipolar configuration consists of two adjacent platinum-iridium wires with a gold-plated needle (Kamen and Gabriel, 2010; Richards, 2008). Fine-wire electrodes are commonly insulated and have a diameter of 50 µm (Kamen and Gabriel, 2010).

2.2.1.2 Surface Electrodes

As the name suggests, sEMG uses electrodes that are placed on the skin surface, above the muscle(s) to be assessed. Like indwelling electrodes, there are different electrode configurations including monopolar, bipolar/single differential, and double differential.

The monopolar configuration requires a single electrode over the muscle of interest, together with a grounding electrode over a bony prominence. Recording EMG signals with this setup is not often used as the data acquired may not be of good quality but may be used in a clinical environment (Richards, 2008).

The bipolar/single differential configuration consists of two electrodes placed 1 cm – 2 cm apart, along the muscle fibre orientation, also with a grounding electrode over a bony prominence. This arrangement will obtain data of higher quality than that of the monopolar arrangement due to the two electrodes removing common signals and amplifying the dissimilar ones (Richards, 2008). Because of this, the reduction of noise can be more effective, generating a more accurate representation of the EMG signals produced (Richards, 2008).

The double differential configuration is comprised of three electrodes situated on a single unit. This setup works similarly to the bipolar/single differential, although since there are now three electrodes being compared the noise reduction is even more effective (Richards, 2008). An additional benefit of this is the reduction of pick up volume, which isolates surrounding muscles, along with the cross talk from them (Richards, 2008). However, the use of this configuration may not be suitable for large, superficial muscles, or where cross talk is not likely to occur, as the number of motor units being tested may be affected due to the reduced pick up volume (Richards, 2008). It is essential for all the above configurations that a grounding electrode be used. The purpose of this electrode is to provide the system with a reference point where voltages can be measured.

Various conductive metals including gold, silver/silver chloride (Ag/AgCl), stainless steel, and nickel can be used for the electrodes (Kamen and Gabriel, 2010; Richards, 2008). Furthermore, surface electrodes can be reusable or disposable. Reusable electrodes can be constructed with either bar or circular arrangements with nickel discs, or disposable electrodes using silver/silver chloride discs (Ag/AgCl) (Richards, 2008). Silver electrodes are used commercially in a clinical environment along with an electrolyte gel, which reduces resistance when recording (Pylatiuk et al., 2009; Tam and Webster, 1977).

sEMG electrode size should also be taken into consideration when being used. Selectively levels can be higher when using electrodes with smaller diameters and detection areas as any given electrodes can be placed closer together (Loeb and Gans, 1986). Smaller electrodes may have a diameter of 0.5 cm and should be used for muscles of the face and upper extremities. These electrodes can be placed 1 cm from each other. On the other hand, larger electrodes can be 1 cm in diameter and can have an inter-electrode distance of 2 cm (Cram, 2010).

Despite how versatile sEMG can be, there are limitations to consider. For instance, the surface electrodes will be fixed on one area of the skin while the muscle underneath is moving when contracting, potentially recording signals from a different area of the muscle than intended (Richards, 2008). Observing and ensuring the electrodes do not move over the myotendinous junctions or innervation zones as the muscle moves can reduce how much the signal is affected (Richards, 2008). Another limitation of sEMG is that the electrodes can only record reliable signals from superficial muscles (Richards, 2008). This is further exaggerated by the fact that subcutaneous tissue resists the input and output of electrical impulses (De Luca, 1997; Petrofski et al., 2008; Richards, 2008).

2.2.1.3 Skin Preparation

To obtain high quality EMG data, resistance must be taken into consideration. Skin resistance should not be too high because the accuracy of the signal will be reduced. On the other hand, low quality signals will be produced due to electrical shorting between the electrodes if the skin resistance is too low (Richards, 2008). An optimal resistance has not been agreed on, but it has been recommended that a resistance of 10 kΩ should be sought after (Richards, 2008). Additionally, the accuracy of the EMG signal below a frequency of 100 Hz has been shown to be reduced when the skin resistance is larger than 100 kΩ (Richards, 2008). The number of dead skin cells determines the amount of resistance that is provided, with the younger population having less than the older population (Richards, 2008).

Before applying the surface electrodes, it is recommended that the skin be prepared by, firstly, removing hair to improve the electrode adhesion to the skin (Konrad, 2006). The skin should then be cleaned to remove dead skin cells, oil, sweat and dirt. Abrading the skin with the use three to four strokes of fine sand paper will remove any dead skin cells. Cleaning the area with an alcohol wipe should be used in conjunction with the sand paper. Generally, good skin preparation can be indicated by the skin turning slightly red (Konrad, 2006). Preparing the skin in this way can potentially decrease the resistance to values as low as 5 kΩ from as high as 200 kΩ (Freivalds, 2011).

2.2.1.4 Signal Quality

The quality of a signal can be influenced by various factors; some more difficult to control than others. These factors can be categorised as causative, intermediate, and deterministic. Causative factors can be divided into intrinsic and extrinsic groups: extrinsic causative factors have a rudimentary effect on the EMG signal, such as electrode placement and structure, whereas intrinsic causative factors are anatomical, physiological, and biochemical features of a muscle (De Luca, 1997; Reaz et al., 2006). Intermediate factors are physical and physiological occurrences affected by causative factors, which then affect deterministic factors (De Luca, 1997; Reaz et al., 2006). For example, crosstalk from surrounding muscles and superposition of action potentials can be classified as intermediate factors (De Luca, 1997; Reaz et al., 2006). Deterministic factors have a direct influence on the information in EMG signals and force (De Luca, 1997; Reaz et al., 2006). Examples of this include the amount of active motor units, the firing rate of motor units, and the amplitude, length, and shape of action potentials (De Luca, 1997; Reaz et al., 2006).

When positioning sEMG electrodes the location of innervation zones should be taken into consideration. When an electrode is placed directly over an innervation zone, motor unit action potentials interfere with the EMG signal, meaning the muscle power spectra could be assessed incorrectly (Saitou et al., 2000).

It is essential that a reference, or grounding, electrode is used when using EMG. Its purpose is to reduce the amount noise that interferes with the relatively small signals being recorded. Because of this, the reference should be placed directly on tissue not affected by electrical impulses, which is usually over a bony prominence (De Luca, 2002).

2.2.2 Signals

Signals may appear in various forms including sine, cosine, square, triangle, and sawtooth waves. Henceforth, the term *signals* will be associated with the sine wave.

To understand signals several fundamental terms must first be understood. A *period (T)* can be defined as the amount of time taken for one full cycle of a given sine wave to be completed (Floyd and Buchla, 2013). Periods can be measured by using three methods, as shown in Fig.1:

- Measuring from one zero crossing to the next corresponding zero crossing.
- Measuring from one positive peak to the next.
- Measuring from one negative peak to the next.

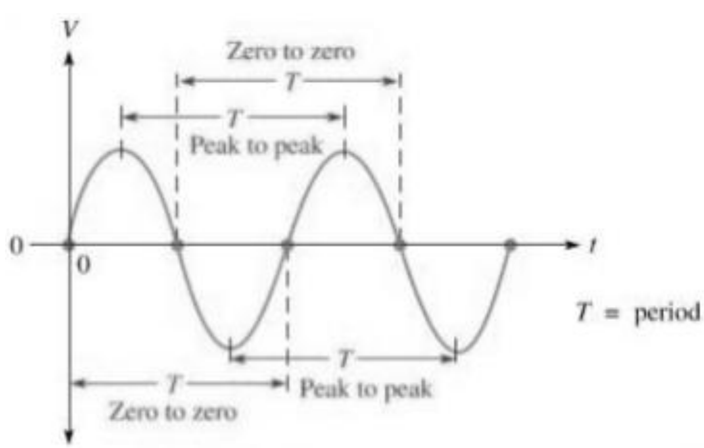

Figure 1. Sine wave period measurements (From Floyd and Buchla, 2013).

The *frequency* (f) of a sine wave is defined as the number of cycles completed in a single second (Floyd and Buchla, 2013) and is measured in Hertz (Hz). One Hz is equal to one cycle per second. Therefore, a lower frequency means less cycles per second and vice versa. The relationship between period and frequency can be represented as follows:

$$f = \frac{1}{T}$$

(2)

Or

$$T = \frac{1}{f}$$

(3)

The maximum positive and negative voltage of a sine wave, with respect to zero, is known as the *peak value* (Floyd and Buchla, 2013). Despite there being positive and negative peak values, a single peak value is used as the two are of an equal magnitude. This single peak value is known as the *amplitude* of the sine wave, represented by V_p and is measured from zero to the peak value (Floyd and Buchla, 2013). For example, the peak amplitude of the sine wave displayed in Fig. 2 is 10 V, while its peak-to-peak amplitude is 20 V.

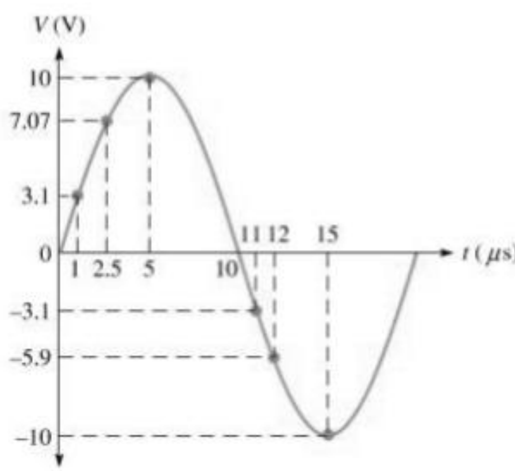

Figure 2. Instantaneous values of voltage, t = time (From Floyd and Buchla, 2013).

When recording EMG, it is common to experience *noise* that interferes with the wanted signals. Several types of noise can be identified (Reaz et al., 2006):

- Noise generated by nearby electronics which is unable to be removed but can be reduced with high-quality equipment.
- Ambient noise sourced from electromagnetic waves from the surface of the human body and the earth's surface.
- Motion artifact caused by the interface and wires of electrodes. Additionally, patient movement can present anomalies in the signal.
- Inherent signal variability caused by motor unit firing rates has a frequency typically ranging between 0 – 20 Hz. This noise can be removed by filtering.

2.2.3 Signal Processing and EMG analysis

The quality of EMG signals may be affected by various factors. For example, interference from noise can be assimilated while travelling through various tissue, causing the amplitude and frequency to attenuate (Kamen and Gabriel, 2010; Reaz et al., 2006). Additionally, electrodes, particularly surface electrodes, are prone to acquire signals from multiple motor units, meaning signals can overlap (Reaz et al., 2006). The importance of processing these signals is increasing, especially for clinical diagnostics, because the importance of managing and rehabilitating motor disabilities is recognised (Reaz et al., 2006). Signal processing techniques are generally used on raw EMG data to obtain three groups of information: time, frequency, and amplitude (Kamen and Gabriel, 2010).

Before EMG signals can be processed, the analogue voltages must be converted to digital signals by analogue-to-digital (A/D) conversion (Göker, 2014; Konrad, 2006). EMG systems may internally consist of an A/D converter, otherwise an external data acquisition board can be connected to accomplish A/D conversion (Pozzo et al., 2004). Additionally, an appropriate A/D sampling rate, or sampling frequency, should be selected. *Sampling* is the process of converting continuous-time signals, where values are uncountable and may consist of real numbers, to discrete-time signals, where values are countable and consist of natural numbers (Beerends et al., 2003). For a signal frequency spectrum to be accurately converted, the sampling rate of the input voltage must be at least twice the value of the greatest expected signal frequency (Konrad, 2006).

Data analysis generally consists of four steps. Before any filter can be applied to the raw signal, it should be ensured that the signal is centred around zero on the Y-axis. *Zero offset* occurs with the interference of external electrical signals, usually generated by nearby electrical equipment (Robbins, 2014). The occurrence of zero offset during the filtering process will generate inaccurate results. Confirming the presence of zero offset can be achieved by calculating the mean value of the signal. If the mean value calculated is not equal to zero, it should be subtracted from each data point to correct zero offset whilst preserving the signal symmetry (Haykin and Liu, 2009).

Low frequency noise interfering with the signal is an issue that can affect the zero offset correction process (Robbins, 2014). Therefore, this low frequency noise must be removed with the use of a filter. There are four main categories of filter (Enoka, 2015):

- High-pass filter – reduces low frequencies and maintains the high frequency element.
- Low-pass filter – reduces high frequencies and maintains the low frequency element.
- Band-pass filter – removes frequencies lower and greater than stated values, thus retaining a specific range of frequencies.
- Band-stop filter – reduces the incidence of a specific frequency or range. A notch filter is an example of a band-stop filter which removes one specific frequency.

Using this information, a high-pass filter should be applied to the signal if low frequency noise is present and vice versa.

Once the signal has been filtered it can then be rectified. The process of *rectification* involves removing all values below zero. This can be achieved by using one of two methods:

- Half Wave Rectification – Values below zero are removed. Despite being a simple method, this essentially halves the capacity of the signal.
- Full Wave Rectification – Each value is squared, then the square root of the new value is calculated. This has the outcome of converting all negative values to positive ones, retaining the capacity of the original signal.

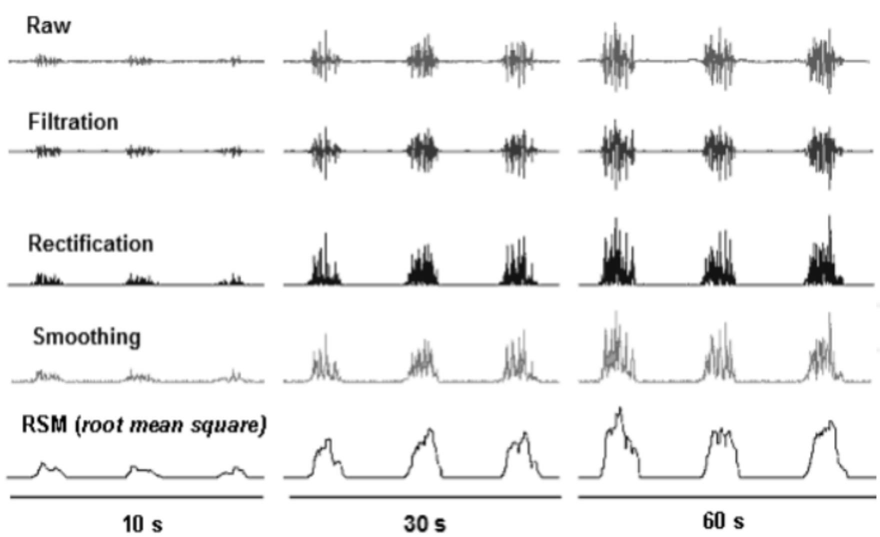

Figure 3. Stages of signal processing (From Altimari et al., 2012).

After the signal has been successfully rectified, signal smoothing can be completed. *Smoothing* is a technique used to identify the underlying pattern of the data. This smoothed signal is important as it clearly presents data without the interference of noise. One method of smoothing is known as the *Root Mean Square* (RMS). RMS uses a section of a signal, called a *window* or *linear envelope*, to convert every instantaneous value, both positive and negative, to positive values (Robbins, 2014). Conversion is achieved by squaring every value. The sum of every squared number is calculated and divided by the number of samples to result in a mean square. The square root of the mean square is then calculated to produce the RMS. Robbins (2014) summarises the process with the following equation:

$$RMS = \sqrt{\left(\frac{Sum\ of\ data\ points\ within\ window}{Length\ of\ window}\right)^2} \tag{4}$$

When specifying the window length, the speed of the movement and signal length should be taken into consideration. Rapid EMG changes from quick movements can be identified with small windows, with the cost of reduced smoothing of the original signal. Conversely, better signal smoothing can be achieved with a larger window, with the cost of losing trends of the original EMG signal. The benefits of using both a small and large window can be amalgamated by overlapping the two (Robbins, 2014).

2.2.3.1 Time and Frequency Domain

The time and frequency domains are, fundamentally, two perspectives of the same signal. In general, raw EMG data needs to be processed to form a reasonable understanding about what has been recorded. This is because raw EMG data appears as countless lines displaying the scale of a signal in the time domain, which can be difficult to interpret (Richards, 2008). For this reason, raw EMG data in the time domain are simplified using *frequency domain* processing (Soderberg, 1992). The frequency domain characterises a signal by representing the variation of frequency within (Sundararajan, 2001).

Processing EMG signals in the time domain is commonly achieved using the RMS or the average rectified values (De Luca, 1997). The RMS signifies the power of a signal, thus possessing a categorical meaning, which is appropriate for EMG signals identified during voluntary contractions (De Luca, 1997). However, there is no categorical meaning of an average rectified value as this only determines the area under the EMG signal (De Luca, 1997). Conversion from the time domain to the frequency domain is achieved using the *Fourier Transform* (Soderberg, 1992).

2.2.3.2 Frequency Analysis

Frequency distribution of muscle EMG can be easily analysed with the use of various software. The majority of sEMG frequency power is, generally, situated between 10 Hz and 250 Hz (Konrad, 2006). The *Fast Fourier Transform* (FFT) is used to calculate the power distribution and is visualised as a total power spectrum, where frequency power distribution, or magnitude, is displayed on the Y-axis and frequency (Hz) is displayed on the X-axis (Konrad, 2006).

2.2.4 Applications of EMG

The applications of EMG beneficial in many fields including neurophysiological and medical research, rehabilitation, ergonomics, and clinical diagnostics to name a few (Barbero et al., 2012). Moreover, EMG has uses in several branches of biomechanics (Konrad, 2006), including:

- Anthropometry - the measurement of body segments and proportions.
- Kinematics - the measurement of distance, angles, velocity, and acceleration.
- Kinetics - the measurement of forces.
- Kinesiological EMG - the measurement of muscle activity levels.

In neurophysiological and medical research EMG can be utilised for functional neurology (Barbero et al., 2012); this approach investigates subtle nervous system deviations before a clear pathology become evident. Examples of pathologies include multiple sclerosis, restless leg syndrome, and tremor disorders. EMG can be used in assessing functional neurology as the location and the degree of the dysfunction needs to be assessed. In addition to this, the effects ageing on muscle performance can be assessed with EMG (Barbero et al., 2012). The technique can measure muscle strength and compare values to the average of the same age. These results can then be used to determine if further training is necessary.

There are many applications of EMG in the rehabilitation field. For example, EMG is often used to monitor the effectiveness and progression of a patient's treatment. This can help to adapt and tailor treatment plans to the specific needs of the patient to optimise recovery rates. Progress of the rehabilitation can be observed by comparing the muscle activation pre- and post-injury. This continuous assessment can help practitioners decide if the current plan should continue to be followed or if a new treatment strategy might be more suitable. Additionally, those who have undergone amputation may find benefits of EMG for controlling paralysed limbs from an external source or the use of prosthetics (Barbero et al., 2012).

With regards to the ergonomics, EMG can be employed to analyse the impact of workplace design for a more ergonomic work environment. EMG can assess how effective office chairs are at supporting back muscles by placing electrodes at various sites on the back while the individuals work in various positions for differing durations. Linked to this type of application, risk prevention in ergonomics using periodic monitoring of individual's posture could potentially detect the early stage development of postural disorders (Barbero et al., 2012).

With the use of EMG in clinical diagnostics, several neuromuscular diseases can be identified as the signals differ from the norm. These atypical waveforms have characteristics which may relate to certain pathological conditions. Therefore, each disorder can be related to a specific signal deviation. The source of neuromuscular diseases may stem from a nerve cell disorder, disruption of the action potential delivery from the axon, the motor unit failing to transmit neuromuscular signals, defects in the sarcolemma, or general muscle imperfections (Najarian and Splinter, 2012). The lack of nervous excitation often causes an EMG signal to be partially or even completely absent. The capability of a muscle to receive action potentials will be reduced if a nerve cell between its motor unit and the spinal cord has degraded at numerous sites (Najarian and Splinter, 2012).

EMG signals produced by those who are affected with a neuromuscular condition that impair motor unit structure and function will appear altered. These signals may indicate the presence of abnormal spontaneous activity or abnormal voluntary electrical activity (Kimura, 2001; Rubin, 2009). Some examples of neuromuscular diseases include myopathy, Parkinson's disease, and anterior horn diseases including amyotrophic lateral sclerosis (ALS)/motor neuron disease (Najarian and Splinter, 2012). For each of these diseases EMG can be used to determine the type of nerve damage involved.

Additionally, fatigue can be measured with the use of sEMG as the frequency of muscles has been shown to decline with force over time. Muscle fatigue is an occurrence from persistent maximal or sub-maximal contractions where muscle force generation or maintenance is reduced (Bigland-Ritchie et al., 1983). Fatigue in muscles is visible in their EMG as noticeable reductions can be seen in amplitude, power, and frequency (Najarian and Splinter, 2012). As more fatigue becomes apparent it has been shown that the frequency spectrum shifts from higher to lower frequencies due to muscle fibres experiencing reduced conduction velocity (Arendt-Nielson and Mills, 1988; Conforto et al., 2013).

2.3 Controlled Modulation of Force

One factor which influences muscle function is the varying types of muscle fibres. The largest components are fascicles which are made up of smaller muscle fibres (MacIntosh, Gardiner and McComas, 2006). Muscle fibres are classified as one of three types, each with differing properties: type I (slow twitch), type IIA or type IIB (both fast twitch) (Draper and Marshall, 2014). Draper and Marshall (2014) summarise the muscle fibre types by stating that type I fibres, are red in colour due to their high myoglobin content, which provides a large oxygen storage (Ordway and Garry, 2004; Shankman and Manske, 2011), have a slow rate of contraction, relatively low contraction force and are highly resistant to fatigue. On the contrary, type IIB fibres have a very fast rate of contraction, relatively high contraction force, low fatigue resistance, but are red in colour, like type

I fibres. Finally, type IIA muscle fibres are situated between these two by possessing fast contraction speeds, but medium contraction force and fatigue resistance. Knowledge of the differences between the fibre types could begin to provide an explanation for any differences of EMG-force relationships, should any be found.

Predominant muscle fibre types can be predicted with a simple test: the one repetition maximum (1RM) is first established. After this, as many repetitions as possible at 80 per cent of the 1RM should be completed. It is suggested that if less than seven repetitions were completed the muscle group may be composed of more than 50 per cent fast twitch fibres. If between seven and twelve repetitions were completed the proportion of slow twitch and fast twitch fibres may be equal. Finally, if more than twelve repetitions were completed the muscle group may consist of more than 50 percent slow twitch fibres (Pipes, 1994).

As previously mentioned, there are currently two widely accepted theories of force modulation in *Henneman's Size Principle* (Henneman et al., 1964) and *rate coding* (Enoka, 2015). Henneman's Size Principle states that when under load, smaller motor units are recruited before larger ones (Henneman, Somjen and Carpenter, 1964). A motor unit is said to activate more frequently as muscle tension increases (Winter, 1990). As tension increases and the motor unit approaches its maximum firing rate, larger motor units are recruited and begin firing (Winter, 1990). Once muscle tension decreases the larger motor units stop firing first (Winter, 1990).

With regards to muscle fibre types, type I, or slow twitch, are recruited first as they typically reside in small motor units (Freidvalds, 2011). The recruitment of type I fibres first allows for the initial force production to be precise relative to the action taking place without exceeding the necessary force required (Freidvalds, 2011). The recruitment of type I fibres is followed by the relatively medium-sized type IIA fibres, then finally the larger type IIB, fast twitch fibres (Freidvalds, 2011).

Rate coding is the control of the rate at which action potentials are discharged from motor neurons (Enoka, 2015). Rate coding has presented a strong, linear relationship between action potential discharge rate and force production (Enoka, 2015). Motor units each have discharge rates, which can be described as the rate, as pulses per second (pps), at which action potentials are discharged from neurons (Enoka, 2015). Motor neurons with higher thresholds are believed to have greater firing rates than those with lower thresholds (De Luca and Contessa, 2015). The idea was based on an observation that a lower-threshold motor neuron with a smaller diameter presents a longer after-hyperpolarisation and lesser firing rates than those with a higher-threshold and a larger diameter (De Luca and Consetta, 2015). Minimal discharge rates of motor units with low-thresholds can range from 5 to 8 pps and 10 to 15 pps for high-thresholds (Enoka, 2015). Additionally, motor units have maximal discharge rates ranging from 20 to 60 pps during a gradual contraction but can reach up to 120 pps during a short, rapid contraction (Enoka, 2015).

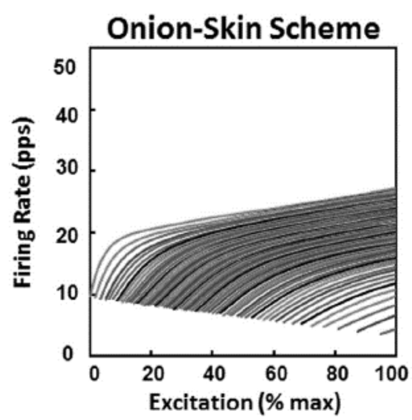

Figure 4. Onion-Skin model for vastus lateralis. Simulated firing rate spectrum with increasing excitation to motorneurons is displayed (From De Luca and Contessa, 2015).

When sEMG records four or more motor unit action potentials, the individual signals become harder to distinguish due to increased levels of superposition (De Luca et al., 2006). dEMG is a technique which acquires specific motor unit action potentials with the breakdown of sEMG signals (De Luca et al., 2006). This information may be useful when investigating the firing rates of individual motor units during various contraction intensities. The use of dEMG has led to the development of a concept referred to as the onion-skin model. Studies have shown that motor units which have been recruited earlier maintain greater firing rates than those recruited later (Fig. 3).

2.4 Influences on Muscle Function

2.4.1 EMG-Force Relationship

Movement at any joint is reliant on both joint range of motion (RoM) and muscle length (Reese and Bandy, 2013). RoM is the movement available at a joint and is affected by bony structures and connective tissues, such as ligaments, at the end range. On the other hand, muscle length allows a joint, or joints if the muscle is biarticular, to move through its RoM.

The relationship between EMG amplitude and the force generated by a muscle can provide a clearer understanding of the amount of neuromuscular activation occurring during an isometric contraction (Saito and Akima, 2013). However, due to the limited number of studies conducted researching this, this understanding may not be clear for some muscles as there have been conflicting results regarding the EMG-force relationship being linear or non-linear. A linear relationship is one where EMG amplitude steadily increases as force increases. On the contrary, the amplitude in a curvilinear relationship begins to increase at a faster rate as force increases. Another potential outcome is the elbow point relationship. This is where two linear segments of different gradients converge at one point. This means that the same level of muscle activity may begin at a different contraction intensity for a muscle with a linear compared to one with a curvilinear or elbow point relationship. From a functional perspective, clinicians can use these different relationships as guidelines for determining how much a muscle should be contracted to incite a specific level of activity for targeted muscles.

The linearity of the EMG-force relationship has been shown to rely of the relationships between motor unit recruitment and firing frequency (Solomonow et al., 1990). Linear relationships have been observed when motor unit firing frequency ensues after full motor unit recruitment. On the other hand, a non-linear relationship is observed when motor unit firing frequency and recruitment occur concurrently.

The EMG-force relationship of various muscles has been investigated to some degree over the past 70 years. For example, soleus (Lippold, 1952; Close et al., 1969; Bigland-Ritchie, 1981) and rectus femoris (Komi and Vitasalo, 1976; Alkner et al., 1981) have consistently presented a linear relationship.

On the other hand, biceps brachii (Zuniga and Simons, 1969; Bigland-Ritchie, 1981; Lawrence and De Luca, 1983; Praagman et al., 2003), first dorsal interosseous (Milner-Brown and Stein, 1975; Bigland-Ritchie, 1981; Lawrence and De Luca, 1983; Madeleine et al., 2001), and brachioradialis (Bigland-Ritchie, 1981; Praagman et al., 2003) have not presented consistent linear relationships. The reasoning for the different relationships is not fully understood, but it has been suggested that muscle physiology is the key factor affecting them (Bigland-Ritchie, 1981).

The effect of joint angle on EMG-force relationship has been studied on several occasions. For instance, elbow angle has been reported to have no significant effect on the EMG-force relationship of biceps brachii and brachioradialis during flexion (Doheny et al., 2008). Likewise, the relationship for triceps brachii during extension was not significantly different between 45° - 120° of flexion (Doheny et al., 2008). Despite this, angles of 10° and 30° of flexion had a significant effect, recording greater EMG amplitudes at low level of force (Doheny et al., 2008). The EMG-force relationship of the quadriceps group during extension has also been shown to be significantly affected by knee joint angle (Saito and Akima, 2013). For each muscle in the quadriceps group, lower EMG amplitudes were recorded at 30° of flexion than 60° and 90° at each 20, 40, 60, 80, and 100 per cent of exerted force (Saito and Akima, 2013).

It should be noted that previous studies may have reported relationships to be curvilinear or non-linear while they were, in fact, an elbow point. If this was the case, this could potentially provide an explanation for differences between muscles with reported curvilinear relationships. It can be suggested that relationships be carefully examined to determine linearity before conclusions are made (Table 1).

2.4.2 EMG-Length Relationship

The relationship between muscle length and EMG activity has been investigated considerably, although yielding dissimilar results. As previously stated, the length-tension relationship is represented by an Inverted-U curve, whereby sarcomeres in shortened and lengthened states generated less force than when at their optimal length (Friedvalds, 2011). Despite this, EMG activity has been shown to be relatively unaffected by changes in joint angle (Brownstein et al., 1985; Eloranta and Komi, 1981; Leedham and Dowling, 1995; Vredenbregt and Rau, 1973). Muscles while in a lengthened state have displayed a decrease in EMG activity, although this may have been due to fatigue or varying participant effort (Andriacchi et al., 1984; Heckathorne and Childress, 1981; Leedham and Dowling, 1995; Lunnen et al., 1981; Soderberg and Cook, 1984).

It has been suggested that the relationship between torque and EMG activity can be an indication of energy efficiency (Mohamed et al., 2002). When a high torque is output with low EMG activity, this can be considered as more energy efficient than the vice versa (Mohamed et al., 2002).

2.4.3 Force-Velocity Relationship

Hill's muscle model describes muscle tension using both active and passive components. The Hill muscle model consists of two components in series and one component in parallel (Knudson, 2007). The active tension of muscle is represented by the contractile element, whereas the passive tension is represented by the parallel elastic component and series elastic component, as shown in Fig. 4 (Knudson, 2007).

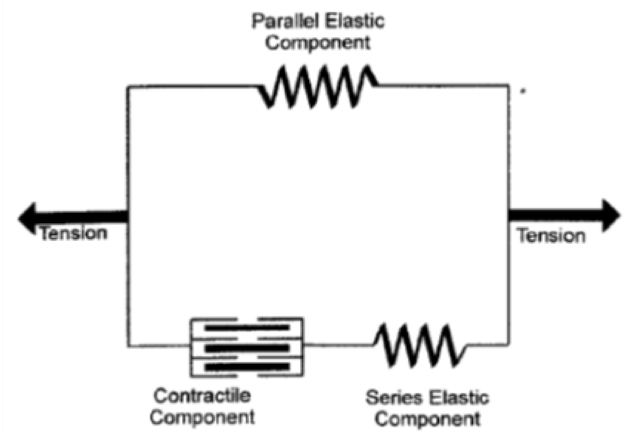

Figure 5. Hill muscle model (from Knudson, 2007, pp.53).

Muscle contraction intensity is heavily influenced by the rate of change of muscle length. As the velocity of a movement during a concentric contraction increases it has been shown that muscle force decreases (Epstein and Herzog, 1998; Freidvalds, 2011). There are two explanations for this occurrence: first, an increase in velocity reduces the efficiency of cross-bridge bonding between sliding filaments and, secondly, muscle tissue being affected by damping (Freidvalds, 2011). On the contrary, an increase in velocity during an eccentric contraction means force exertion is also increased, and vice versa. Concentric and eccentric contractions are considered to have a positive and negative velocity, respectively, while isometric contractions have zero velocity (Fig. 5).

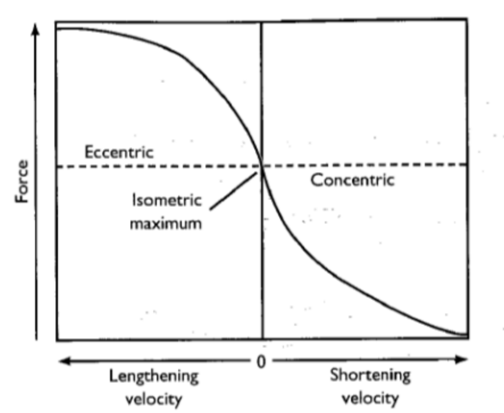

Figure 6. Force-Velocity relationship of skeletal muscle (from Hall, 2015, pp.160).

2.4.4 Length-Tension Relationship

An important characteristic of skeletal muscle is its capability to exert force relative to its length (Maganaris, 2001). The length-tension relationship is instigated from sarcomeres within muscle fibres which, at a joint, affect the relationship between torque and position (Kawakami et al., 2000).

The amount of muscle force produced is dependent on a muscle's shape. pennation angle (PA), physiological cross-sectional area (PCSA), ratio of fibre types, joint structure, and the tendon attachment on the bone (Kawakami et al., 2000). Generally, muscles with a large PA and PCSA are better suited for force production and endurance activities. On the other hand, the purpose of muscles with a small PA and PCSA is shortening velocity for rapid movements (Fukunaga et al., 1992; Kenney, Wilmore and Costill, 2015). Regarding muscle fibre type, a greater predominance of type I fibres indicates the primary purpose of a muscle is force production over shortening velocity. In contrast, more type II fibres suggest a muscle is used for shortening velocity over force production (Kenney et al., 2015).

The underlying physiology of the triceps surae group has been the focus of many studies over the years. Specifically, research examining the percentage of fibre types (Edgerton, Smith and Simpson, 1975), PAs, and fibre lengths (Fukunaga et al., 1992; Kenney, Wilmore and Costill, 2015) have been key areas of study. Fukunaga et al. (1992) established that muscles such as soleus, with a large PCSA, large PAs, and small fascicle lengths prioritise tension production. Additionally, the primary purpose of muscles, such as lateral gastrocnemius, with a small PCSA, small PAs, and large fascicle lengths is generating velocity over tension (Fukunaga et al., 1992).

Changes in sarcomere length, as described by the Sliding Filament Theory (Huxley, 1971), plays a key role in the length-tension relationship. When a muscle is in a shortened or lengthened state less tension is generated than when at its optimal length (Friedvalds, 2011). A systematic review by Burkholder and Lieber (2001) determined the optimal sarcomere length in human skeletal muscle to be 2.64 µm, where optimal sarcomere length was defined as the length which generated the most tension (Fig. 6).

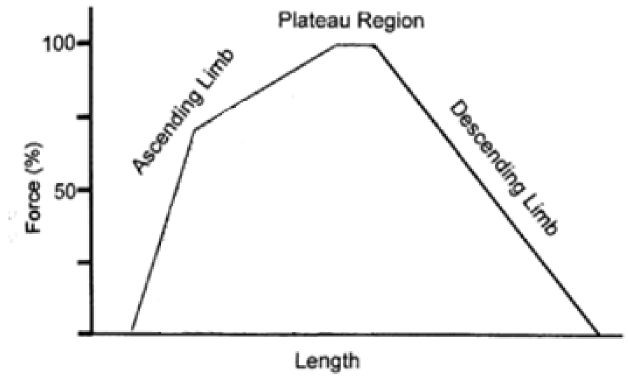

Figure 7. Length-Tension relationship (adapted from Kundson, 2007, pp.85).

The decrease in tension generation is explained by the number of cross bridges formed in each state. For instance, as a muscle lengthens the tension decreases linearly due to less cross bridges being established, until eventually none can be formed (Friedvalds, 2011). Contrary to this, as a muscle shortens tension decreases due to actin filament overlap (Friedvalds, 2011). Furthermore, myosin filaments begin to collide with the titin and Z-disks, physically preventing any movement occurring and, therefore, tension is unable to be generated (Friedvalds, 2011).

2.5 Applications of the Current Study

Fundamentally, an application of current study is to expand on and be a source of informative research. Previous research has not investigated the effects of numerous muscle characteristics simultaneously. Following this, there is no previous research which has studied the EMG-force relationship of 15 muscles over three joints at one time. The methods and results of the current study are envisioned to deliver information on the different EMG-force relationships of various muscles and provide explanations for the results.

The current study could also be used as a basis to future clinical research. The methods exercised, and the limitations involved can be considered and improved upon to obtain more results to expand on this study. This could allow for potential future research to develop improved rehabilitation protocols.

In a rehabilitation environment, clients may be required to contract targeted muscles at a specified intensity to incite an anticipated level of activation. This is done to assist the healing process and strengthen the muscle in question to reduce the likelihood of potential future injury. In particular, those who have undergone surgery or sustained muscular injury could have reduced strength and may need to regain an appropriate level of neuromuscular control. An example of a technique used for rehabilitation is Proprioceptive Neuromuscular Facilitation (PNF) stretching which aims to restore RoM and increase strength (Hindle et al., 2012). However, if the client is asked to contract at 40 per cent of their maximum, for example, and their perception of this is inaccurate relative to any pre-measured values using equipment such as an isokinetic dynamometer, the effectiveness of the treatment may become hindered by causing further harm.

As well as the client misperceiving their contraction intensity, clinicians may be unaware of the differences of EMG-force relationships between muscles. With regards to the relationship, one muscle may present with linear relationship while another may present with a curvilinear relationship. If a clinician assumes all muscles possess a linear EMG-force relationship and asks a client to contract a muscle at a specific intensity, the predicted level of muscle activation of any muscle with a curvilinear relationship will be inaccurate. This could mean that the desired level of neuromuscular control will not be achieved as quickly or prominently as expected.

The data obtained from this study should provide an accurate guide to how much muscle activation is achieved from a specific contraction intensity. Practitioners will be able to refer to the graphs from this study to quickly and easily establish the correct amount of force to be applied. It is also envisioned that the current study will act as a foundation for future studies to further research additional variables and populations. The current study gathered healthy volunteers to participate, although there is potential that a population with similar injuries or genetic disorders may present with differing EMG-force relationships.

2.6 Aims of the Current Study

The primary purpose of this research is to determine the type of EMG-force relationship each muscle in question possesses and investigate reasons as to why these are occurring. Additionally, frequency analysis will be conducted, compared, and discussed in comparison to previous research.

The secondary purpose of this research is to gather data on various muscle characteristics spanning years of previous research. This will mean the data will be compiled in one convenient document with explanations and conclusions for characteristics that do and do not appear to influence the EMG-force relationship.

3. Methods

3.1 Ethics

Ethical approval was gained from the Institute of Health Research (IHR) Ethics Committee. Participants were required to read an information sheet and complete a medical screening form and informed consent form before testing. Storage of all participant information complied with the Data Protection Act (1998). This includes participant information being destroyed once the study was completed, as well as keeping information and data collected anonymous. Only researchers participating in this study had access to this information. Participants had the freedom to withdraw from the study at any time. If data had already been collected from the participant, it was excluded during data analysis.

3.2 Participants

For this study, 31 healthy participants were recruited, of which 19, 6, and 6 had all three, two, and one joint(s) tested, respectively, totalling 25 of each joint tested (Table 1). The recruitment process consisted of convenience sampling whereby students and staff at the University of Bedfordshire were asked to participate.

Table 1. Participant demographics.

Joint	Number of Participants	Males	Females	Age (Years)	Height (m)	Mass (kg)
Ankle	25	13	12	24.08 ± 4.70	1.71 ± 0.08	67.03 ± 17.17
Elbow	25	12	13	24.80 ± 4.83	1.71 ± 0.08	67.96 ± 18.35
Knee	25	15	10	24.16 ± 4.36	1.72 ± 0.08	71.65 ± 24.29
Overall	31	18	13	24.55 ± 4.64	1.72 ± 0.08	72.08 ± 23.67

3.3 Instrumentation

The Isokinetic Dynamometer (Biodex 3, Suffolk, UK) was connected to an NI USB-6343 Multifunction Data Acquisition (DAQ) unit (National Instruments, Newbury, UK) via three coaxial cables relaying torque, velocity, and position. All channels were set to floating source.

Six sEMG sensor SX230 electrodes (Biometrics, Newport, UK) were connected to analogue channels one through six of a DataLINK Subject Unit (Biometrics, Newport, UK) when testing the ankle, channels one through five when testing the knee, and channels one through four when testing the elbow.

Data was transferred to a DataLINK K800 amplifier (Biometrics, Newport, UK) from the DataLINK Subject Unit via an R6000 data transfer cable (Biometrics, Newport, UK). In addition to this, analogue outputs from the amplifier were transferred to the DAQ unit via an R2000i transfer cable. The DAQ unit relayed data from the amplifier and isokinetic dynamometer to a laptop via USB, where signals were recorded in SignalExpress (v. 15.0.0, 2015, National Instruments, Newbury, UK). EMG data recorded in Signal Express were converted and saved as ASCII files to be analysed later.

3.4 Equipment Validation

Before data could be collected, the testing environment needed to be arranged for optimal quality EMG signals. To begin with, there was a considerable amount of noise interfering with the EMG. This issue needed to be resolved as it would have been very difficult to use the data collected.

The noise could have been caused by nearby electrical equipment, electrical impulses from the heart, or from wires crossing over (De Luca, 1997). Several steps were taken to identify the source and address this issue. Biodex, the manufacturers of the isokinetic dynamometer, were contacted for advice as their machine was the largest piece of electrical equipment in the testing environment. It was suggested that a different power source may reduce noise levels. Considering this, the power sources of the DataLINK and NI DAQ board were moved from the wall to the isokinetic dynamometer itself. Furthermore, Biodex serviced and passed the dynamometer, which ensured there were no faults with the machine.

In addition to this, a series of tests were conducted to identify any other potential source of the noise in the data collection room. The noise tests consisted of recording EMG data for three seconds under various conditions, including different rooms, lights on/off, DataLINK plugged into different wall sockets, any surrounding electrical devices switched on/off, and the use of a reference electrode/electrocardiography (ECG) electrode. An electrode was placed on the right biceps brachii during each test with the reference electrode around the left wrist, or ECG electrode placed on C7 of the spine.

Each test was recorded for three seconds and saved as ASCII files with engineering unit conversion applied. The values in each file were converted into graphs, where the first 500 data points were selected to represent the overall data (Appendix 2). Points of interest included the frequency and amplitude of the signals.

3.5 Use of Mid-Range Validation

One volunteer agreed to participate in the validation testing. To justify the use of the mid-range, the right ankle, knee, and elbow RoM (°) were measured using a clinical goniometer (MIE Medical Research Ltd., Leeds, UK). Three angles were calculated from the measured RoM: the midrange = total RoM / 2, the midrange * 0.5 which favoured extension/dorsiflexion, and the midrange * 1.5 which favoured flexion/plantarflexion.

The isokinetic dynamometer arm was then positioned for each joint by taking the joint to end range, setting the clinical goniometer to zero, and then moving the joint in accordance with the calculated values. Three MVCs for each antagonistic movement were completed. Each repetition lasted for three seconds with a rest period of thirty seconds. Peak torque (Nm) was recorded and compared at each angle for each joint (Appendix 3). As this has previously been researched (Algere et al., 2014) no additional analysis was conducted.

The results showed that in all but one instance torque produced at Mid * 0.5 and Mid * 1.5 was greater or less than the torque produced at midrange. This reiterated the theory of the length-tension relationship; force production is influenced by muscle length. For this reason, the midrange was chosen as the angle of interest during data collection.

3.6 Protocol

This research was a cross-sectional study. Muscle function of the right elbow, knee, and ankle was assessed. The right side was chosen to be tested over the left or dominant side as previous research had only used the right side (Lawrence and De Luca, 1983; Madeleine et al., 2001; Praagman et al., 2003). Individual testing sessions lasted up to an hour per joint. Participants were required to be tested on at least two separate days due to the amount of time testing took, as well as possible leg fatigue if knee and ankle were tested on the same day (ACSM, 2013). Before any data was collected, a researcher measured the participants' height (m) and body mass (kg) with a Seca 213 Stadiometer Height Measure (Seca, Birmingham, UK) and a Seca 813 Robusta Electronic Scale (Seca, Birmingham, UK), respectively. This data was recorded in a Microsoft Office Excel (2016, Microsoft, USA) spread sheet.

The isokinetic dynamometer was set up in accordance with the joint tested. Participants were seated upright in the isokinetic dynamometer when testing all joints. Ankle testing involved participants resting their leg on a padded support while the ankle was secured in-line with the axis of movement at an average midrange angle of 37.6°. The shank was parallel to the floor (Appendix 4). For elbow measurements, the upper right limb was secured to a padded support while the elbow was, on average, in 79.6° flexion (Appendix 5). When testing the knee, the knee was aligned with the axis of movement and the participants' shanks were secured to the movement arm of the dynamometer (Appendix 6). The knee was fixed at an average flexion angle of 75.4°.

First, a warm up period consisting of ten repetitions of isokinetic contractions at a rate of 120° per second (Jee, 2015) for each muscle in the antagonistic pair against very low resistance was provided to the participants. The isokinetic dynamometer was set up per the joint that was to be tested, with each joint being fixed at midrange. Midrange of each joint was determined by measuring the maximum RoM using a clinical goniometer and dividing the value by two. Measuring the mid-range of each individual participant was opted for over choosing a set mid-range. The latter may have given a mechanical advantage or disadvantage to a muscle over its antagonistic partner as every participant would likely have a different RoM. Additionally, this was done for each participant as muscle physiology or tightness may have affected RoM. Additionally, ageing has been shown to affect RoM, therefore, potentially influencing results if a predetermined midrange angle was used instead (Farley et al., 2006; Roach, 2001).

sEMG electrodes were placed on TA, GM, GL, Sol, FL, and FB for the ankle, RF, VM, VL, BF, and ST for the knee, and both heads of BB, TB Long, TB Lateral, and BR for the elbow, all in accordance with the SENIAM recommendations (Hermens et al., 2000). A reference electrode was placed around the right wrist when testing all joints (Hermens et al., 2000). Participants were required to complete a maximal voluntary contraction (MVC) by performing a separate sustained contraction of each muscle in an antagonistic pair against the resistance offered by the isokinetic machine, for five seconds each. Isometric contractions were then completed with each joint being contracted at 20, 40, 60, and 80 per cent of the MVC in a randomised order. Randomisation of joint and force order was included to reduce the risk of order effect and the influence of fatigue. Percentages of the MVC were determined by using the maximum torque (Nm) achieved during the MVC * 0.2, * 0.4, * 0.6, and * 0.8 for 20, 40, 60, and 80 per cent, respectively. A visual representation of the force to be applied against resistance was provided on a monitor for participants to aim for. Two lines were shown: the upper line was the value to aim for while the lower line was a 10 per cent buffer. Participants were to aim their torque bar between the two lines as accurately as possible.

To reduce the risk of fatigue and potential injury, after each set of contractions participants were provided with a rest period of three minutes as the activity involved resistance exercises (ACSM, 2013).

An International Physical Activity Questionnaire (IPAQ) was completed retrospectively by each participant to determine their level of physical activity during the data collection period. Participants were grouped into high, moderate, and low activity levels. Table 2 displays how many participants were group into each level of physical activity.

Table 2. Number of participants in each category of physical activity levels per joint.

Joint	Low	Moderate	High
Ankle	2	6	15
Elbow	2	6	15
Knee	2	5	16
Overall	3	6	19

3.7 Data Analysis

EMG signal analysis was completed in Visual 3D (v.6.01.07, 2017, C-Motion, Germantown, MD). Raw EMG data were originally sampled at 2000 Hz. EMG signals were smoothed using Root Mean Square (RMS) with a window size of 500 data points. Each signal was multiplied by a constant of two and then subtracting a constant of four. Then, the mean instantaneous values of the signals from each muscle were then calculated and subtracted from the aforementioned values to correct for zero offset. Signal-to-Noise ratio (SNR) was shown to be high, therefore no band-pass filter was applied to the EMG data.

A lowpass Butterworth filter with a cut-off frequency of 6 Hz was applied to torque data associated with each joint movement. A small window of torque data during a resting period was used to calculate a mean value, which was subtracted from each torque value to correct for zero offset.

The EMG data of each muscle were coupled with the respective torque data; EMG data of flexor and extensor muscles were associated with torque data for flexion and extension, respectively. Two markers were set 512 data point either side of the peak torque, to obtain 1024 data points, of each MVC which were used to calculate a mean value. This process was repeated for every contraction at each intensity to obtain mean EMG and torque values. Global mean values were calculated for MVC percentage contractions four repetitions were completed for each antagonistic movement per set. The mean values calculated were exported to a Microsoft Excel spreadsheet

Mean values were converted into percentages of the MVC for each participant were calculated by dividing each value by the MVC and multiplying by 100. The same process was used to convert torque data into percentages of the MVC.

Frequency analysis was performed on the data using MATLAB (v.8.5, 2015, MathWorks, Massachusetts, USA) (See Appendix 7 for MATLAB code used). 1024 data points of each repetition of each set performed used to calculate mean values previously were loaded into MATLAB. As stationary signals were produced from isometric contractions the FFT was used to convert time domain data into frequency domain data. Once data was converted into the frequency domain the median frequency was calculated. Median frequencies were produced for every contraction at each intensity for each participant. These were exported to a Microsoft Excel spreadsheet where the mean was calculated for individual participants at each contraction intensity. A median and mean value were then calculated, where the median was plotted on the EMG-force relationship graphs.

Some data were unusable due to them being of poor quality, so were therefore excluded from the final EMG, torque, and frequency data to reduce the influence on the final results.

4. Results

Table 3 shows that the R^2 value for every muscle is at least 0.93 when a linear trendline is used. The coefficient of determinations of each muscle using a linear trendline were all extremely close to 1 that using a polynomial trendline offered negligible difference. For example, when second order polynomial trendline was used the largest increase of an R^2 value was by only 0.033. Therefore, if a polynomial trendline was used it would become even more difficult to distinguish between linear, curvilinear, and elbow point. With regards to the linear trendline and the data from the current study, an R^2 value between 0.930 - 0.979 indicates an elbow point relationship, between 0.980 – 0.989 indicates a curvilinear relationship, and ≥ 0.990 indicates a linear relationship.

Table 3. Coefficient of determinations of linear and second order polynomial trendlines for each muscle.

Joint	Muscle	Coefficient of Determination (R^2) of Trendline		
		Linear	Second Order Polynomial	Difference
Ankle	Fibularis Brevis	0.999	1.000	0.001
	Fibularis Longus	0.992	0.997	0.004
	Gastrocnemius Lateralis	0.964	0.998	0.033
	Gastrocnemius Medialis	0.968	0.996	0.028
	Soleus	0.985	0.999	0.015
	Tibialis Anterior	0.983	1.000	0.017
Elbow	Biceps Brachii	0.965	0.972	0.007
	Brachioradialis	0.930	0.940	0.010
	Triceps Brachii - Lateral Head	0.995	0.997	0.002
	Triceps Brachii - Long Head	1.000	1.000	0.000
Knee	Biceps Femoris	0.987	1.000	0.013
	Rectus Femoris	0.968	0.997	0.029
	Semitendinosus	0.997	0.998	0.001
	Vastus Lateralis	0.971	0.988	0.017
	Vastus Medialis	0.967	0.991	0.024

Using the results from the IPAQ participants were grouped into high, moderate, and low levels of physical activity. EMG-force relationships with standard deviations were generated and a linear trendline was applied to each. The R^2 values of each trendline were determined and summarised in Table 4.

Table 4. Coefficient of determinations of linear trendlines for each muscle. Participants categorised by activity levels.

Joint	Muscle	Coefficient of Determination (R^2) of Linear Trendline		
		High Activity	Moderate Activity	Low Activity
Ankle	Fibularis Brevis	0.999	0.976	0.962
	Fibularis Longus	0.995	0.969	0.744
	Gastrocnemius Lateralis	0.969	0.941	0.958
	Gastrocnemius Medialis	0.972	0.965	0.778
	Soleus	0.983	0.977	0.986
	Tibialis Anterior	0.987	0.948	0.999
Elbow	Biceps Brachii	0.964	0.976	0.995
	Brachioradialis	0.943	0.941	0.915
	Triceps Brachii - Lateral Head	0.999	0.957	0.988
	Triceps Brachii - Long Head	0.999	0.993	1.00
Knee	Biceps Femoris	0.976	0.981	0.939
	Rectus Femoris	0.978	0.944	0.967
	Semitendinosus	0.990	0.990	0.994
	Vastus Lateralis	0.983	0.931	0.956
	Vastus Medialis	0.968	0.978	0.954

A visual inspection of each graph and plotting of a linear trendline with its equation and coefficient of determination (R^2) were used to determine the linearity of an EMG-force relationship. Using this method, five muscles were classified as having a linear relationship, three as curvilinear, and seven as elbow point, as summarised in Table 5.

For each graph displayed force and EMG amplitude are presented as a percentage of their respective MVCs. Standard deviations were calculated on the percentage values meaning 100 per cent contraction intensity has a deviation of 0 for the EMG-force relationships, both vertically and horizontally. This does not represent raw data.

The EMG-force relationship of GM and GL is shown to be an elbow point at 60 per cent of the MVC, as shown in Fig. 8a and Fig. 9a, respectively. On the contrary, Sol is shown to possess a curvilinear EMG-force relationship, as shown in Fig. 10. Moreover, Fig. 11 and Fig. 12 shown the relationship of both FL and FB to be linear, respectively.

With regards to the power spectrum, an elbow point is observed at 60 per cent of the MVC for the frequency of GM, which ranges from 126 – 153 Hz. On the other hand, GL has no elbow point but, instead, is shown to generally decrease with increases contraction intensity, ranging from 133 – 153 Hz. Similarly, the frequency observed with Sol decreases with increasing force, although it initially increases. The frequency range of Sol is 123 – 143 Hz. Contrary to the triceps surae, the frequency of FB ranges from 112 – 136 Hz and FL from 116 – 130 Hz, generally increasing as force increases. Fig. 13 shows the EMG-force relationship of TA to be curvilinear. Likewise, the frequency of TA increases as contraction intensity increases, ranging from 104 -142 Hz.

Vertical error bars in the following figures represent EMG/frequency standard deviations. Additionally, horizontal error bars represent torque standard deviations.

4.1 Ankle Plantarflexors

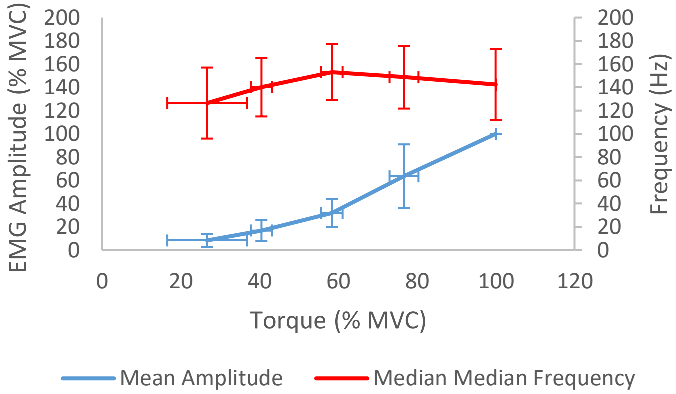

Figure 8a. Normalised EMG data with corresponding frequency at each contraction intensities for gastrocnemius medialis. Frequency range: 126-153 Hz. Data excluded: 20 per cent (n = 7), 40 per cent (n = 3), 60 per cent (n = 2), 80 per cent (n = 2), 100 per cent (n = 2).

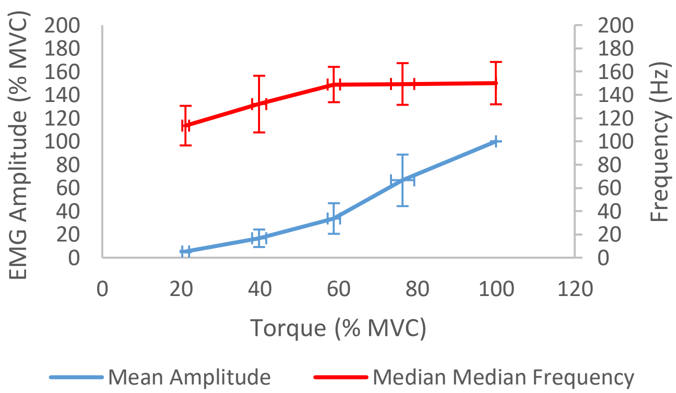

Figure 8b. Gastrocnemius medialis of participants grouped into moderate physical activity level category.

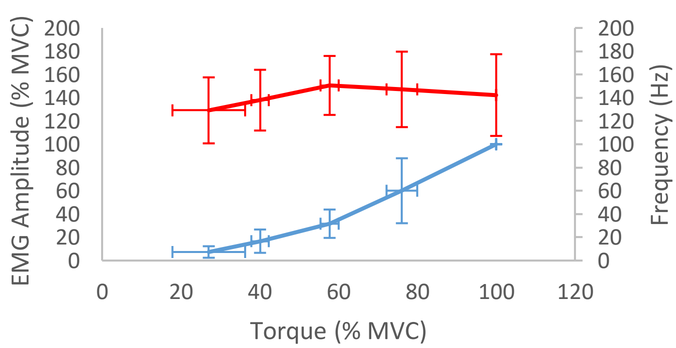

Figure 8c. Gastrocnemius medialis of participants grouped into high physical activity level category.

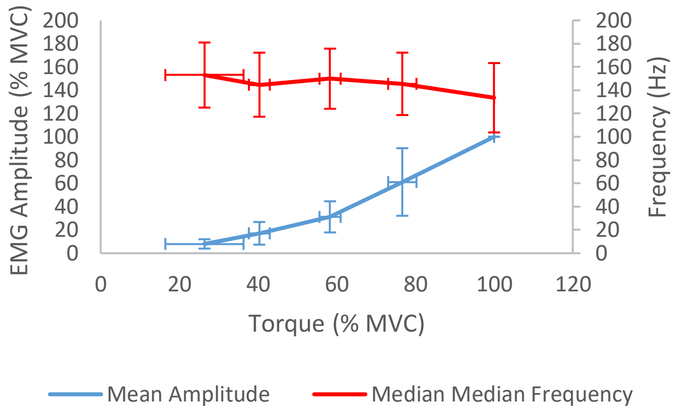

Figure 9a. Normalised EMG data with corresponding frequency at each contraction intensities for gastrocnemius lateralis. Frequency range: 133-153 Hz. Data excluded: 20 per cent (n = 6), 40 per cent (n = 2), 60 per cent (n = 1), 80 per cent (n = 1), 100 per cent (n = 1).

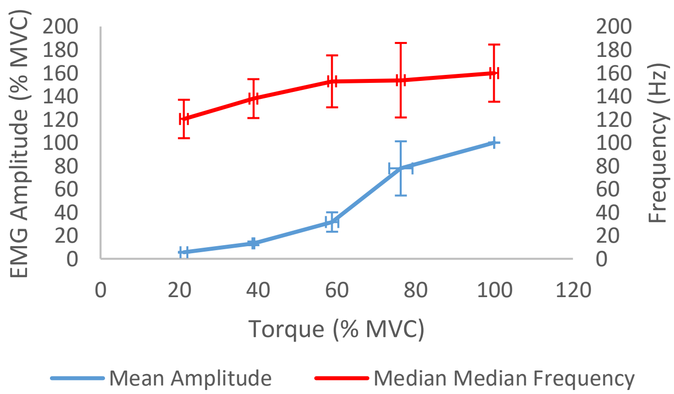

Figure 9b. Gastrocnemius lateralis of participants grouped into moderate physical activity level category.

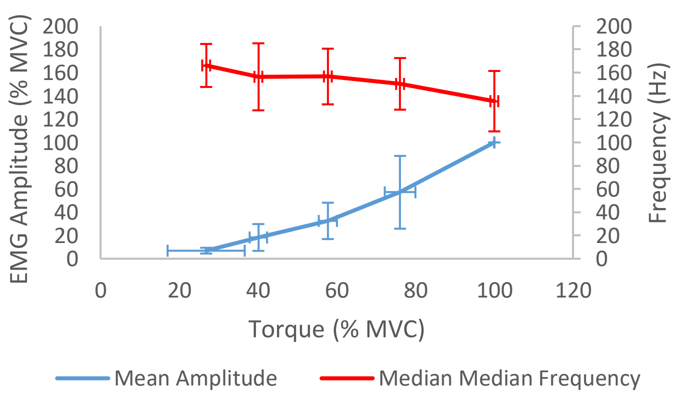

Figure 9c. Gastrocnemius lateralis of participants grouped into high physical activity level category.

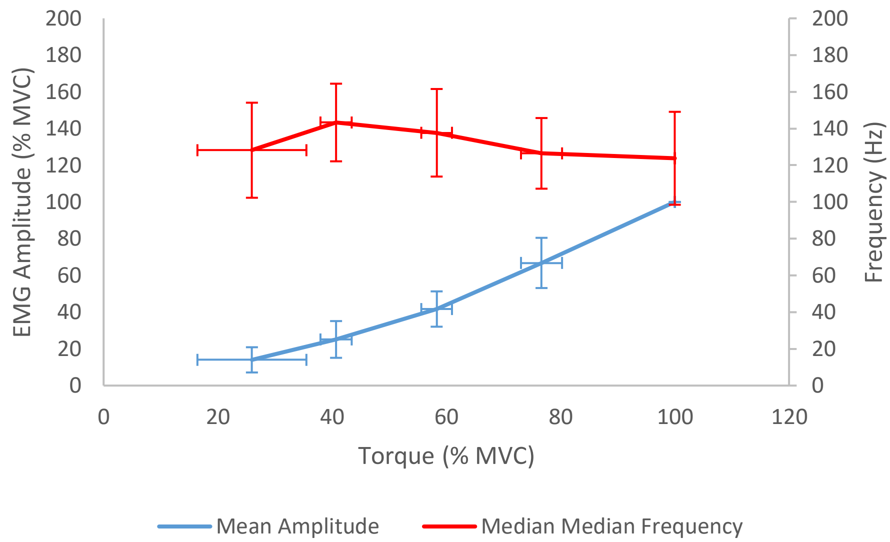

Figure 10. Normalised EMG data with corresponding frequency at each contraction intensities for soleus. Frequency range: 123-143 Hz. Data excluded: 20 per cent (n = 4), 40 per cent (n = 3), 60 per cent (n = 1), 80 per cent (n = 1), 100 per cent (n = 1).

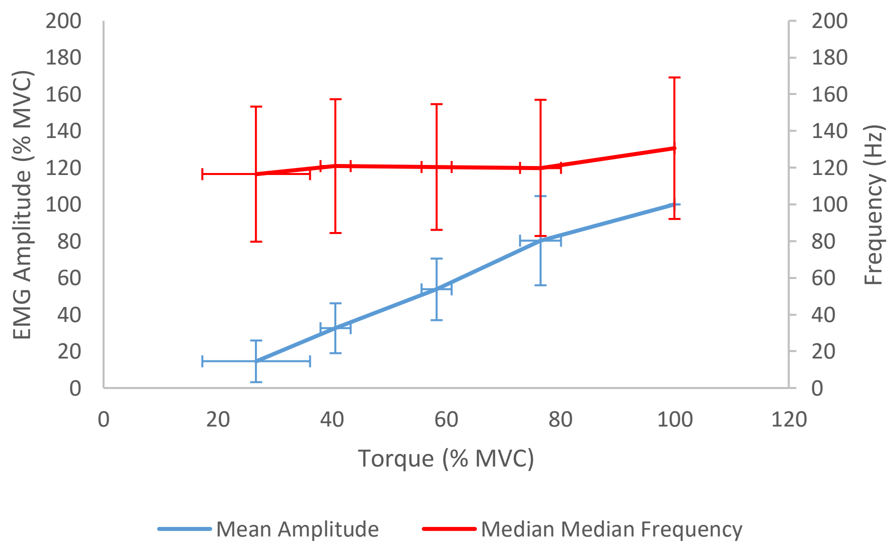

Figure 11. Normalised EMG data with corresponding frequency at each contraction intensities for fibularis longus. Frequency range: 116-130 Hz. No data excluded.

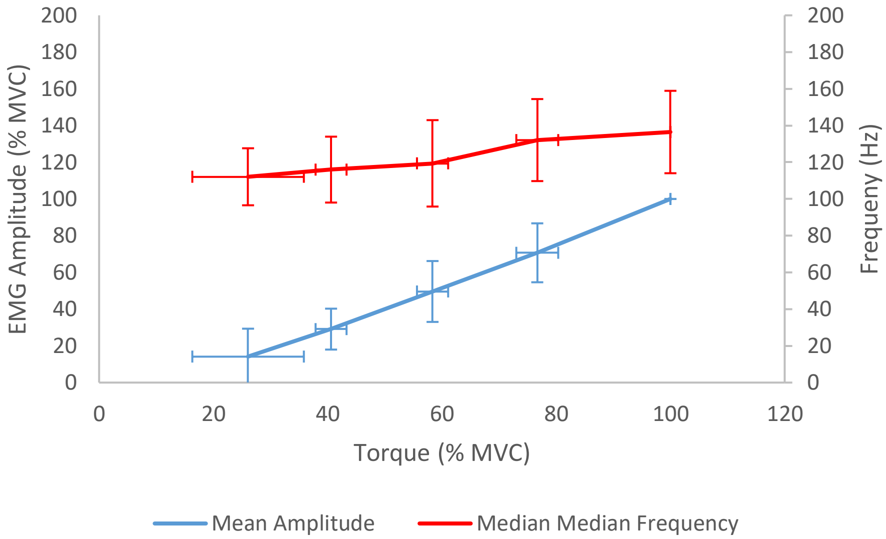

Figure 12. Normalised EMG data with corresponding frequency at each contraction intensities for fibularis brevis. Frequency range: 112-136 Hz. Data excluded: 20 per cent (n = 5), 40 per cent (n = 2), 60 per cent (n = 2), 80 per cent (n = 2), 100 per cent (n = 2).

4.2 <u>Ankle Dorsiflexors</u>

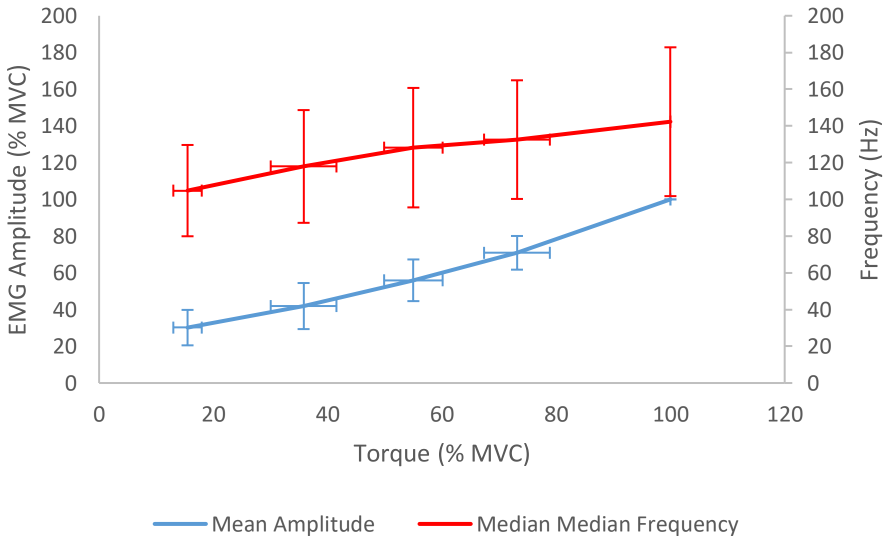

Figure 13. Normalised EMG data with corresponding frequency at each contraction intensities for tibialis anterior. Frequency range: 104-142 Hz. No data excluded.

The EMG-force relationships of TB Lateral and TB Long are both linear, as shown in Fig. 14 and Fig. 15, respectively. On the other hand, Fig. 16 and Fig. 17 show BB and BR to each possess an EMG-force relationship with an elbow point at 95 per cent of the MVC.

Regarding the power spectrum, the frequency of TB Lateral ranges from 81 – 87 Hz and TB Long from 69 – 74 Hz; both are relatively consistent. However, the frequency of BB ranges from 66 – 72 Hz and BR from 70 – 74 Hz. An elbow point is observed in the frequencies of both elbow flexors at 95 per cent.

4.1 <u>Elbow Extensors</u>

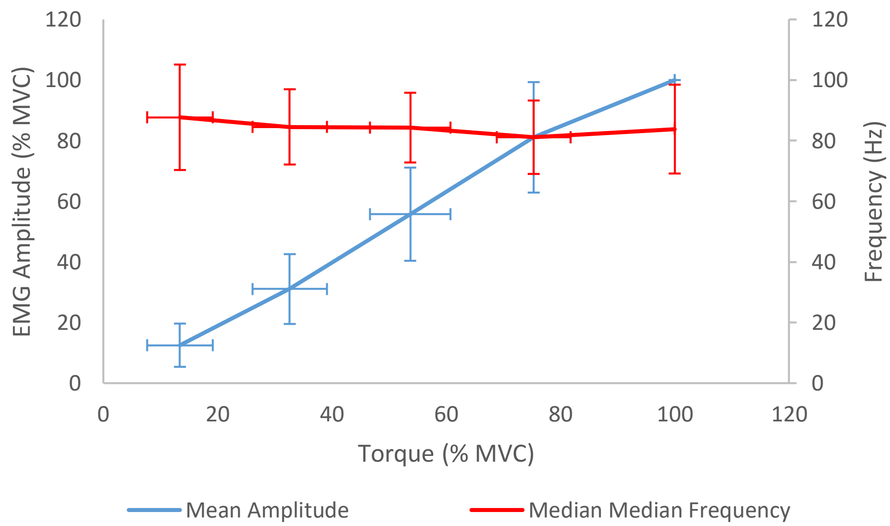

Figure 14. Normalised EMG data with corresponding frequency at each contraction intensities for triceps brachii – lateral head. Frequency range: 81-87 Hz. Data excluded: 20 per cent (n = 2), 40 per cent (n = 1), 60 per cent (n = 1), 80 per cent (n = 1), 100 per cent (n = 1).

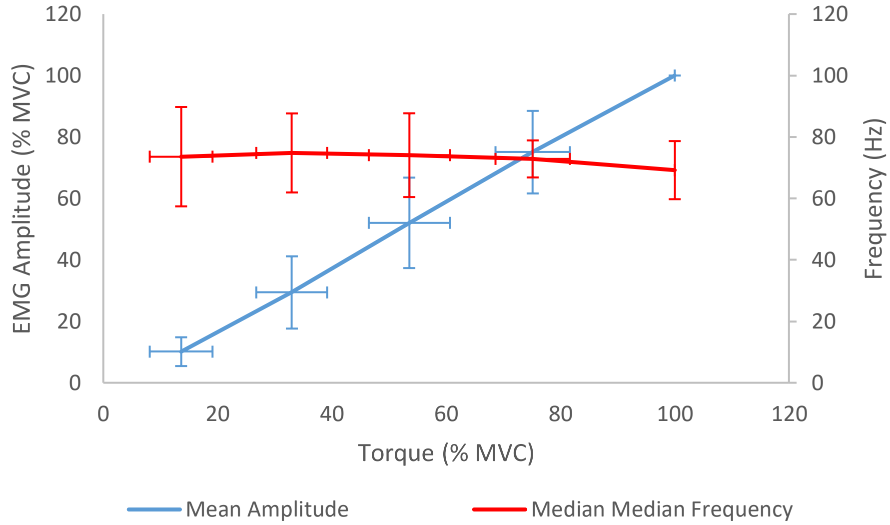

Figure 15. Normalised EMG data with corresponding frequency at each contraction intensities for triceps brachii – long head. Frequency range: 69-74 Hz. Data excluded: 20 per cent (n = 2), 40 per cent (n = 1), 60 per cent (n = 1), 80 per cent (n = 1), 100 per cent (n = 1).

4.2 Elbow Flexors

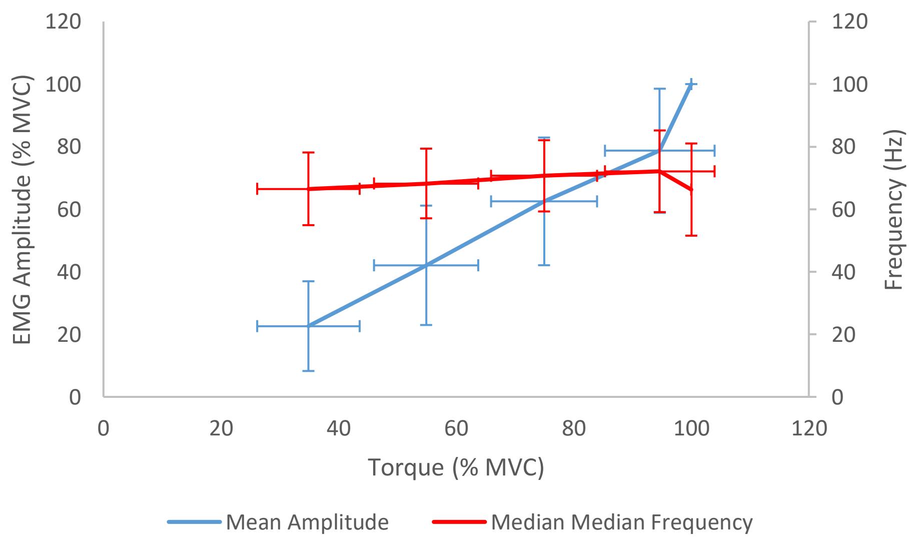

Figure 16. Normalised EMG data with corresponding frequency at each contraction intensities for biceps brachii. Frequency range: 66-72 Hz. Data excluded: 80 per cent (n = 1).

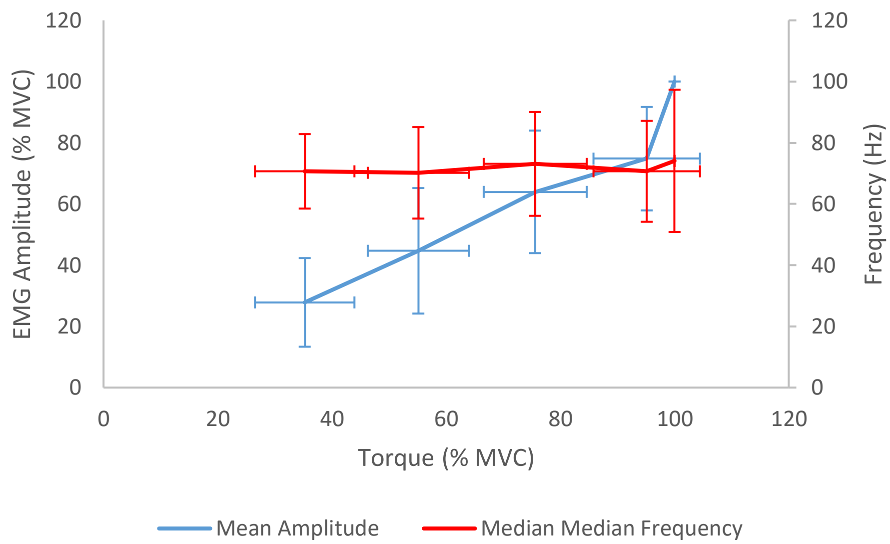

Figure 17. Normalised EMG data with corresponding frequency at each contraction intensities for brachioradialis. Frequency range: 70-74 Hz. Data excluded: 20 per cent (n = 3), 40 per cent (n = 3), 60 per cent (n = 3), 80 per cent (n = 4), 100 per cent (n = 3).

Fig. 18, Fig. 19, and Fig. 20 show the relationship of RF, VL and VM to be an elbow point at 80 per cent of the MVC, respectively. ST is shown to have a linear EMG-force relationship, as shown in Fig. 21, while Fig. 22 shows the relationship of BF to be curvilinear.

The frequencies of all three knee extensors have a similar range: RF is shown to range from 70 – 82 Hz, generally increasing as contraction intensity increases. Moreover, the frequency of VL is maintained between 65 – 69 Hz and VM between 64 – 70 Hz; at 60 per cent of the MVC the frequency begins to decline. On the power spectrum BF ranges between 77 – 90 Hz; there is an increase until 80 per cent of the MVC, where it then begins to decrease at the elbow point. On the other hand, ST ranges from 66 – 92 Hz, although continues to decline as contraction intensity increases.

4.1 Knee Extensors

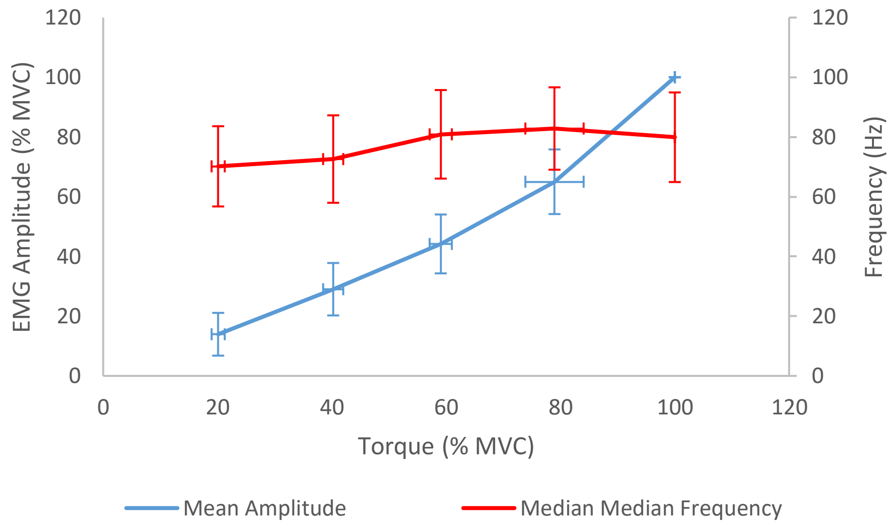

Figure 18. Normalised EMG data with corresponding frequency at each contraction intensities for rectus femoris. Frequency range: 70-82 Hz. No data excluded.

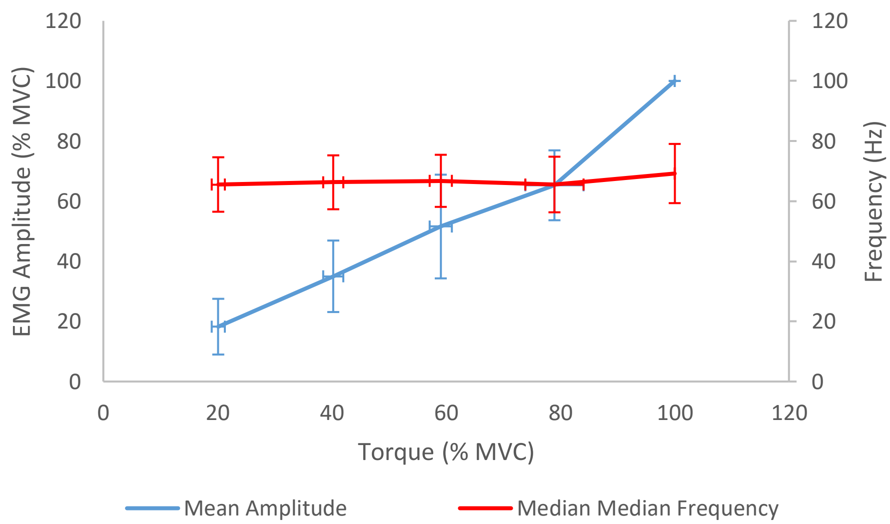

Figure 19. Normalised EMG data with corresponding frequency at each contraction intensities for vastus lateralis. Frequency range: 65-69 Hz. No data excluded.

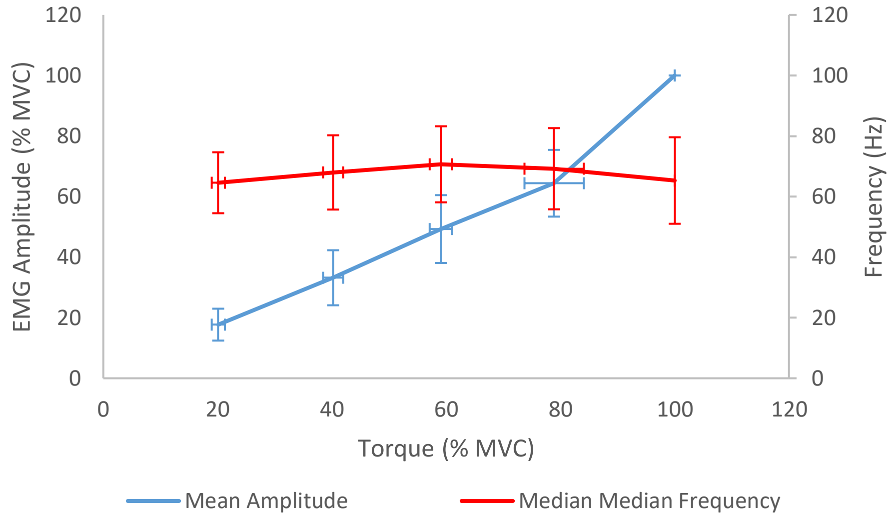

Figure 20. Normalised EMG data with corresponding frequency at each contraction intensities for vastus medialis. Frequency range: 64-70 Hz. Data excluded: 80 per cent (n = 1).

4.2 <u>Knee Flexors</u>

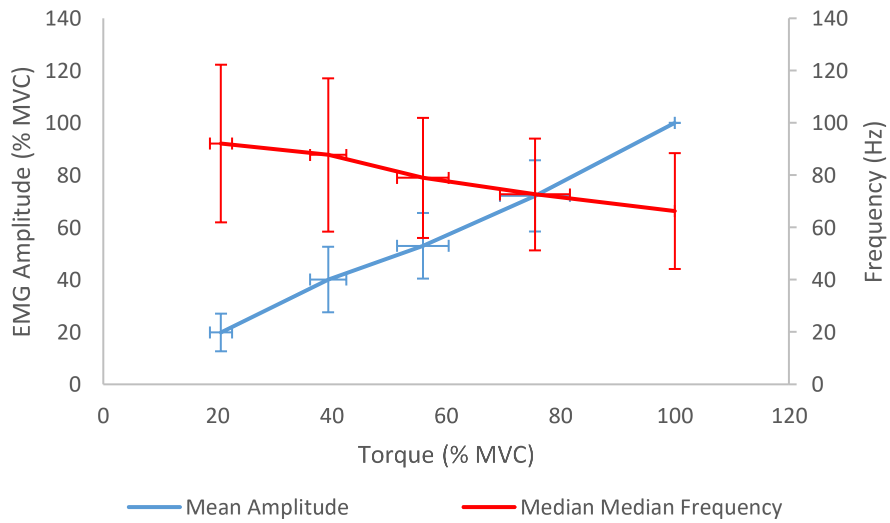

Figure 21. Normalised EMG data with corresponding frequency at each contraction intensities for semitendinosus. Frequency range: 66-92 Hz. No data excluded.

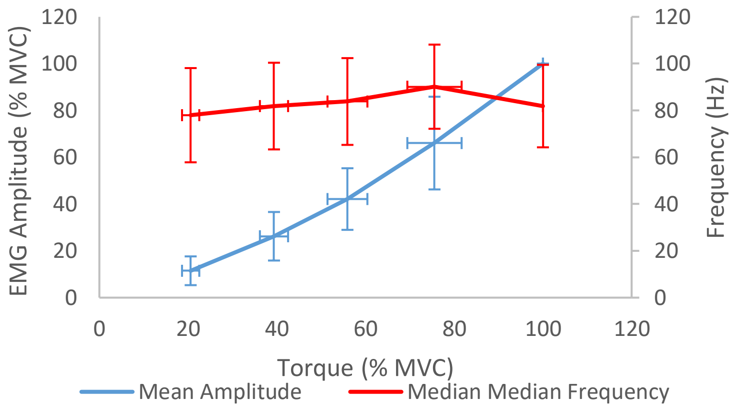

Figure 22a. Normalised EMG data with corresponding frequency at each contraction intensities for biceps femoris. Frequency range: 77-90 Hz. No data excluded.

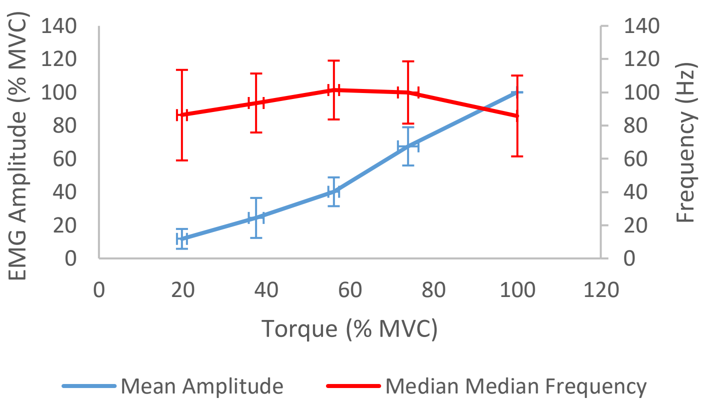

Figure 22b. Biceps femoris of participants grouped into moderate physical activity level category.

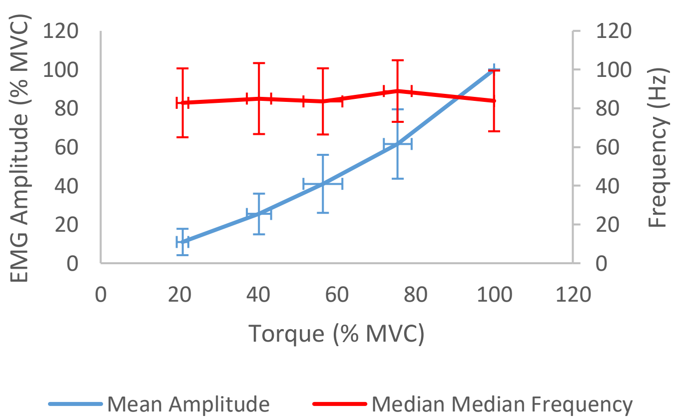

Figure 22c. Biceps femoris of participants grouped into high physical activity level category.

Table 5. Summary table of overall responses.

Joint	Muscle	EMG-Force Relationship Type		
		Linear	Curvilinear	Elbow Point
Ankle	Fibularis Brevis	✓		
	Fibularis Longus	✓		
	Gastrocnemius Lateralis			✓
	Gastrocnemius Medialis			✓
	Soleus		✓	
	Tibialis Anterior		✓	
Elbow	Biceps Brachii			✓
	Brachioradialis			✓
	Triceps Brachii - Lateral Head	✓		
	Triceps Brachii - Long Head	✓		
Knee	Biceps Femoris		✓	
	Rectus Femoris			✓
	Semitendinosus	✓		
	Vastus Lateralis			✓
	Vastus Medialis			✓
Total	15	5	3	7

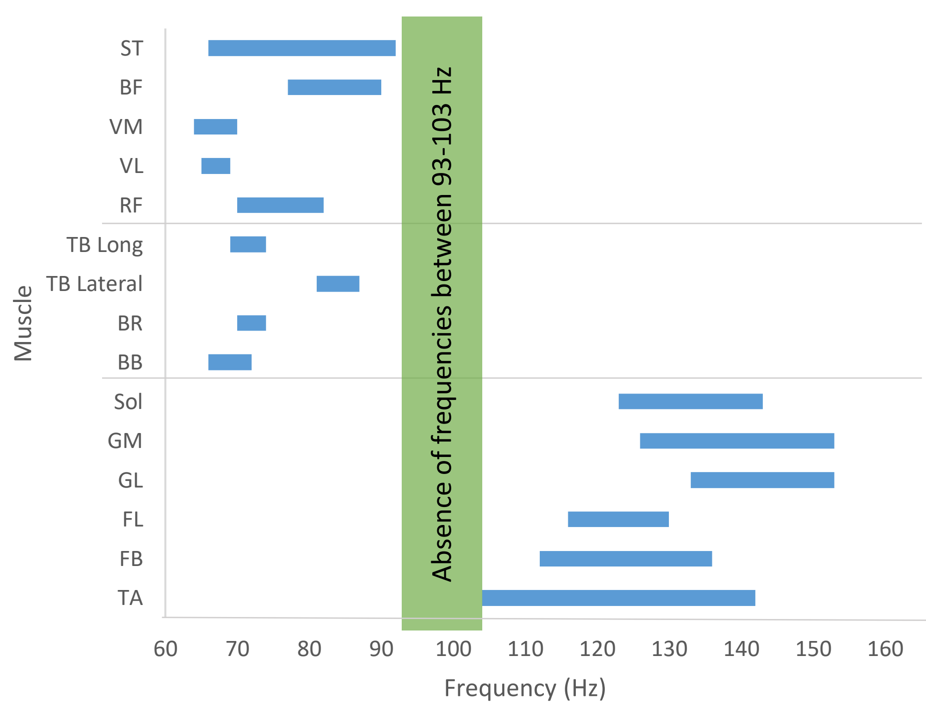

Figure 23. Frequency ranges of muscles.

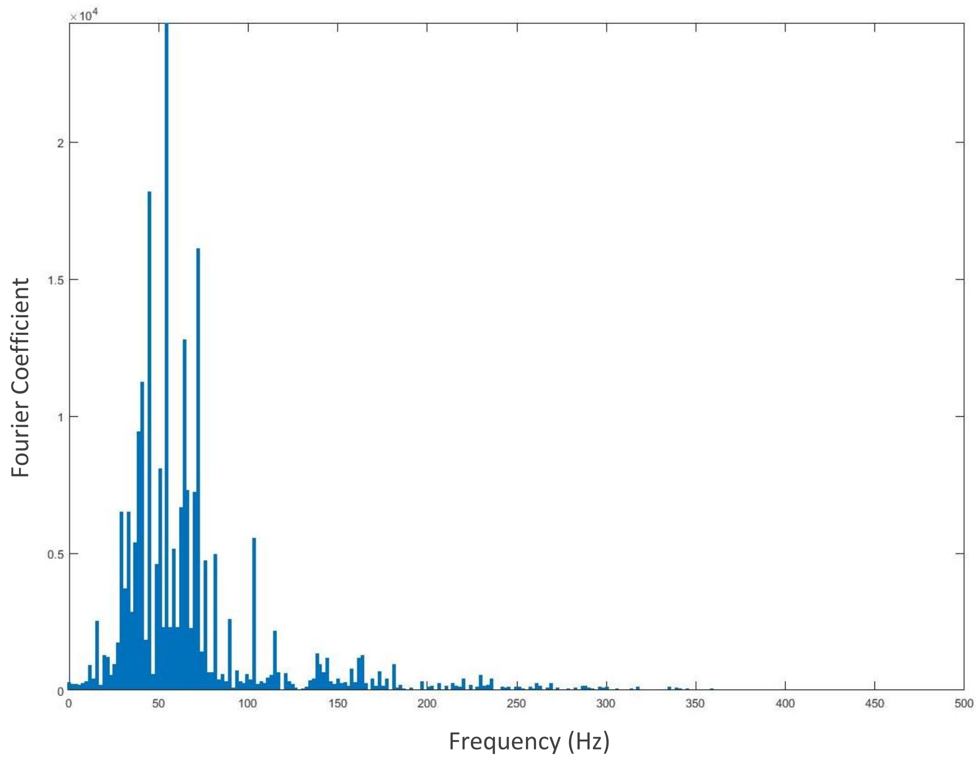

Figure 24a. Vastus lateralis frequency spectrum. 80 per cent of the MVC.

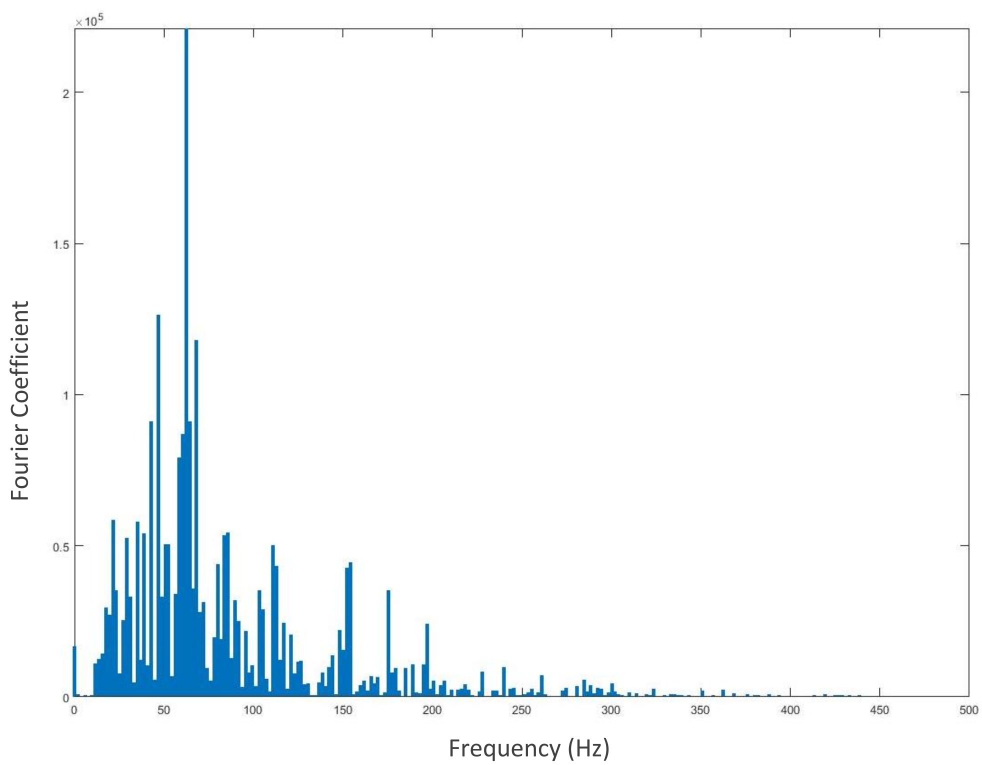

Figure 24b. Noise in the vastus lateralis frequency spectrum at 80 per cent of the MVC.

5. Discussion

The primary aim of this study was to investigate the relationship between muscle force and EMG amplitude. Key findings of the current study were that all muscles tested had extremely similar R^2 values, although upon visual inspection of individual graphs there were some relatively clear differences. These include FB, FL, ST, TB Lateral, and TB Long presenting with linear EMG-force relationships, BF, Sol, and TA presenting with curvilinear relationships, and BB, BR, GL, GM, RF, VL, and VM, presenting with elbow point relationships.

The EMG-force relationship has been known to vary between participants. The current study is no exception as muscles such as GM, GL, and BF had high standard deviations, particularly at 80 per cent of the MVC. The participant age range was between 19 – 35 years with an average age of 24.55 ± 4.64 years. High deviations may have been caused by the difference in physiology between those who were 19 years old and those who were 35 years old, for example. The high deviations may have also originated from the relatively even split between males (n = 18) and females (n = 13). Recruiting only either males or females may have shown a decrease in standard deviations. Additional reasons as to why these deviations are high have been considered further in the discussion.

Furthermore, linear relationships have been associated with smaller muscles and non-linear relationships with larger muscles (Basmajian and De Luca, 1985). This could be explained by rate coding and/or Henneman's Size Principle. In most instances the force and, in turn, amplitude deviations of each muscle are rather varied. The variation of force could be due to how well the participants could perceive their force production while amplitude variations may be dependent on joint angle, contraction velocity, and strength (Kuriki et al., 2012).

5.1 EMG-Force Relationships

5.1.1 Ankle Muscles

GM and GL are two heads of the triceps surae group, both presenting with an elbow point relationship at 60 per cent of the MVC as shown in Fig. 8a and Fig. 9a, respectively. The similarity between the two gastrocnemius heads comes as no surprise due to the relatively large number of commonalities which are discussed further on. Some characteristics are consistent throughout the population, such as both gastrocnemius heads converging at the achilles tendon to insert on the posterior surface of the calcaneus. The triceps surae group is completed with Sol constituting the third head.

While GM, GL, and Sol share the same insertion point, the EMG-force relationship of Sol does not present as an elbow point but is more curvilinear as shown in Fig. 10. These results disagree with previous research which found the EMG-force relationship of gastrocnemius and Sol to be linear (Bigland-Ritchie, 1981, Lippold, 1952). The participant demographics were not stated, and the methods used were seldom detailed in the study by Bigland-Ritchie (1981). Without this information, a conclusion cannot be made about how much age or training status influenced the results and cannot be compared to the current study. Lippold (1952) recruited 30 participants, although information such as age and training status were also not stated. Participants were in a seated position with the knee fixed at approximately 90º flexion. It is unclear as to which angle the ankle joint was fixed at. In the current study the knee was fixed at approximately 120º, which could provide an explanation as to why these results and those of Lippold (1952) differed.

FL and FB both initially present with a linear EMG-force relationship. as shown in Fig. 11 and Fig. 12, respectively. TA was the only dorsiflexor to be measured and has shown a slightly curvilinear EMG-force relationship, as shown in Fig. 13.

Participants had relatively poor force control at the lower intensities for plantarflexors; the standard deviation for each plantarflexor muscle at 20 per cent was, on average, ± 10 per cent. Although visual feedback on contraction intensity was provided to the participants, they were unable to finely control contractions at 20 per cent of their MVC. Plantarflexion is frequently performed throughout the day to achieve actions such as walking, running, and even jumping, thus the plantarflexor muscles must be powerful enough to keep up with the demands. As these muscles are demanding high force production it may be difficult to control the plantarflexors at lower forces.

Conversely, the standard deviation of TA at 20 per cent of the MVC was ± 2.5 per cent and increased to around ± 5.8 per cent as contraction intensity increased. The greater control at lower contraction intensities is a reasonable finding since dorsiflexion is a relatively weak action compared to plantarflexion. TA is not required to move the entire mass of an individual, unlike the triceps surae group. Participants may have had less control at higher intensities as it is somewhat unnatural to contract TA at 100 per cent, especially for everyday usage.

The amplitude deviation of both GL and GM increases as contraction intensity increases and is especially noticeable at 80 per cent of the MVC. The amplitude deviation of Sol, on the other hand, remains relatively consistent with increasing levels of force. Physical activity level was thought to be a potential cause of the high deviations. Participants had previously been categorised by level of activity (Table 2) so Fig. 8b, Fig. 8c, Fig. 9b, and Fig. 9c display the EMG-force relationships of the moderate and high levels. Low level activity was not included due to there being only two participants maximum in this category at one time. Despite grouping participants, the standard deviations of GL and GM of those with moderate and high activity levels are still relatively large and the elbow point remained unchanged. This suggests that physical activity level did not affect the EMG-force relationship. However, these indifferences could be explained by two reasons:

1. IPAQ does not distinguish between the type of activity an individual participated in. For example, one participant may primarily train the lower body while another may train their upper body more. This would affect fibre type ratio, thus the EMG-force relationship, of the upper and lower limbs. These variations are still grouped into the same category which could explain the large standard deviations.
2. The standard deviation was larger in the high physical activity group. This may be due to the sheer difference of number in each group; the high activity group contained 15 participants while the moderate activity group contained only 6 participants.

The R^2 value of GM was 0.965 moderate activity and 0.972 for high activity, while for GL these values were 0.941 for moderate activity and 0.969 for high activity (Table 4). These values are in range of remaining elbow point relationships.

Like the two gastrocnemius heads FL also shows an increasing deviation of amplitude with force, although FB and Sol have a similar trend of maintaining a regular deviation. This could have been caused by potential knee rotation differences between participants. Despite the increasing or consistent trends, all the plantarflexor amplitude deviations show some level of increase between 20 per cent and 80 per cent of the MVC. Unlike the plantarflexors, the amplitude deviation of TA initially slightly increases only to decrease as contraction intensity increases.

5.1.2 Elbow Muscles

TB Lateral and TB Long both have a linear relationship, as shown in Fig. 14 and Fig. 15, respectively. Furthermore, BB and BR both show an elbow point EMG-force relationship as shown in Fig. 16 and Fig. 17, respectively. Previous research has also reported the EMG-force relationship of BB and BR to be non-linear (Bigland-Ritchie, 1981; Zuniga and Simons, 1969). Similarly, TB has previously been found to have a non-linear relationship (Bigland-Ritchie, 1981), which disagrees with the results of the current study. As previously mentioned, the methods used by Bigland-Ritchie (1981) were seldom discussed. However, it was concluded that the differences between muscles with a linear and curvilinear relationship are likely due to physiological characteristics of the muscles, namely mixed fibre types instigating the non-linear relationships. The training status of the participants in the current study and in the Bigland-Ritchie (1981) study may have differed, which could have affected fibre type ratio, although this is indeterminate as these details were not provided.

Generally, participants were unable to accurately control the contraction intensity at every percentage of the MVC. On average, all contraction intensities indicated during data collection were performed 15 per cent more than required; 20 per cent was performed at 35 per cent, 40 per cent was performed at 55 per cent, etc. The staggered forces are the likely cause of the elbow point at 95 per cent; the force and EMG amplitudes are normalised and presented as percentages meaning MVCs are always 100 per cent. To support the hypothesis of participants having poor force perception for elbow flexion, standard deviations for BB and BR force were ± 9 per cent across all contraction intensities. On the contrary to BB and BR, the percentage of force exerted for the elbow extensors was shown to be between 5 - 7 per cent less than what was expected during data collection. This suggests that participants had relatively poor force perception when contracting muscles of the elbow.

The standard deviations of the TB Lateral and TB Long amplitudes generally increase with increasing levels of force. In contrast, the amplitudes deviations of BB and BR are relatively large, ranging from ± 14 per cent to ± 20 per cent with no apparent pattern as contraction intensity increases. The flexion angle of the elbow at its mid RoM caused the forearm to be in a near-vertical position. Being in this position, some participants may have isometrically supinated their forearm whilst flexing their elbow to maintain a good grip, while some participants may not have.

5.1.3 Knee Muscles

With regards to the knee extensors the EMG-force relationship of RF, VL, and VM show an elbow point relationship, as shown in Fig. 18, Fig. 19, and Fig. 20, respectively. The elbow point for both the VL and VM relationship is at 79 per cent of the MVC and was at 78 per cent for RF. Participants were able to accurately contract at the required contraction intensity which suggests that participants had excellent force perception at every level of contraction in comparison to the ankle and elbow muscles, especially at 20 per cent of the MVC. For all three knee extensors, the standard deviations for force as a percentage of the MVC from 20 - 60 per cent were ± 1-2 per cent. At the 80 per cent contraction intensity the standard deviation slightly increased to ± 5 per cent.

ST and BF present a curvilinear and linear EMG-force relationship, as shown in Fig. 21 and Fig. 22a, respectively. Contrasting the general consistency of the knee extensors, the standard deviations of the knee flexors gradually increased as contraction intensity increased; at 20 and 80 per cent contractions the deviation was ± 2 and ± 6 per cent, respectively. As with the knee extensors, this implies that participants had good force perception at lower intensities, although at 60 and 80 per cent of the MVC participants contracted at, on average, 5 per cent less than what was expected.

The standard deviation of the RF amplitude steadily increases as force increases from 20 – 80 per cent. On the other hand, the amplitude deviation of VL and VM increases until 60 per cent, where it then decreases and remains constant, respectively. With regards to the knee flexors the amplitude deviations increase as force increases, although ST remains unchanged at 40 per cent and 60 per cent. Like GM and GL, Fig. 22b and Fig. 22c show physical activity level categories of participants for BF. These results showed a similar outcome whereby the relationship still had a relatively large standard deviation at 80 per cent of the MVC. The coefficient of determination (R^2) of a linear trendline for moderate physical activity group was 0.981 and 0.976 for the high activity group (Table 4). Compared to R^2 value of 0.987 for BF shown in Table 1, these values are somewhat lower but still classify the relationship as curvilinear. These indifferences could, again, be due to the reasons as discussed for GM and GL previously.

5.2 Variables

Considering the amount of research that has been conducted investigating how force affects EMG amplitude, there has been little explanation for the variables and characteristics affecting the types of relationship that may present for individual muscles. There is a range of variables that may have affected the EMG-force relationship, such as fibre type ratio, nerve innervations, PA, PCSA, and more, which will be addressed and discussed ahead. Muscle groups with apparent trends, or a clear lack thereof, will be discussed in the pertinent sections.

In addition to the following characteristics discussed, blood supply, muscle thickness, muscle volume, and muscle innervations were additional characteristics reviewed. Neither of these appeared to influence the EMG-force relationship so were, therefore, omitted from this discussion.

5.2.1 Fascicle and Fibre Length

Muscle fibres arranged in bundles and surrounded by perimysium are known as muscle fascicles. *Fascicle length* is relatively easy to measure with ultrasound and Magnetic Resonance Imaging (MRI), so is often used over the term *fibre length.* Previous studies have measured either fascicle or fibre length but often not both regarding the same muscle.

Muscles with similar EMG-force relationships have been shown to possess dissimilar fascicle lengths between joints. For example, the fascicle length of GL has been reported to range from 56 – 74 mm and GM from 34 – 62.7 mm (Kawakami et al., 1998; Maganaris et al., 1998; Narici et al., 1996; Thom et al., 2007) while both having an elbow point relationship.

Contrary to this, the knee extensors have relatively similar fascicle lengths: RF can range from 69 – 134.9 mm, VL from 74.5 – 100.5 mm, and VM has been measured at 105 mm (Ando et al., 2014; Baroni et al., 2013; Blazevich et al., 2007; Erskine et al., 2009; Scanlon et al., 2013). Additionally, with regards to the elbow flexors, the fascicle lengths of BB have been reported to range from 141.3 – 160 mm and BR to be 192.1 mm (Kikuchi, 2010; Langenderfer et al., 2004).

Although each of the aforementioned muscles possess an elbow point EMG-force relationship the fascicle lengths differ quite substantially. It should also be noted that the elbow point of the relationships is at different percentages of the MVC for each joint, suggesting that fascicle length does has some influence on the EMG-force relationship.

A similar trend can be observed for muscles with a linear EMG-force relationship. For example, the fibre length of TB Long has been reported to be 10.86 mm (Lim et al., 2001). On the other hand, the length of fibres in ST has been shown to be 89.85 mm (Friederich and Brand, 1990). Finally, the fibre lengths of FB and FL have been recorded at 39.65 mm and 44.35 mm, respectively (Friederich and Brand, 1990).

Generally, muscles with the same action around a joint possess similar fascicle and fibre lengths but differ in comparison to muscles around other joints. Moreover, fascicle and fibre length naturally change as joint angle changes. Therefore, the length of muscle fascicles and fibres are directly affected by joint angle, which appears to have a large influence on the EMG-force relationship.

5.2.2 Fibre Type Ratio

The presence of one muscle fibre type over the other can be a suitable indicator of the purpose that muscles serve. Muscles that contain a higher percentage of type I fibres are better suited for activities that require endurance and force production, whilst a greater presence of type II fibres indicates a muscle is suited for rapid, explosive movements and shortening velocity (Kenney et al., 2015).

As previously stated, GM and GL presented similar EMG-force relationships while Sol differed only by the lack of an elbow point. The ratio of type I to type II fibres of GM and GL has previously been recorded at 50.8:49.2 and 43.5:56.5 for GM and GL, respectively, while Sol was shown found to have a fibre type ratio of 86.4:13.6 (Johnson et al., 1973). The findings of a second study coincide with this as the fibre type ratio of gastrocnemius was shown to be about 50:50 and Sol to be 70:30 (Edgerton et al., 1975). These findings suggest that the purpose of GM is a balance between force production and shortening velocity while the purpose of GL somewhat favours shortening velocity for rapid movements. Contrary to this, the predominance of type I fibres suggests the purpose of Sol is to facilitate endurance activities, which could explain the absence of an elbow point in the EMG-force relationship.

In addition to Sol containing predominantly type I muscle fibres, TA and FL were also found to share this trait (Johnson et al., 1973) suggesting their purpose is suited to force production and endurance. On the other hand, the purpose of BB, BR, TB, Lateral, TB Long, and RF appears to be shortening velocity as all were found to be comprised of predominantly type II fibres (Edgerton et al., 1975; Johnson et al., 1973). In some instances, studies have reported muscles such as BF, ST, VL, and VM to predominantly possess either type I or type II fibres (Edgerton, et al., 1975; Garrett et al., 1984; Johnson et al., 1973), suggesting that these muscles may be quite versatile.

Continuing from this, it should be noted that the EMG-force relationship of each of the muscles with a greater type I fibre percentage does not present with an elbow point. Interestingly, the muscle with a contrary fibre type predominance each include the presence of an elbow point relationship, except for TB Lateral and TB Long. This suggests that the EMG amplitude of muscles with a higher percentage of type II fibres suddenly increases at a greater rate as contraction intensity increases, noticeably between 60 - 80 per cent of the MVC. Moreover, the EMG amplitude of muscles with predominantly type I fibres appear to be relatively consistent with increasing contraction intensity.

5.2.3 Pennation Angle

The PA of a muscle is often considered a determining factor of force production and has been defined as the angle between the line of force and the alignment of muscle fascicles (Lee et al., 2015). A larger PA typically indicates a muscle's ability to produce greater force than those with a smaller PA (Fukunaga et al., 1992; Lieber and Friden, 2000).

As previously stated, GM and GL have presented with an elbow point EMG-force relationship while Sol is curvilinear. The PA of GM, GL, and Sol have been reported to be between 15.8° - 27.7° depending on joint angle, 10.9° - 18°, and 21° - 25°, respectively (Kawakami et al., 1998; Magnanaris et al., 1998; Morse et al., 2005; Narici et al., 1996). Although GM and GL have a slightly differing PA they possess a similar relationship. This suggests that a similar PA is unlikely to influence the EMG-force relationship by much. However, the somewhat larger PA of Sol may be a factor in the absence of an elbow point in its relationship. Like the gastrocnemius, RF, VL, and VM all possess an elbow point relationship while the PA of each has been shown to be between 6 – 12°, 9 - 12°, and 10 - 15°, respectively (Ando et al., 2014; Blazevich et al., 2007). On the other hand, research has also reported the PA of RF to be as large as 27.9° (Erskine et al., 2009), VL to be 17.2° (Baroni et al., 2013), and VM to be between 16.5 - 18° (Blazevich et al., 2007), indicating PA does not affect the relationship.

Like Sol, BF has also been shown to possess a relatively large PA of 20.94° (Kellis et al., 2010) and share a curvilinear relationship. This trend implies that curvilinear EMG-force relationships become more apparent the larger the PA is.

The current study has shown that FB and TA share a linear relationship. The PA of both FB and TA has previously been recorded as 12° (Friederich and Brand, 1990). On the other hand, TB Lateral and TB Long have also been shown to have a linear relationship, although their PA has both been recorded to be rather varying due to several factors. The PA of the TB heads has ranged from 5° - 55°, with greater muscle thickness associated with hypertrophy relating to a greater PA (Kawakami et al., 2006). Studies have found the PA of the TB heads to be 12.3° and 15° while others have reported the PA of TB Lateral and TB Long to be 26° and 12°, respectively (Gonzalez et al., 1996; Kawakami et al., 1994; Langenderfer et al., 2004).

With regards to research conducted on the PA of muscles, it appears it may have a slight influence on the EMG-force relationship. Despite the trends observed between muscles with similar PAs possessing similar relationships, some muscles with similar relationships have entirely different PAs. This suggests that other muscle characteristics have a much greater influence on the EMG-force relationship.

5.2.4 Physiological Cross-Sectional Area

PCSA is the area of a muscle and is measured at the largest cross section, perpendicularly to the muscle fibre orientations. PCSA is used over the anatomical cross-sectional area (ACSA) for pennate muscles as the ACSA measures perpendicularly to the longitudinal axis of a muscle instead of its fibres. Parallel muscles can also be measured with the PCSA as the fibres are aligned with the longitudinal axis.

A calculation for determining PCSA was established by Alexander and Vernon (1975) and is as follows:

$$PCSA = \frac{muscle\ mass}{\rho\ \cdot fibre\ length} \tag{5}$$

where ρ is muscle density.

PCSA is directly affected by the changes in PA and fibre length (Lieber and Friden, 2000).

The PCSA of GL, GM, VL, and VM has been shown to be similar despite crossing different joints; 24 cm^2 – 48.5 cm^2, 51 cm^2 – 68 cm^2, 71.1 cm^2, and 44.4 cm^2, respectively (Albracht et al., 2008; Erskine et al., 2009; Morse et al., 2005; Thom et al., 2007). A larger PCSA of a muscle has been associated with increased levels of training; Bamman et al. (2000) reported the PCSA of the untrained triceps surae to be 291.2 cm^2 while the trained to be 337.2 cm^2. It should be noted that as well as the PCSA of these muscles being similar, they all possess an elbow point EMG-force relationship.

Of all the muscles analysed in the current study it has been shown that Sol has greatest PCSA at 122.21 cm^2 – 131 cm^2 (Albracht et al., 2008; Friederich and Brand, 1990) and even as large as 230 cm^2 (Fukunaga et al., 1992). The EMG-force relationship of Sol is a curvilinear one, comparably to BF, although the PCSA of the latter has been reported to be much less at 18.23 cm^2 (Friederich and Brand, 1990).

In addition to this, the PCSA of FB, FL, ST, and TA are alike at 12.45 cm^2, 16.13 cm^2, 13.2 cm^2, and 12.68 cm^2, respectively (Friederich et al., 1990) and all have a linear EMG-force relationship. Furthermore, TB Lateral and TB Long have linear relationships despite having a PCSA of 2.81 cm^2 – 4.68 cm^2 and 2.6 cm^2 – 4.68 cm^2, respectively (Kawakami et al., 1994; Kikuchi, 2010; Langenderfer et al., 2004).

Overall, it appears PCSA may have a slight influence the EMG-force relationship as trends can be found between the two for some muscles. As PA and fibre length directly PCSA they may have attenuated the effect of PCSA on the EMG-force relationship.

5.3 Frequency Analysis

Muscle frequency, or power spectrum, has been the subject of research for some time, particularly when investigating the effects of fatigue on muscle performance. The amount of motor units recruited and the motor unit discharge rate, or rate coding, both influence the magnitude of activation and, consequently, muscle force (Duchateau and Baudry, 2014). A larger discharge rate means more motor units are recruited, equating to a faster and more powerful contraction and increasing frequency values.

The current study investigated the EMG-force relationship while aiming to reduce the effect of fatigue. Along with the effects of fatigue, muscle fibre type has additionally been shown to affect the power spectrum (Gerdle et al., 1991). Fig. 23 displays the frequency range of each muscle tested in the current study.

Fig. 8a, Fig. 9a, and Fig. 10 present the frequencies of GM, GL, and Sol to range from 126 – 153 Hz, 133 – 153 Hz, and 123 – 143 Hz, respectively. These results are somewhat in agreement with Bilodeau et al. (1994) who found the median power spectrum of GM to range from 100 - 150 Hz, GL from 95 – 120, and Sol from 80 – 110 Hz. Ramp and step contractions were performed by males and females, which were all compared and showed that sex influenced the power spectrum while the type of contraction did not. The knee angle used in the aforementioned study and the current study could explain the differences; Bilodeau et al. (1994) fixed the knee at 90° of flexion whereas knee flexion angle was greater than 90° in the current methods. Despite this difference, Sol should not have been affected by knee flexion angle. It should be noted that the power spectrum of GM and GL declines at the elbow point of the EMG-force relationship, likely due to the relatively even fibre type ratios. Moreover, this could indicate that fatigue may have potentially occurred at 60 per cent of the MVC.

In the present study the median power spectrum of TA ranges from 104 – 142 Hz, continuously increasing as force increases, as shown in Fig. 13. This finding agrees with Roy and De Luca (1989) who also found the median power spectrum of TA to be between 90 – 150 Hz. The ankle muscles having the greatest frequencies over the knee and ankle muscles is understandable as those of the ankle are tasked with locomotion and maintaining posture when standing. These actions require the muscles to be in constant use so a high discharge rate to recruit more motor units is necessary. Building on this, the triceps surae group may have higher frequencies than the remaining three as their action is plantarflexion, which raises the body so requires the plantarflexors to be powerful with high discharge rates.

The knee extensors have all shown to have a relatively similar median power spectrum; RF ranges from 70 – 82 Hz, VL from 65 – 69 Hz, and VM from 64 – 70 Hz as shown in Fig. 18, Fig.

19, and Fig. 20, respectively. This agrees with Bilodeau et al. (2003) as they compared males with females and used the statistical mean to analyse the power spectrum of the knee flexors to find similar results; 65 – 75 Hz for RF, 60 – 90 Hz for VL, and 70 – 90 Hz for VM. On the other hand, Pincivero et al. (2001) found the median frequency of RF, VM, and VL to range between 119 – 133 Hz, 97 – 116 Hz, and 132 – 162 Hz, respectively. This could be explained by the knee flexion angle being fixed at 60° whereas the average knee flexion angle used in the present study was 75.4°. The slight difference of results between the current study and that of Bilodeau et al. (2003) could be due to the difference in statistic used for reporting the frequency data.

The present study found the frequency of BF and ST 79 – 90 Hz and 66 – 92 Hz, respectively. This corresponds with Kellis and Katis (2008) who found the frequency of BF to be 80 – 120 Hz and ST to be 80 – 110 Hz during ramp contractions. Despite the similarity of range between the current study and Kellis and Katis (2008), ST differed as in the current study the frequency continuously declined as force increased. The inter-participant variability was also relatively high, especially at the lower contraction intensities. These occurrences may be due to two factors: ST could be easily fatigued, or the electrodes were placed occasionally directly over innervation zones.

In the current study the knee muscles have shown to exhibit lower frequencies than the ankle muscles. A potential explanation for this could be that, although both are used for locomotion, the knee muscles do not have to work to move the entire body mass.

The frequency of the elbow flexors has been shown to range from 66 – 72 Hz for BB and 70 – 74 Hz for BR, as shown in Fig. 16 and Fig 17, respectively. Previous research regarding BR has shown its median frequency to range from 65 – 75 Hz, which concurs with the current study (Doheny et al., 2008). Additionally, the findings of BB are in general agreement with previous research; the use of different masses caused the median frequency to range from 54 -74 Hz (Thongpanja et al., 2013) while a range of elbow flexion angles caused the median frequency of

BB to range from 46 – 67 Hz (Thongpanja et al., 2015). Moreover, joint angle has been shown to alter the median frequency of BB to range between 65 – 75 Hz (Doheny et al., 2008). The small difference of results between Thongpanja et al. (2015) and Doheny et al. (2008) may be attributed to the methods used; the former used five levels of isometric contraction from 1 – 5 kg, while it is assumed that the latter used only one level of isometric contraction.

Doheny et al. (2008) found the median frequency of triceps to range from about 80 – 90 Hz. The current study found the median frequency of TB Lateral to be 81 – 87 Hz and 69 – 74 Hz for TB Long, as shown in Fig. 14 and Fig. 15, respectively. Again, this difference between the two studies may be because of the use of various flexion angles used by Doheny et al. (2008). The elbow muscles exhibiting lower frequencies is also logical since they are not supporting the mass of the entire body or used for maintaining posture. Both heads of triceps brachii were shown to have higher frequencies than the elbow flexors. This could be due to firing frequency occurring before full motor unit recruitment, as concluded by Solomonow et al. (1989) to explain the linear EMG-force relationships. A non-linear relationship arises from the motor unit recruitment and firing frequency occurring concurrently (Solomonow et al., 1989).

In frequency analysis either the mean or median is commonly chosen as the preferred measure of central tendency. The mean provides a relatively accurate representation of the sample population, although is easily influenced by outliers in a skewed dataset. For this reason, the mean should be used for normally distributed data. Contrary to this, the median is not influenced by extreme outliers as the middle value of the dataset is used. Calculating the median value is more suitable to be used on skewed data over the mean.

During general data analysis, outliers may be present due to bad data or genuine extreme values. If a dataset contains outliers due to poor data, this data will be excluded to reduce to influence of contamination. Despite this, outliers may still be present and could affect the final results. For the current study, both the mean of the median frequency and median of the median frequency were calculated and presented similar results (Appendix 8).

5.4 Limitations

There may have been limitations which could have influenced some results. Training history, diet/nutritional supplements, sex, age, ethnicity, or body type may have had some effect on the results. Some extrinsic factors that may have affected the results include EMG sensor placements possibly not being in ideal locations, such as near to an innervation zone.

The isokinetic dynamometer itself can be identified as a limitation when considering ankle torque measurements. Before collecting data from participants, it was recommended that gravity correction of the dynamometer attachments was completed to give more accurate results. Although this was possible when using the elbow and knee attachments, it was not possible for the ankle due to how the machine was manufactured. Because of this the on-screen force bar used to guide the participants would sometimes be greater than the value they were asked to contract at. This meant that they either did not contract their plantarflexors or slightly contracted but potentially provided inaccurate data. Slightly dorsiflexing before plantarflexing occasionally remedied this issue but not in all cases. With regards to the EMG equipment, the sEMG electrodes were unable to target specific motor units. This means cross-talk from neighbouring muscles may have interfered with the signal, particularly for smaller muscles such as BR. The use of indwelling electrodes may have given more been more reliable readings as these can be positioned more accurately.

When applying force participants were seated with their hip and knee flexed (Appendix 4) to isometrically contract the antagonistic muscle pairs of the ankle. The ankle muscles are primarily used to plantarflex and dorsiflex during locomotion when standing upright. There may be instances where this seated position is relatable to exercises such as the leg press, although this works the knee extensors, so is not relevant to the ankle. The setup of the ankle position on the isokinetic dynamometer is not functional for everyday use and may have produced differing results if the equipment permitted a more suitable position.

With regards to joint positioning, due to the angle the knee was required to be at the GL and GM may have been at a disadvantage over Sol. The gastrocnemius is a biarticular muscle crossing both the knee and ankle. Because of this it is important that the length-tension relationship is considered to understand how knee angle is a factor affecting force production and EMG amplitude. It has previously been shown that as knee flexion angle increases, force production and EMG amplitude decreased for both the medial and lateral heads of gastrocnemius, whilst increasing for soleus (Cresswell et al., 1995). Cresswell et al. (1995) also found that the lateral head of gastrocnemius was affected less by flexion angle than the medial head. This may have been due to lateral head's pennation angle being smaller than medial head's (Fukunaga et al., 1992; Kawakami et al., 1998), thus being more aligned to the line of force.

Following this, the knee flexors and extensors have a similar issue due to them also being biarticular, crossing the knee and hip. When calculating the midrange of knee flexion for each participant hip angle was not taken into consideration. As the knee flexion mid-range was being fixed into place it may have been more accurate to position the hip at the same angle, forming a Z-angle. This would have ensured that the length of the knee extensors was similar to the flexors as the origins and insertions distance would have been mirrored.

Furthermore, as only one fixed position was used for the duration of testing synergist muscles were grouped and included in the overall movements. For example, the primary purpose of the fibularis muscles is to evert the foot, while only assisting with plantarflexion. If eversion was tested alongside plantarflexion the results may present an alternate EMG-force relationship. Similarly, during knee flexion one position was used to measure BF and ST; hip rotation may have benefitted the relevant muscle and produced a different relationship. Relating to this point, tension impedance, particularly for BF and ST, could have affected force production as pressure on one point of a muscle may have resulted in a reduction of tension over different areas. A standing position may have resolved this potential issue over a seated one, although the isokinetic dynamometer used was unable to accommodate for this.

In section 3.4 the presence of noise and the actions taken to reduce it were detailed. The results of the noise reduction in the time domain can be seen in Appendix 9. Fig. 24a shows the frequency spectrum of vastus lateralis for a participant in the current study. It was noticed that, although the noise appeared to be reduced in the time domain, noise still appeared to be affecting the frequency domain. Comparing to Fig. 24a, Fig. 24b shows frequencies on the lower end of the spectrum with greater spikes than typically expected. This indicates noise at the lower frequencies which may have potentially lowered the overall median frequency. It is possible that applying a notch filter to remove these frequencies would have led to an increase in median frequency.

6. Conclusion

The results of the current study have shown that people tend to under- or overestimate their perception of contraction intensity; the elbow muscle results particularly highlight this. In the current study participants often could not consistently maintain their force around the intensity required of them, even with the use of visual guidelines. In a clinical environment this would be more apparent as the client would be merely assuming their perception of contraction intensity to be correct since visual cues may be not available to guide them. Exceeding the required force could, potentially, cause further injury to the client while under exerting could diminish the effectiveness of the treatment.

After reviewing a number of previous studies, characteristics such as nerve innervation, blood supply, appear to have little-to-no influence on the relationship. On the other hand, muscle fibre type and joint angle appear to be the two major factors contributing to changes in the EMG-force relationship. On several occasions it has been shown that joint angle affects force production, thus EMG amplitude. With regards to the median frequency of muscles, the results from the current study generally agree with previous research about the ranges of median frequencies for each muscle in question. It has been shown that sex, fibre type, joint angle and, in turn, muscle length largely contribute to the differences observed.

Future studies may find a range of results if the focus was investigating how the EMG-force relationship differs with individuals of varying training statuses. Someone who lives a sedentary lifestyle may present with different EMG-force relationships for the muscles in question than those who participate in vigorous physical activities, such as digging, heavy lifting, or aerobics. Likewise, those who take part in moderate level physical activities, such as light cycling, carrying light loads, or playing a team sport may find their EMG-force relationships differ from those who do vigorous physical activities. This could also be applied to people who participate in different sports and at different levels. For example, comparing those who play football against those who play badminton, or only those who play football at elite against semi-professional players.

The use of various joint angles could also be expanded upon as this will alter the muscle length, thus force production. If equipment permits, the EMG-force relationship of muscles during dynamic movements could be analysed as the joint angle will be consistently changing, so is more applicable to functional usage. Otherwise, a series of maximal isometric contractions at various joint angles could be measured to simulate a dynamic movement.

The body position of the participants while on the isokinetic dynamometer is another aspect that could be reviewed for future research. For example, as previously noted, a limitation of the current study was that the ankle joint was testing in a non-functional position; people seldom plantarflex nor dorsiflex in a seated position with their lower leg parallel to the floor (Appendix 4). Instead, an isokinetic dynamometer with the ability to allow these movements in a more functional position, such as standing or standard sitting, should be sought after.

With regards to the knowledge of sEMG, one issue that can be brought to attention from the current study is that grouping data together may disguise underlying patterns within the data. Grouping participants into categories may provide a clearer insight into what could be influencing the EMG-force relationship. Using further sub-categories for type of activity could prove useful at a later time.

7. References

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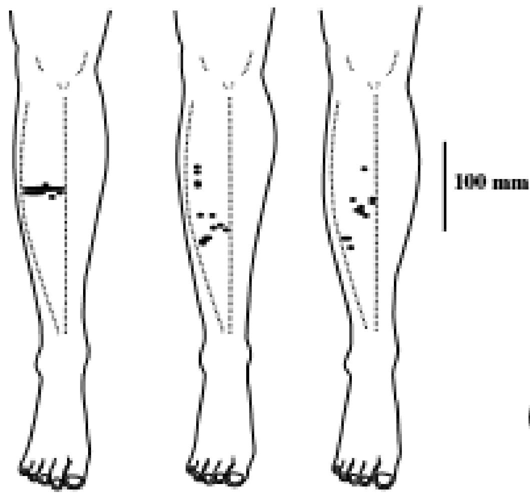
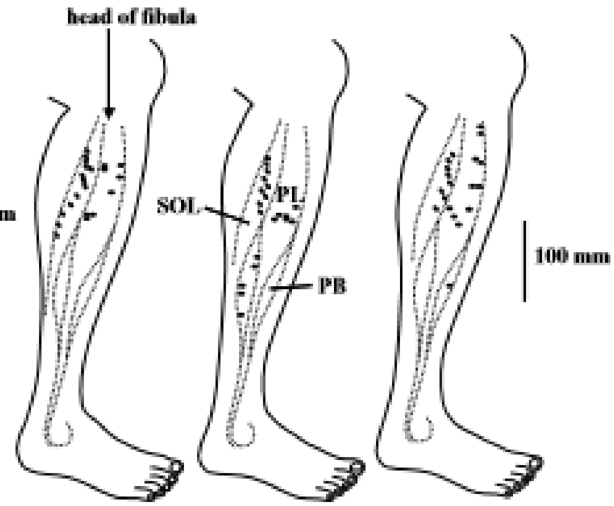
8. **Appendices**

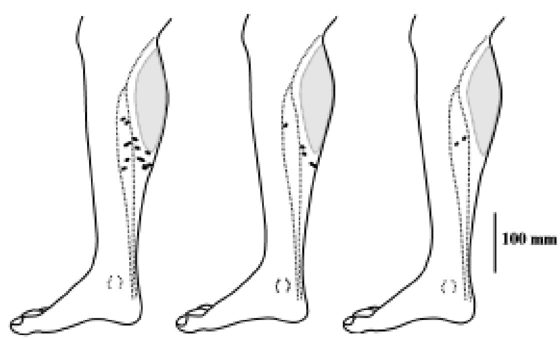
Appendix 1.1 – Distribution of Innervation Zones in the Ankle Muscles



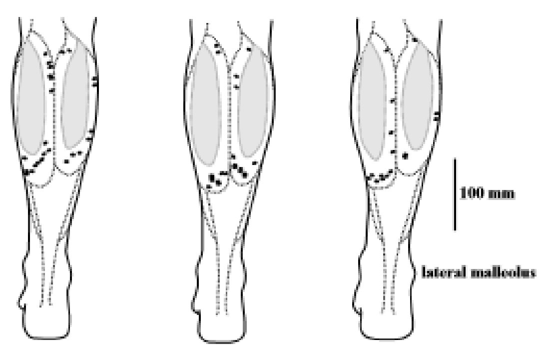
Tibialis anterior



Soleus, Fibularis longus, Fibularis brevis

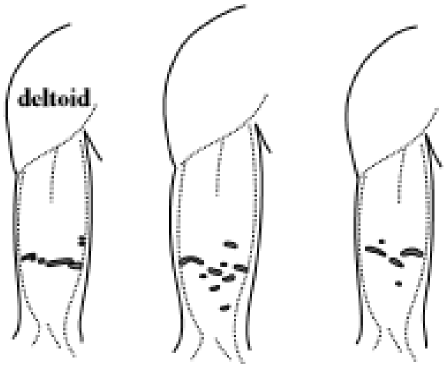


Gastrocnemius medialis

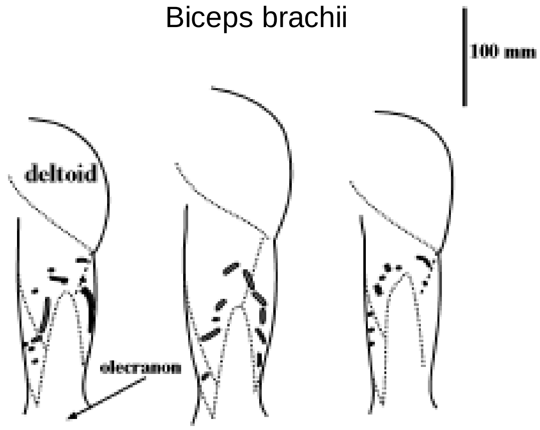


Gastrocnemius medialis and Gastrocnemius lateralis

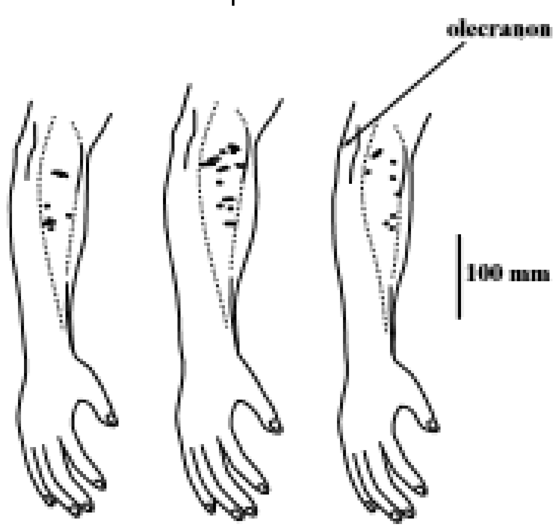
Appendix 1.2 - Distribution of Innervation Zones in the Elbow Muscles



Biceps brachii

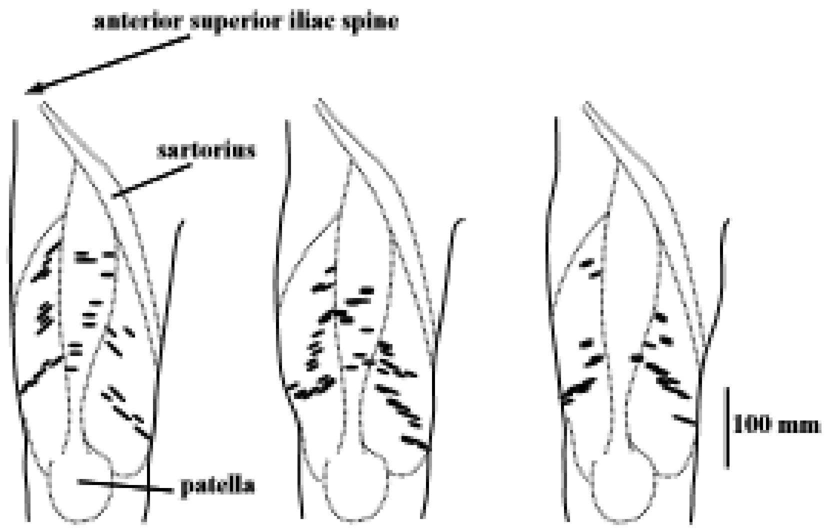


Triceps brachii

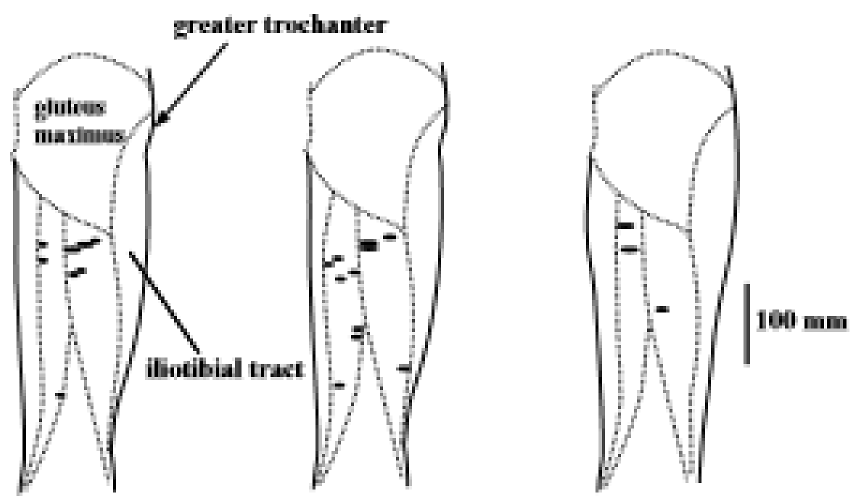


Brachioradialis

Appendix 1.3 – Distribution of Innervation Zones in the Knee Muscles

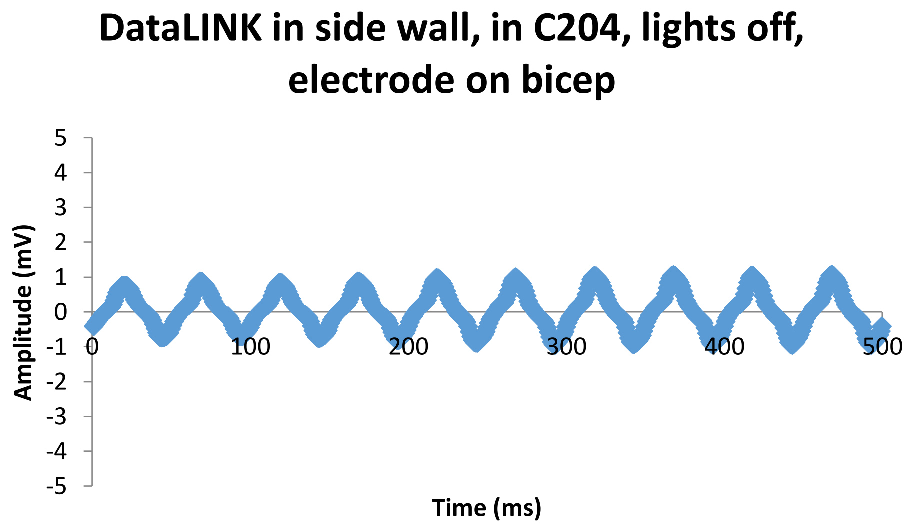


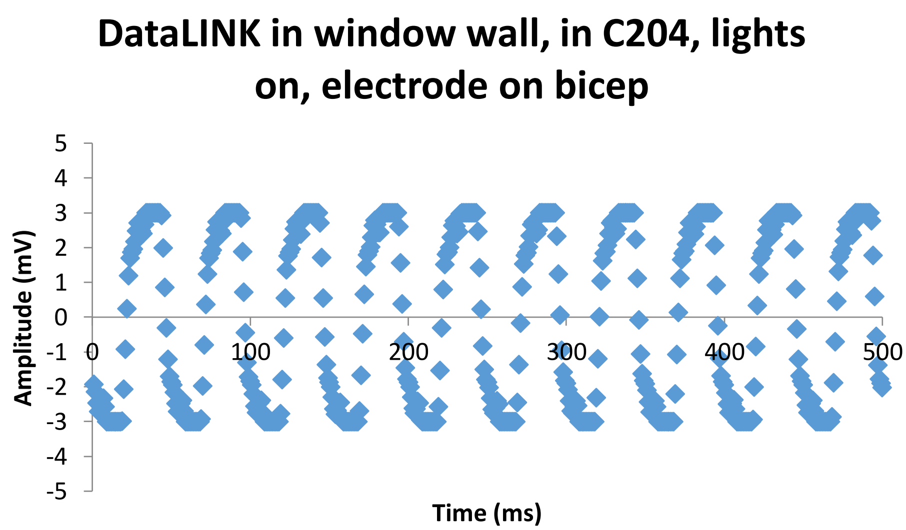
Vastus lateralis, Rectus femoris, Vastus medialis



Semitendinosus and Biceps femoris

Appendix 2 - Noise Test Graph Examples





Appendix 3 – Validation Testing Results

Joint	Angles (°)				Peak Torques (Nm)					
	RoM	Midpoint	Mid*0.5	Mid*1.5	At Midpoint		Mid * 0.5		Mid * 1.5	
					Ext/Plantar	Flex/Dorsi	Ext/Plantar	Flex/Dorsi	Ext/Plantar	Flex/Dorsi
Ankle	75	37.5	18.75	56.25	37	27	50	22	29	27
Knee	160	80	40	120	138	66	108	19	133	73
Elbow	150	75	37.5	112.5	30	27	22	24	35	19

Appendix 4 – Isokinetic dynamometer setup for ankle joint testing

Appendix 5 – Isokinetic dynamometer setup for elbow joint testing

Appendix 6 – Isokinetic dynamometer setup for knee joint testing

Appendix 7 – MATLAB Code for Frequency Analysis

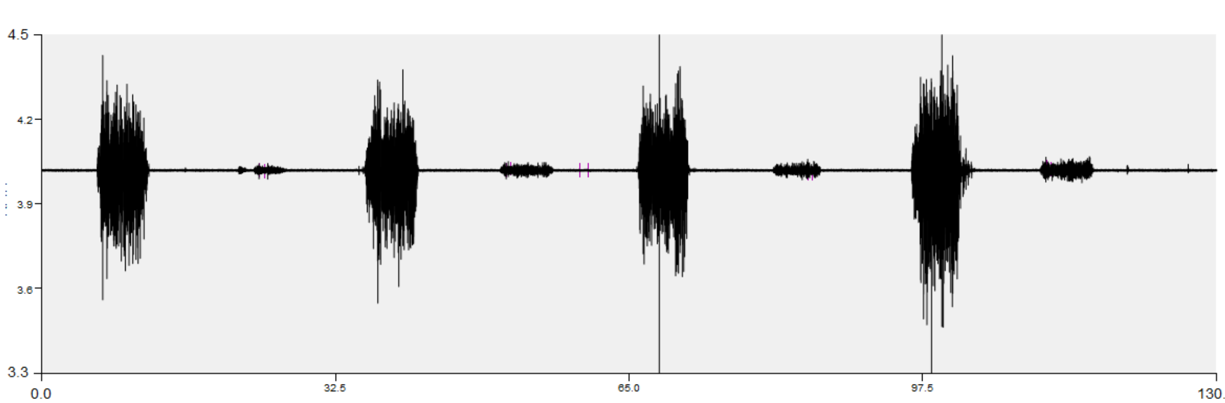
```
%% Analysis of Ankle EMG Data
%% Loading and Checking the GL Data
% The first section of the code loads the ankle data files and checks that
% there is the same number of contractions in each file
Num_contractions_GL = zeros(5,1);
  for i = 1:5; %This sets up an iteration (1,2,3,4,5)
    filename=['Ankle_GL_',num2str(i*20),'.txt'];
    %The above line uses the iteration to generate a name which changes
    %in the format of the data filenames.
    Ankle = dlmread(filename,'',5,1); %This line loads the datafiles,
    % ignoring the first 5 rows and the first column,
    Ankle = Ankle(1:1024,:); %The original files have 1025 data points,
    %This line of code reduces each column to 1024 data points.
    Num_contractions_GL(i)=length(Ankle(1,:)); %This command counts the
    %amount of columns in each file - this represents the amount of
    %contractions exported from Visual3d
      %Note that the use of (i) after the variable name tells MatLab
      %to create a vector, rather than simply keep saving the numbers
      %as a single variable overwriting itself.
  end

%%
%%Calculating the Median Frequencies
%MatLab does not like dealing with data which is not in a matrix format.
%Therefore, step one is to create an empty array which data can be added to.
%This process helps create code, plus makes code run faster. It is called
%'pre-allocation'.
AnkleGL = zeros(5,max(Num_contractions_GL));
%The above code will generate an array of '0's with 5 rows, the number of
%columns will be set by the maximum number of contractions detected in the
%raw EMG files exported from Visual3D.
  for i = 1:5; %This sets up an iteration (1,2,3,4,5)
    filename=['Ankle_GL_',num2str(i*20),'.txt'];
    %The above line uses the iteration to generate a name which changes
    %in the format of the data filenames.
    Ankle = dlmread(filename,'',5,1); %This line loads the datafiles,
    % ignoring the first 5 rows and the first column,
    Ankle = Ankle(1:1024,:); %The original files have 1025 data points,
    %This line of code reduces each column to 1024 data points.
    for j=1:length(Ankle(1,:)); %This code sets up an iteration, within
      %the above iteration, to cycle through the imported columns.
      [frqscale, power] = drFFT2(Ankle(:,j),2000);
      AnkleGL(i,j)=medFFT(frqscale,power); % once the data has been
      % loaded into MatLab, the code runs through each column
      % transforming the data into the frequency domain and
      % calculating the median frequency (MF). The MF is then saved
      % in to the pre-allocated array.
    end
  end
xlswrite('Ankle',AnkleGL,'GL') % This command saves the median frequencies
%in an excel files named 'ankle' creating a tab called 'GL'. As further
%code is developed it is therefore possible to add other tabs and reduce
%the number of spread sheets generated.
```

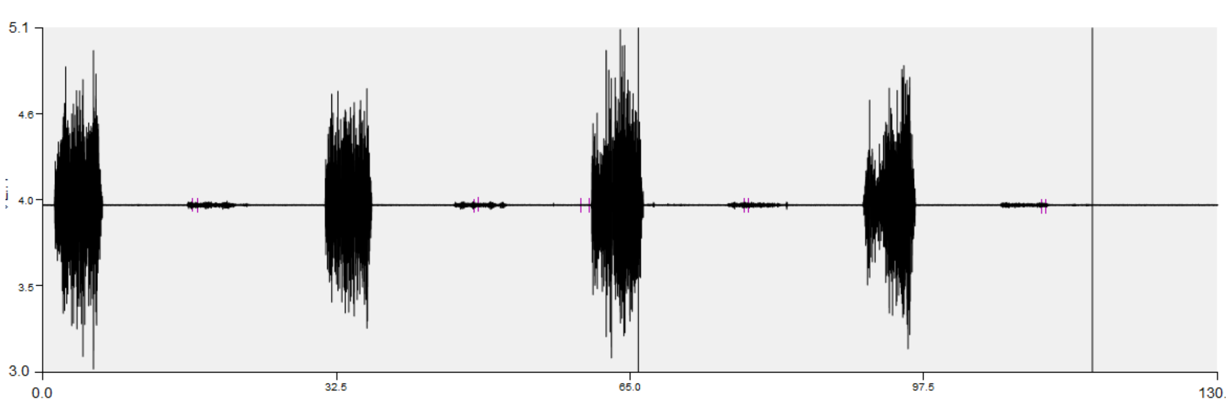
Appendix 8 – Percentage of the Mean of Median Frequency in Median of the Median Frequency (Hz)

Joint	Muscle	Mean of Median Frequency (Hz)	Median of Median Frequency (Hz)	% of Median Frequency
Ankle	Fibularis Brevis	122.62	123.20	99.53
	Fibularis Longus	126.61	121.64	104.09
	Gastrocnemius Lateralis	144.3	145.32	99.3
	Gastrocnemius Medialis	143.98	142.11	101.32
	Soleus	128.6	131.87	97.52
	Tibialis Anterior	118.53	125.15	94.72
Elbow	Biceps Brachii	71.38	68.76	103.81
	Brachioradialis	74.81	71.73	104.28
	Triceps Brachii - Lateral Head	84.23	84.31	99.90
	Triceps Brachii - Long Head	73.41	72.90	100.69
Knee	Biceps Femoris	84.88	83.14	102.10
	Rectus Femoris	77.43	77.29	100.18
	Semitendinosus	84.79	79.53	106.61
	Vastus Lateralis	67.23	66.67	100.84
	Vastus Medialis	70.46	67.54	104.31

Appendix 9 – Raw EMG Traces During a Set of Contractions



Raw EMG trace of rectus femoris at 80 per cent of the MVC.



Raw EMG trace of vastus lateralis at 80 per cent of the MVC.